Demonstrating Altered Kinematics from Ligament Injuries with a Three Dimensional Knee Model

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Abstract: A three dimensional physical model of the human knee joint was created in this project. The design enhances previous two dimensional models but its main focus is to simulate altered kinematics from cruciate ligament failure. The model accurately simulates the rolling to sliding ratio within a healthy knee as well as accurate internal rotation simulation as a result of flexion. The model can also be altered to display cruciate ligament injuries ranging from the most severe, Grade III, to slight sprains in the ligament, Grade I. There are several ways in which a physician checks for ligament damage. These injuries can be displayed by implementing common ligament damage tests like the anterior draw test or posterior draw tests. The injurious examples yield increased laxities within the knee which is displayed by an increase in travel of the tibia and femur relative to each other when a force is applied during these tests. For example, increasing the severity of damage to the anterior cruciate ligament in the model was found to cause increasing femoral slip in the posterior direction relative to the tibia, as is observed clinically in an anterior draw test. The end result is an interactive model that can be altered quickly to display both knee kinematics of a healthy and injurious knee.

INTRODUCTION

The motivation for focusing on the kinematics of the knee joint is due to the large number of knee injuries suffered each year by people. In 2003, about 19.4 million people visited the physician's office due to knee issues. The leading reason why people visited the orthopedic surgeon that year was also a result of knee injuries. [15] Seventy percent of the people attending the orthopedic surgeon with knee problems were there as a result of cruciate ligament issues. Since the instatement of Title 9 which allows an equal number of female athletic scholarships as men's, there has been an increase in anterior cruciate ligament (ACL) tears and surgeries do to the fact that female athletes are more susceptible to this type of injury. ACL and posterior cruciate ligament (PCL) injuries have become a major issue in professional sports as well. Last season two multi-million dollar players, Dante Culpepper and Carson Palmer, were sidelined during the NFL season and an entire country wept as the star player of the English National Team, Michael Owen, missed the World Cup as a result of an ACL tear. All of these players received large salaries and their livelihood was jeopardized by these injuries. These players are all prime examples of where cruciate ligament damage is not only a health concern but also a business concern when dealing with superstar athletes.

In this paper, the healthy kinematics and functions of the knee and cruciate ligaments of which the new model is then governed by are discussed. Also, several common ways in which the ligament can fail and ways in which physicians test for ligament damage are introduced. The design of the model is then examined and then the outcome and results of the new model are discussed.

MATERIALS & METHODS

Kinematics

Definition of Axes and Loads: The niche of all components of a skeleton, including the knee, is to transmit loads. All forces and couples that are applied to the knee joint can be expressed in three different directions as seen in Figure 1. The reference axis was aligned with the center of the femur and tibia. The other two axes are purposefully placed perpendicular to this axis to simplify the coordinate system. This coordinate system is dynamic because as the knee flexes, the point of contact between the two bones changes as will be discussed later. The loads can be defined by tensile, compressive and shear stresses and are transferred from one bone to the next through the complex system of ligaments. It is

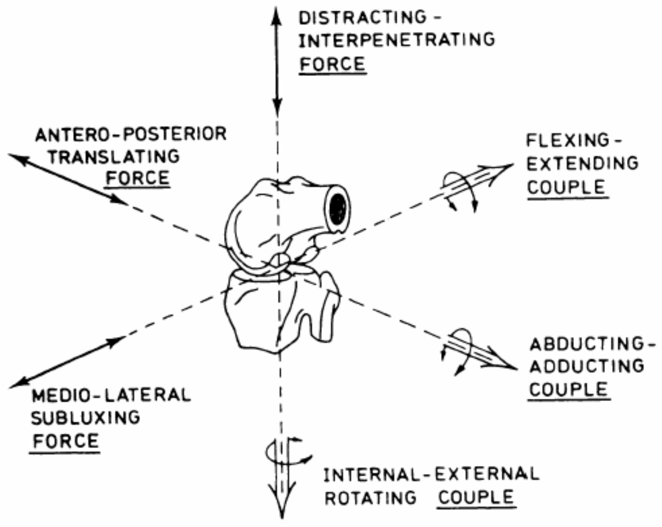


Figure 1: Reference Axis of Knee [1]

important to note that the articular surface of the normal knee joint is so smooth and well lubricated that there is no concern of significant shear stress as a result of friction. The surfaces of the two bones can only translate compressive stresses normal to their surface and ligaments can transfer tension only in the line of fibers. Therefore, the articular surfaces of the knee joint are mainly subjected to compression.

Cruciate Ligaments: The cruciate ligaments can be best described as a mechanism that resists. The ligaments are situated so as to resist flexion in the knee joint. However, they are oriented poorly to resist sliding on the articular surface of the bones, internal-external rotating couples, and movement of the bones in the medio-lateral direction. Other ligaments within the knee help resist these movements but it is the inability of the ligaments to resist the rotating motion about the tibial axis that accounts for most ACL injuries.

Joint Motions: The natural cross linkage set-up of the cruciate ligaments leads to a sliding motion being accompanied by a rolling motion during flexion of the knee. As the knee is flexed, the tibia appears to rotate about some axis in the knee. What is actually occurring internally is a combination of rolling and sliding between the surface of the tibial plateau and the femoral condyles. The crossing of the two ligaments is the driving force behind this motion. As the knee flexes, the tibia not only rolls, about the femur but there is also a sliding of the two bones on each others surfaces. The easiest way to try and visualize it is in a two dimensional model as seen below in Figure 2.

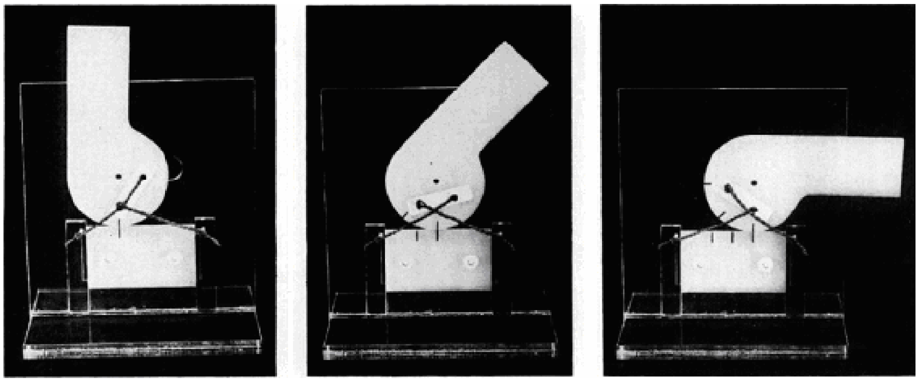


Figure 2: Two dimensional model of knee joint as a 4 bar linkage [1]

The cruciate ligaments are modeled by crossing wires, the femur by a rounded piece of plastic, and the tibia as a flat piece of plastic. The picture to the left displays the knee at full extension or 0° flexion. The lines made on the plastic femur and tibia are where the point of contact occurs between the two bones. The middle picture displays the knee at 30º of flexion. The point of contact between the two bones has now changed and is again marked by new lines. The picture on the right displays the knee at 90º flexion. Once again the point of contact has traveled is marked with another line. The easiest way to measure the rolling and sliding that occurs is through the points of contact between the two bone surfaces. Applying that to this model can be done by measuring the distances between the lines made in the photo on the left and those made on the photo on the rightIn this two dimensional model when the knee flexes the femur is the bone doing all the movement because the tibia is firmly planted in the stand. This is very similar to a person who is squatting with his or her feet firmly planted in the ground. The flexion of the knee in this case causes the femur to roll and slide posterior in reference to the tibia. It is also important to note, that in a four bar linkage model of the knee joint, the point of contact is always located perpendicular to where the cruciate ligaments cross.

The rolling to sliding relationship can be quantified by its slip ratio. The slip ratio δ is the ratio of the distance the point of contact travels along the male surface of the two objects (d_m and, in this case, the distance traveled along the femoral condyle) to the distance traveled between the two points of contact of the female surface (d_f). The tibial plateau is the female surface in the knee. In equation form.

$$\delta = \frac{d_m}{d_f} \tag{1}$$

The distance traveled by point of contact on the femur is twice that traveled by the point of contact on the tibial plateau. This equates to a slip ratio of 2. If the knee joint were a true hinge, then there would only be a rotating motion of the femoral condyle in relation to the tibial plateau. This would result in a distanced traveled by the point of contact in the femur to be some distance while the point of contact on the tibial plateau would remain in the same place creating a d_f of 0 and a slip ratio of infinity. If the femoral condyle only rolled on the tibial plateau than the distance traveled on both surfaces would be equal and result in a slip ratio of 1. The human knee is somewhere in between these two examples but is more closely related to a pure rolling scenario. This ratio is important because it will need to be applied to the new model to ensure accurate simulation of knee kinematics.

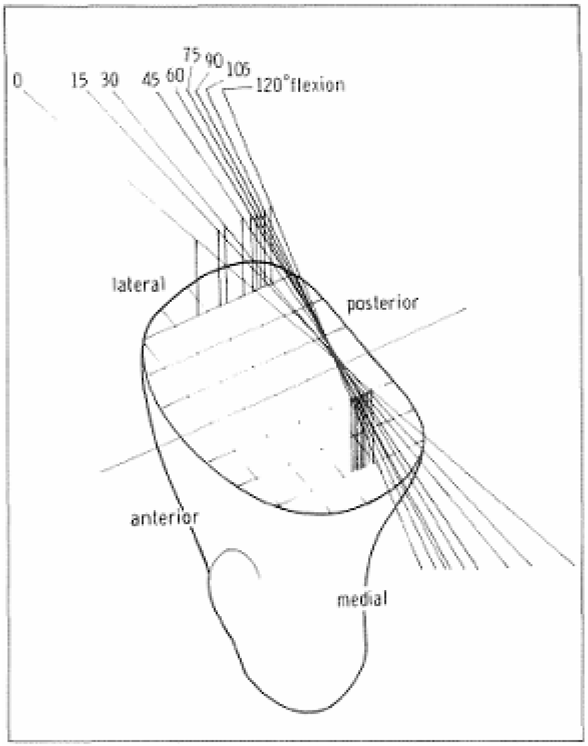


Figure 3: Condyle instant center of rotation (ICR) vs. flexion angle[2]

Although Figure 2 is a good two-dimensional model of the kinematics of the knee in relation to anterior and posterior movements as a result of the cruciate ligaments, it does not accurately model the overall kinematics in a three dimension sense. The reason for this is that the lateral condyle is slightly larger than the medial condyle. The effect of this is easily visualized as a truncated cone. As the cone is pushed it will also turn. The same is true in comparison to the different sized condyles. As the femoral condyles roll and slide one the tibial plateau, the size difference causes a rotation about the distracting-interpenetrating axis. (Figure 1) There is an instant center of rotation (ICR) that exists trough the center of both femoral condyles. As the knee flexes the ICR rotates about the distracting-interpenetrating axis altering the angular relationship of the femur to the tibia. The rotation of the ICR is shown in Figure 3 in relation to the flexion angle. The bottom part of the figure is a sectioned view of the tibia whose surface has been simplified to a flat surface to increase visibility. The line in the figure represents the ICR drawn right through the femoral condyles. At 0º flexion as noted on the top the femoral condyles ICR is perfectly aligned with the medio-lateral axis. (Figure 1) As the knee flexes the ICR rotates about the central axis cause an angular difference between the line drawn through the ICR and the medio lateral axis which is known as the internal rotation angle. The medial side of the ICR moves in an anterior direction in reference to the tibia while the lateral moves in a posterior direction. The internal rotation angle has been studied for both healthy and cruciate ligament deficient knees. For example, Amis et al [12] found that a healthy new has an internal rotation angle of 3º at 20º of flexion and 10º at 90º flexion. This will be elaborated on more under the Model Verification section. This is important because the new 3D model uses these past studies in order to demonstrate healthy and injurious knees.

Ligament Failure Modes

ACL Failure Modes: The cruciate ligaments can best be described as resistant ligaments. However, they offer little resistance to lateral and medial movements of the femur in relation to the tibia as well as unnatural rotations about the distracting-interpenetrating axis. These movements are what account for a majority of ACL injuries. They can be accomplished by forward body flexion and hip abduction. These scenarios often occur when changing directions quickly, landing awkwardly from a jump or trying to slow down when running. Injuries to the ACL can also occur when the knee is internally or externally rotated with valgus or when the foot rolls over putting an unnaturally load and rotational force on the knee joint. A less common way of tearing the ACL is a sudden impact to the knee forcing the tibia in an extreme anterior position in relation to the femur. Although this is uncommon, it mostly occurs in contact sports and car accidents.

PCL Failure Modes:

PCL injuries are a lot less common than ACL injuries and account for only 3% to 20% of knee injuries [16]. The PCL can be damaged under a similar type of impact as the ACL. However, a PCL injury occurs when the tibia is forced posterior in relation to the femur as a result of a sudden impact. This most commonly occurs in car accidents and contact sports similar to an impact injury to the ACL. The PCL also experiences trauma under excessive hyper flexion and hyper extension of the knee. The most common reason for PCL failure is when there is a total knee failure. PCL injuries are accompanied by multiple ligament injuries in 95% of injury cases. Other ligaments that are commonly injured in this type of injury are the ACL as well as lateral ligaments.

Model Design

The model has many unique features that set it apart from a standard anatomical model that can be purchased at distributors such as Saw Bones Worldwide. One of the major features is the cruciate ligament design.

The most important aspect when designing the ligaments for the model were that they were fully replaceable, easily replaceable, visually aesthetic, and anatomically correct. The idea for the replacement ligament system came from current methods in total ligament replacements.

Total ligament replacement is a complicated surgery but overall has positive results. Most ligament surgeries use autographs from the patient's patellar tendon or one of the hamstring tendons. A channel is then drilled through the patient's tibia and into the femur. This is done using drill guides to ensure proper placement of the holes as seen Figure 4.

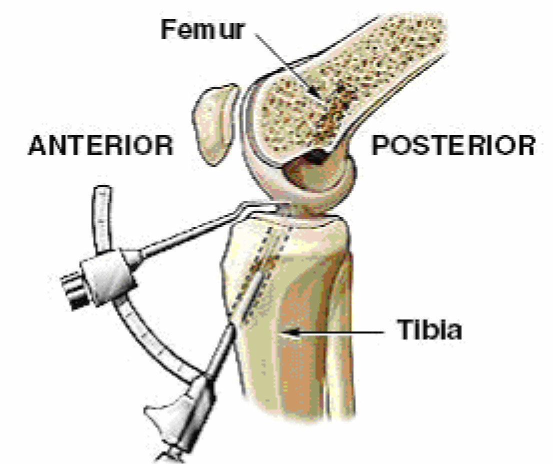


Figure 4:The drawing displays a patients knee being drilled during ACL replacement surgery.[9]

After the holes are drilled in the patient's tibia and femur, it is then time to insert the ligament replacement. The graft is then threaded through the holes where the ligaments once existed and secured using either a bioabsorbable or metal screw. The new model utilizes the same concepts but does not use a screw to secure the ligaments.

The new model simplifies the method to secure the ligaments using spring loaded clamps. On one side of the tunnel the ligament is held in place using a small washer and the ligament is pulled taught. When the ligament is pulled to its correct length the spring loaded clamp is then released holding the ligament at its correct length. The ligament was modeled using an elastic material to simulate its relative properties.

Modeling the cruciate ligaments was also very challenging given their complexities. The cruciate ligaments have complex biomechanical behaviors and are easiest described as a viscoelastic structures. The ACL is comprised of two major fiber bundles. These bundles are known as the anteromedial (AM) and posterolateral (PL) bundles. The new model uses only one elastic strand to model the ACL because of the complexities of using multiple strands. This simplification is acceptable because it will still yield accurate kinematics seeing as this method has been proven to work in myriads of ligament replacement surgeries. However, it is important to note that many people are looking into the effects of creating a double-bundle ACL reconstruction that utilizes two tunnels drilled in the patient's tibia. The goal of these studies is to show a slightly better performance and longevity in this type of replacement but this precision is not necessary in this model.

Another important aspect of the new model is the flexion measuring aid. This is a Plexiglas backing that helps the user measure the angle of flexion of the knee throughout its motion. This is a unique device because it helps the user more accurately determine the angle of flexion rather than approximating it. It is important to note that the flexion measuring aid also rotates. This will be vital when using it to determine the internal rotation of the knee as discussed later.

Bone Design

The anatomical representation of the femur and tibia were purchased from a commercial source (Saw Bones Worldwide, Pacific Research Laboratories, Vashon, WA). It is a left knee made of hard white plastic filled with polyurethane foam. The bones can be seen below in Figure 5.

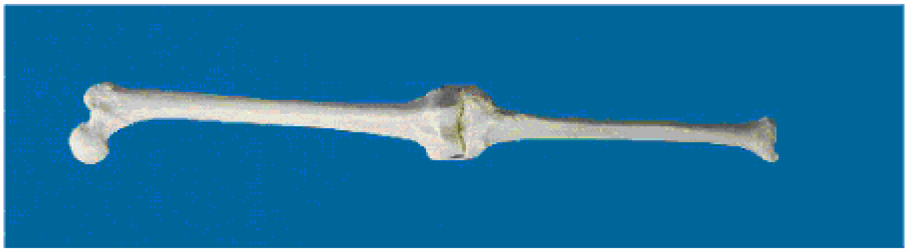


Figure 5: The picture shows the basic bone set-up from which the knee model is made.

The tibia is trimmed to 8” long and the femur trimmed to 8” in order to create a manageable model. A ½” threaded rod was inserted into the bottom of the tibia to create a way to mount the bone to the base of the model.

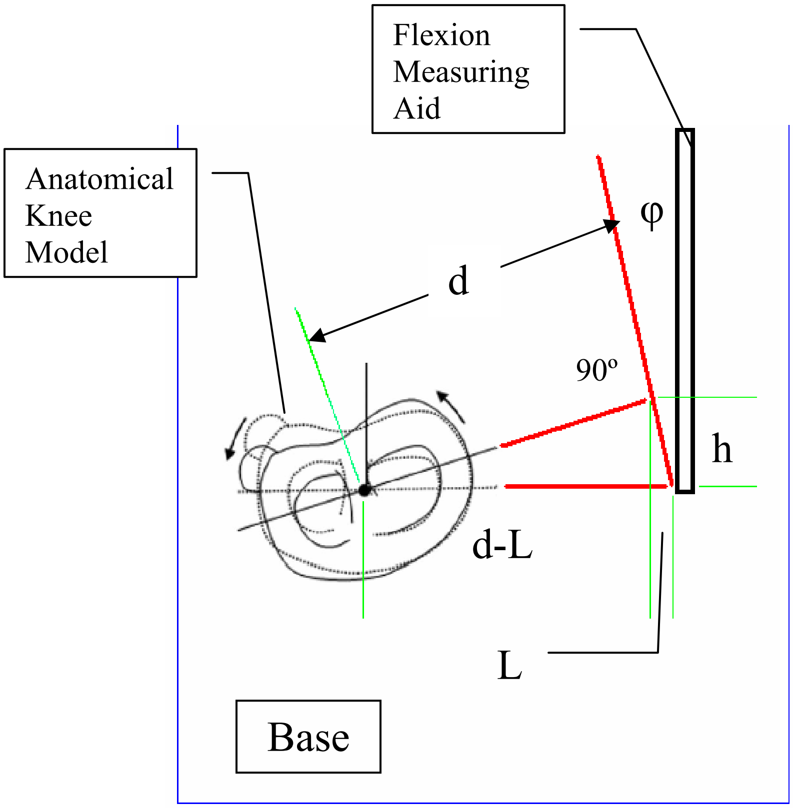
Model Verification

The first step to verifying the model was to check the major kinematics of a healthy knee and see how they compare to that of an actual knee. The two major components of kinematics as discussed before are the anterior and posterior movement of the bones in relation to each other and the internal rotation that also occurs during flexion.

The model was able to display a rolling to slide ratio that was equal to that in a real knee. This ratio was described to be about two with the distance traveled of the point of contact on the femur to be about twice that of the distance traveled by the point of contact on the tibia. The model now had to verify the internal rotations of the two bones to ensure that it was accurately modeling the kinematics.

Under flexion the new model displayed internal rotations similar to those found by Amis et al [12]. These are a 3º internal rotation at 20º flexion and a 10º internal rotation at 90º flexion. Although the internal rotation was easily measured by making marks on masking tape and measuring the angle, a better method of determining the internal rotation angle needed to be produced to make it user friendly.

An easier way of measuring the internal rotational angle was derived using the flexion measuring aid. The figures below are a basic overhead view of the model and are the basis of the derivation. Figure 7 displays all of the components while Figure 7 is a simplified version only displaying the basic geometry.



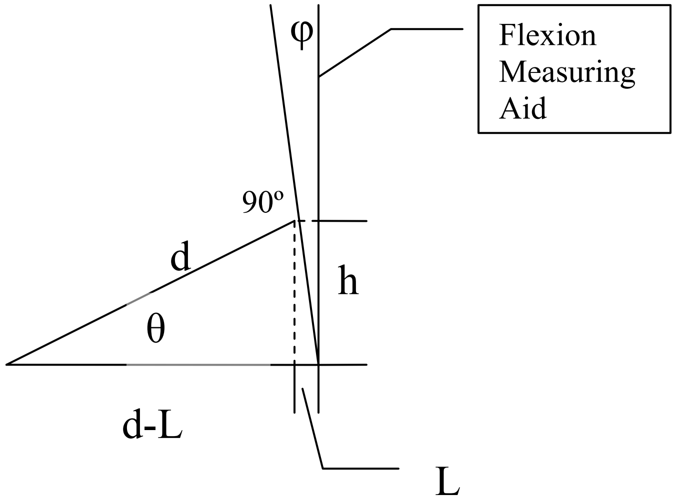


Figure 7: Basic geometric overhead view of model.

Figure 6: Overhead view of model and its geometries.

In the figure above *d* is the distance from the axis of rotation to the flexion measuring aid, θ is the angle of internal rotation, φ is the angle at which the flexion measuring aid needs to be turned in order to keep contact with the pointer protruding from the femur, and *h* and *L* are measured distances within the geometry. Now that the geometry was laid out, a derivation could be used to create an equation to relate the angle φ with the angle θ. The purpose for relating these two angles is to create a simplistic system to measure internal rotation that is more accurate than approximating the values. By measuring the internal rotation we can compare rotation angles in a healthy knee with that of an injured one. The equation found to relate the angles is below and the derivation can be found in Appendix A.

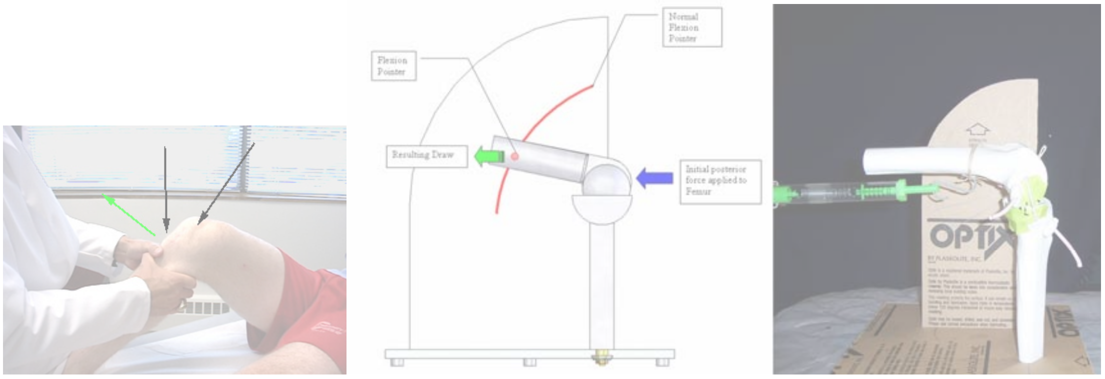
$$\theta = 2\phi \tag{8}$$

The angle of internal rotation can now be easily measured within the model using the pointer secured in the femur and the flexion measuring aid. Corresponding measuring angle marks were made in the base to make these measurements

Ligament Deficient Knee Tests

There are several different ways in which the failure of cruciate ligaments affects the functionality of the knee joint. The ACL's major function is the restraint to tibial anterior translation and is responsible for bearing up to 85% of forces applied to the arc of the knee. [10] Currently there are several methods used by physicians to check for cruciate ligament deficiencies in patients. Applying these widely known techniques to the new model, a visual demonstration of what occurs under the skin during an examination of a patient’s knee can be examined.

The most common type of ACL injury test is the Anterior Drawer Test. In this test the knee is brought to approximately 90 degrees flexion and patient’s foot is held firmly in place and not allowed to rotate. The examiner places his hands the patient’s proximal tibia and applies an anterior translation. Access anterior draw is a sign of a damaged ACL. This method needs to be slightly altered in order to apply it to the new model because the tibia is firmly secured into the base. As a result, a posterior force is applied to the femur instead to check for excessive draw. The visualization of this process is shown below in Figure 8.



(a) (b) (c)

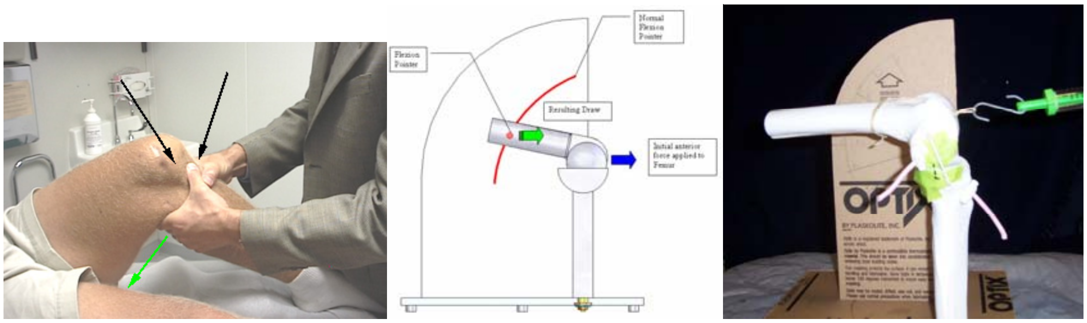
Figure 8: The photo on the left (a) shows an anterior draw test being applied to a patient [14] while (b) shows the anterior draw test being applied to the model and

(c) is the actual application of the anterior draw test using a spring scale to apply the force.

The posterior force applied to the femur causes an anterior draw to occur in terms of the tibia. It is the absence of the ACL that allows this draw to occur. When the knee is fully intact, this posterior force applied to the femur will cause very little deviation from the guide line drawn in red in Figure 1 which will be present on the model's flexion measuring aid. It is important to note that the guideline is not perfectly rounded because of the rolling and sliding motion of the knee. Lachman's test utilizes the same methodology but with the knee placed in 30 degrees flexion. The large deviation from the guideline at 30 degrees flexion displays draw in an injured knee.

Many tests have shown that the ACL has an impact on controlling axial rotation within the knee and that the absence of the ACL increases the laxity and internal rotation possible within the knee. The new model can also be used to show this by utilizing the flexion measuring aid and the flexion pointer. The rolling and sliding of the femoral condyle on the tibial plateau is also accompanied by an internal rotation due to the geometries of the bone as previously discussed. In a healthy knee this rotation occurs around the center of the tibia and accounts for a mean of 3º rotation at 20º of flexion and 10º at 90º of flexion.[12] In a knee with a torn ACL, Andersen et al found the rotational laxity to range from 28º at 10º flexion to 42º at 90 degrees flexion.[11] It is also important to note that Lipke discovered that isolating a failure only in the ACL leads to a significant increase in tibial internal rotation but when other ligaments are ruptured by themselves they have little overall effect on the internal rotation of the knee joint.[13] This enforces the idea that the ACL has a large impact on the internal rotations of the knee. However, the largest effect on internal rotation is when there is failure of several ligaments including the ACL and PCL.

The posterior draw test is used to test for PCL failure and can also be applied to the new model. Similarly to the anterior draw test, the posterior draw test is applied to a patient with his knee at around 90º flexion. Alternatively in this test, a posterior force is applied to the tibia. Posterior draw is a sign that the patient has experienced a PCL injury. PCL injuries are extremely rare on there own but can be tested by this method non-the-less. Since the tibia is firmly secured as mentioned before, an anterior force can be applied to the model to create a simulation of the posterior draw test. This is shown in Figures 9.



(a) (b) (c)

Figure 9: The photo on the left (a) shows a posterior draw test being applied to a patient [14] while (b) shows the posterior draw test being applied to the model and (c) is the actual application of the posterior draw test using a spring scale to apply the force.

The deviation from the red guide line during the Posterior Draw Test demonstrates the absence of the PCL. This can be demonstrated in the model with only the PCL removed or also with both ACL and PCL removed and using both the Anterior and Posterior Draw Test. There is no literature present on the effects of a PCL deficient knee with respect to internal rotation laxity. It is said that research done on the PCL is about ten years behind that of ACL research. [16] Perhaps the reason for this is that it appears to be an impossible case. However, laxity is increased with total failure of both cruciate ligaments.

Results

It was vital to obtain results for testing ligament deficient models in order to displays the difference for the user. A spring scale with a maximum resulting force of 5 N was used in these experiments. It is also important to note that during the Lachman's Test the spring scale was applying a force parallel to the orientation of flexion. The resulting force horizontal that would be applied by a doctor is only half of the 5 N because sin(30) is equal to 0.5. At no point during any of Lachman's Tests did the bones loose contact. It was also assumed that all friction was negligible. This was made because the mass of the femur bone is so small and with a static friction coefficient of plastic on plastic being around one-third it seemed as if this would play a small role in the resulting force. This was also verified because it took a minute amount of force to instigate movement when both ligaments were removed. For this experiment the distances were measured on the model by making marks on the masking tape and measuring the distance at which the points of contact on the femur traveled in relation to the point of contact of the tibia. There were three major tests phases with each ligament. The first was to test the knee model with the entire ligament removed, the second was to test the knee model with slight laxity in the ligament and the final was to test it with a large laxity in the ligament. The laxity was created in the ligaments to try and simulate different levels of cruciate ligament injuries. To accomplish this, the ligament was loosened 12.5 mm past its "healthy" length to create a slight laxity and 25 mm to create a large laxity.

There is a certain amount of human error involved when using measuring tools. To make up for these errors, the same experiment was done several times in a row to get multiple data points that are about some average distance. The initial experiments were done with the ligaments in tact. The force of 5 N (or 2.5 N in the Lachman's Test) resulted in no visible difference. Only when the model was exposed to extreme force did any noticeable draw occur. The new model was then tested by singling out the individual ligaments and then testing them both with deficiencies. The table below displays the offset distance (mm) of five trials in the model with an ACL deficiency

	ACL Deficient Model														
Test	No ACL					Slight ACL Laxity(12.5mm)					Large ACL Laxity (25mm)				
ADT	6.2	5.9	7.9	6.5	6.8	0.0	0.0	0.0	0.0	0.0	3.3	3.0	3.0	2.9	3.2
Lachman	1.6	1.4	1.9	2.4	2.5	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
PDT	0.0	0.2	0.0	0.3	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0

Table 2: Displays draw distance during each test for five separate trials in each ACL deficient model.

There are a few important things to point out here as a result of the data. The first is that there is almost no difference in posterior and anterior laxity in the slightly lax model. However, although this is true there is an increase in internal rotation laxity that occurs. Also, the force applied during the Lachman's Test was not large enough to make a difference in the largely lax model. Another thing that is important to notice was the slight draw that occurred during the PDT with no ACL. The reason for this is that when there is no ACL present in the model, the femur sits slightly posterior to where it normally does in the "healthy" knee model. By applying this anterior force to the femur to simulate a PDT, it pulls the femur back into its "healthy" alignment. The next group of experiments was done on PCL deficient models.

The resulting data from experiments run on the PCL deficient model are shown below. Although it is unheard of for the PCL to fail on its own, it was of interest to see what effects it would have in the theoretical case that it did fail.

	PCL Deficient Model														
Test	No PCL					Slight PCL Laxity(12.5mm)					Large PCL Strain(25mm)				
ADT	1.6	2.7	2.4	2.5	2.7	1.6	1.1	0.8	0.6	1.0	1.9	2.1	1.9	1.7	2.2
Lachman	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
PDT	4.6	4.8	4.3	5.1	5.2	0.0	0.0	0.0	0.0	0.0	3.5	3.3	4.9	3.5	3.7

Table 3:Displays draw distance during each test for five separate trials in each PCL deficient model

A similar phenomenon occurs during the ADT in the PCL deficient knee model as occurs during the PDT in the ACL deficient knee. The displacement between the point of contact in the "healthy" model and that in the injurious model makes up the total displacement in the results. In this case the difference is much larger than in the injured ACL model. The last test was now done combining both injuries.

These results show that by injuring both ligaments you increase the laxity in the joint and the resulting draw during each test.

	ACL & PCL Deficient Model														
Test	No ACL or PCL					Slight ACL & PCL Laxity(12.5mm)					Large ACL & PCL Strain(25mm)				
ADT	6.8	6.2	7.1	7.6	6.7	0.0	0.0	0.0	0.0	0.0	5.9	5.6	5.4	6.2	5.9
Lachman	4.6	4.8	5.1	4.3	4.9	0.0	0.0	0.0	0.0	0.0	3.2	3.3	3.3	3.2	4.0
PDT	9.5	9.2	9.7	9.7	10.0	0.0	0.0	0.0	0.0	0.0	3.7	3.5	3.7	3.0	2.7

Table 3:Displays draw distance during each test for five separate trials in each ACL and PCL deficient model

As expected all of the resulting draws are much greater with dual deficiencies than with only one. The values in yellow are a result of only 3 N versus the usual 5 N that was being applied in the other experiments. This is because the 5 N caused the femur to be pulled right off of the tibia so a smaller value was used to create a resulting draw that could be measured. All of the values were then averaged and are displayed in Table 4 below.

	Averages for all Values								
	None			Small Strain			Large Strain		
	ACL	PCL	Both	ACL	PCL	Both	ACL	PCL	Both
ADT	6.7	2.4	6.9	0.0	1.0	0.0	3.1	2.0	5.8
Lachman	2.0	0.0	4.7	0.0	0.0	0.0	0.0	0.0	3.4
PDT	0.1	4.8	9.6	0.0	0.0	0.0	0.0	3.8	3.3

Table 4: Displays the averages of all draw distances for each experiment

Once again this table drives home the idea that injuring both ligaments results in draws much larger than injuring one. Also, although there are negligible changes in the ligaments abilities to deal with anterior and posterior forces, there is an increased laxity in the internal rotation of the knee joint.

DISCUSSION

The results of ligament deficient tests can be correlated with real injurious knees and help classify the severity of the injury. The knee produces a healthy motion and responds to the tests with very little laxity when the ligaments are tightened to the proper length. These classifications are known as Grades of injuries and are denoted from I to III.

The model will display a Grade I injury as you loosen the ligament and implement the spring loaded clamp up to 12.5 mm away from the healthy clamping point. A Grade I injury is best described as a ligament sprain. The trauma experienced by the ligament is relatively small and some of the fibers within the ligament have become stretched. The remedy for this is rest and ice and requires no surgery.

The model can be used to demonstrate a Grade II ligament injury when the ligament is again loosened between 12.5 mm and 25 mm. This simulates a partial tear in the ligament. This type of injury can result in surgery depending on the severity of the tear. However, if surgery is needed the patient will receive ligament repair surgery instead of complete ligament replacement. The model simulates a more severely injured knee the more it is loosened.

The model will also be able to simulate a Grade III injury. This is demonstrated by totally removing the ligament from the model. A Grade III injury is known as a complete tear of the ligament. This necessitates knee surgery because of the lack of stability and loss of functionality in the knee.

Model verification and results display that this model does mimic the kinematics of the knee. The tests that were chosen to display ligament deficiencies are extremely sensitive when there is a ligament absence as well as highly resistive when the knee is in its healthy stage. If this model were to be built again in several years it would be important to pursue research that is currently being done on double bundle reconstructions to see what types of strides are made in that area and to see if they could be applied to better improve the new model.

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APPENDIX A:

The first equation created was related h to θ and d.

$$d \cdot \sin(\theta) = h \tag{2}$$

An equation can also relate *h* to angle *φ* which will give me an initial way to relate the two angles to each other.

$$h = \frac{L}{\tan(\phi)} \tag{3}$$

Before these two equations could be related together, *L* had to be removed from equation 3. The reason for this is that *L* is an unknown value. Since the variable *d* is known, an equation was determined to relate these two variables using the larger triangle.

$$\cos(\theta) = \frac{d - L}{d} \tag{4a}$$

$$d \cdot \cos(\theta) = d - L \tag{4b}$$

$$d(1 - \cos(\theta)) = L \tag{4c}$$

Equation 4c was then used to substitute for *L* in equation 3.

$$h = \frac{d(1 - \cos(\theta))}{\tan(\phi)} \tag{5}$$

Equation 5 and equation 2 could now be set equal to each other in order to relate the two angles.

$$d \cdot \sin(\theta) = \frac{d(1 - \cos(\theta))}{\tan(\phi)} \tag{6}$$

This equation was rearranged and solved for *φ*.

$$\tan(\phi) = \frac{(1 - \cos(\theta))}{\sin(\theta)} \tag{7a}$$

$$\tan(\phi) = \frac{1}{\sin(\theta)} - \frac{\cos(\theta)}{\sin(\theta)} \tag{7b}$$

$$\tan(\phi) = \cos ec(\theta) - \cot an(\theta) \tag{7c}$$

$$\boxed{\phi = \arctan(\cos ec(\theta) - \cot an(\theta))} \tag{7d}$$

This is the final equation that was created to relate the two angles. Now it had to be applied to realistic angles. On thing that is important to note is that the radial distance from the center of rotation to the measuring aid cancels itself out and has no effect on the final equation. This is extremely important because the center of rotation changes after the ACL has failed from a central position to that of a medial position. Therefore, the absence of a radial distance term means that the same equation can be used in both situations without having to manipulate any values.

Using previous studies a healthy knee usually rotates about 3 ° at a 20 ° flexion and 10 ° at 90 ° flexion. In an injured knee that rotational angle increase to 28 ° at 10 ° flexion and 42 ° at 90 ° flexion.[12] Therefore, the total range of rotation that needs to be measured is from 0 to 45 degrees flexion. A MatLab script was created to take in these values for θ and calculate the value for the angle φ using equation 6d. There is a very interesting relation shown here between θ and φ. No matter what the value of θ is, φ is always half of it. This even holds true when the step is decreased to tenths of a degree.

The final relationship between the two angles can be simply shown by the equation below:

$$\theta = 2\phi \tag{8}$$

The angle of internal rotation can now be easily measured within the model using the pointer secured in the femur and the flexion measuring aid. Corresponding measuring angle marks were made in the base to make these measurements