The effects of graft size and insertion site location during anterior cruciate ligament reconstruction on intercondylar notch impingement

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ABSTRACT

Background: Intercondylar notch impingement is detrimental to the anterior cruciate ligament (ACL). Notchplasty is a preventative remodeling procedure performed on the intercondylar notch during ACL reconstruction (ACLR). This study investigates how ACL graft geometry and both tibial and femoral insertion site location may affect ACL-intercondylar notch interactions post ACLR. A range of ACL graft sizes are reported during ACLR, from six millimeters to 11 mm in diameter. Variability of three millimeters in ACL insertion site location is reported during ACLR. This study aims to determine the post-operative effects of minor variations in graft size and insertion location on intercondylar notch impingement.

Methods: Several 3D finite element knee joint models were constructed using three ACL graft sizes and polar arrays of tibial and femoral insertion locations. Each model was subjected to flexion, tibial external rotation, and valgus motion. Impingement force and contact area between the ACL and intercondylar notch compared well with experimental cadaver data from literature.

Results: A three millimeter anterior–lateral tibial insertion site shift of the maximum size ACL increased impingement force by 242.9%. A three millimeter anterior–proximal femoral insertion site shift of the maximum size ACL increased impingement by 346.2%. Simulated notchplasty of five millimeters eliminated all impingement for the simulation with the greatest impingement. For the kinematics applied, small differences in graft size and insertion site location led to large increases in impingement force and contact area.

Conclusions: Minor surgical variations may increase ACL impingement. The results indicate that notchplasty reduces impingement during ACLR. Notchplasty may help to improve ACLR success rates.

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1. Introduction

The anterior cruciate ligament (ACL) acts as a major motion stabilizer for the knee joint by preventing anterior tibial displacement and providing torsional stability [1–6]. The ACL is the most commonly injured ligament within the knee [7–9].

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These painful injuries occur upwards of 400,000 times annually in the United States [8,10,11]. ACL reconstruction (ACLR) and rehabilitation are costly, and represent an annual $1 billion expense in the United States [10–12]. This cost is expected to rise with the anticipated increase in ACLR performed annually [13].

Up to 90% of ACL injured patients elect to have ACLR. ACLR can be performed with either an autograft or allograft. Allografts are not as common due to the risks of infection and host rejection. Autografts are common, and can be performed using a bone-patellar tendon–bone (BPTB) graft, or with a hamstring graft using the semitendinosus in either a single or anatomical double bundle technique. Several review papers have reported that no significant benefit exists between the various techniques, and recommend the use of a single bundle BPTB graft [14–16].

Reinjury may be more likely for athletes who have undergone ACLR [17,18]. Wright et al. reported that six percent of patients who underwent ACLR had a reinjury within two years [19]. Salmon et al. reported that 12% of patients with an ACLR had a reinjury within a five year follow-up window [20]. In this same patient population, in a 15 year follow-up window, Leys et al. reported up to a 34% reinjury rate [21].

The ACL reinjury rate reportedly varies depending on the reconstruction technique [22–24]. Several studies have compared autograft vs allograft techniques for ACLR reconstruction. Ellis et al. reported that 35% of patients who underwent allograft BPTB replacement required revision surgery within one year, compared to three percent for patients who had autograft replacement [22]. Vishal et al. reported that with a mean follow-up of 49 months, 0.7% of patients who had an autograft BPTB replacement required revision surgery compared to 9.7% for patients who had an allograft BPTB replacement [23].

Differences have been reported regarding the reinjury rate of ACLR patients [17,19,20]. However the reinjury rate has been reported as high as six times greater than healthy patients [17]. Because of these high surgical revision rates, it is important to understand the potential factors contributing to reinjury. Inaccurate placement of the ACL insertion sites during ACLR has been reported in 10% to 40% of ACL tunnel placements [25,26]. Up to three millimeters of variation in graft placement from the anatomical insertion site has been reported with experienced surgeons [27]. A range of ACL graft sizes have also been reported for use during ACLR [28–31]. Wilson et al. reported an average BPTB graft diameter of 9.9 mm and an average cross section of 44.6 mm², with a 2.2 mm standard deviation for the diameter, and a 23.1 mm² standard deviation for the cross sectional area [29]. Magnussen et al. reported graft sizes between seven millimeters and nine millimeters in diameter [28]. Tuman et al. reported that in a group of 106 patients, the average graft size was 7.7 mm in diameter, with two percent having grafts six millimeters in diameter, and one percent having grafts 10 mm in diameter or larger [30].

Several investigations have reported impingement between the ACL and the intercondylar notch during knee joint motion [7,32,33]. Park et al. performed a study which evaluated a three-dimensional (3D) finite element (FE) model of ACL impingement within the femoral intercondylar notch [7]. Park et al. validated their model with experimental data collected from an instrumented cadaver [7]. Knee flexion, external tibial rotation, and valgus motion were applied to the cadaver, and the contact area and impingement force data were collected. The same kinematic data were applied to the FE model. The contact area and impingement force results predicted in the FE model were in close agreement with the cadaver experiment, validating the FE model as a useful tool for predicting ACL impingement.

Femoral notchplasty is a surgical procedure in which the intercondylar femoral notch is widened during ACLR to prevent ACL impingement with the femoral intercondylar notch. Intercondylar notch impingement is thought to be a leading cause of ACL injury [8,12,34]. Notchplasty is commonly performed during ACLR; however a standard protocol has not been well defined. There is no recommended amount of bone removal during notchplasty, and different studies suggest varying amounts of notchplasty [35–37].

The purpose of the present study is to understand how surgical variations affect ACL-intercondylar notch interactions post ACLR. This is important because ACL-intercondylar notch impingement may lead to ACL injury [7,8,12,32,34,38]. The results of this study have the potential to improve ACLR success rates. Furthermore, as this study provides a method to quantify the amount of notchplasty to be performed during ACLR, the results may support the use of surgical notchplasty to reduce the risk of ACL reinjury.

2. Methods

A subject-specific 3D FE model of a male left knee was created from sagittal view magnetic resonance images (MRI) using the method provided by Homyk et al., Yang et al., Orsi et al. and Haut Donohue et al. [39–43]. The model is seen in Figure 1a. Details of the MR data acquisition can be found in our previous work [39–42]. The MRIs were converted into 3D solid structures using solid modeling software packages Rhinoceros (Robert McNeel & Associates, Seattle, WA) and SolidWorks (Dassault Systemes, France). The solid structures were imported into the FE software package ABAQUS (Dassault Systemes, France) and converted to an FE mesh for use in kinematic FE simulations.

A free meshing technique was used for the cartilage and meniscus using four-node linear tetrahedral elements. The ACL was meshed using hexahedral elements. Details justifying these element types and any volumetric locking effects rising from using these elements are provided in Orsi et al. [41].

Bone was modeled rigid as it is much stiffer than the soft tissue it interacts with [39–43]. The articular cartilage was modeled as isotropic linear elastic and the meniscus was modeled as transversely isotropic linear elastic with the material properties shown in Table 1. The menisci were attached to the tibial plateau with linear spring elements. The transverse ligament was modeled as a single spring element which connected the anterior horns of the medial and lateral meniscus.
Contacts between the femoral articular cartilage, tibial articular cartilage and medial and lateral menisci were included as well as contact between the ACL and the femoral intercondylar notch wall. Contact was incorporated using a frictionless finite-sliding formulation where separation and sliding of finite amplitude and arbitrary rotation of the surfaces were allowed. Contact interaction normal to the contacting surfaces was constrained using the standard penalty enforcement method.

The posterior cruciate ligament (PCL), medial collateral ligament (MCL), and lateral collateral ligament (LCL) were modeled as multi-bundled nonlinear spring elements. Ligament insertion sites were determined from the MRI, similar to the group’s previous work [39–42]. The nonlinear spring force-displacement relationship used was defined as a piecewise continuous function,

\[
f = \begin{cases} 
\frac{2}{3} k \varepsilon^2, & 0 \leq \varepsilon \leq 2\varepsilon_0 \\
 k(\varepsilon - \varepsilon_0), & \varepsilon > 2\varepsilon_0 \\
 0, & \varepsilon < 0 
\end{cases}
\]  

(1)

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Constitutive model</th>
<th>Properties</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cartilage</td>
<td>Isotropic elastic</td>
<td>$E = 15.0$ MPa, $\nu = 0.45$</td>
</tr>
<tr>
<td>Meniscus</td>
<td>Transversely isotropic elastic</td>
<td>$E_p = 140$ MPa, $E_L = E_T = 20$ MPa</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\nu_{xx} = 0.2, \nu_{yy} = \nu_{zz} = 0.3$</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$G_{xx} = G_{yy} = 57.7$ MPa, $G_{yy} = 8.3$ MPa</td>
</tr>
</tbody>
</table>
where $f$ is the tensile force, $k$ is a stiffness parameter and $2\varepsilon_0$ is the lower bound strain limit for the linear ligament behavior. $\varepsilon$ is the strain in the ligaments defined as $\varepsilon = (L - L_0)/L_0$, where $L$ is the ligament length and $L_0$ is the unstretched zero-load length of the ligament. At full knee extension the initial reference strain, $\varepsilon_i$, is listed in Table 2 for each bundle. $L_0$ is found using $\varepsilon_i$ along with the initial reference length of the ligament, $L_0$, using $L_0 = L_0/(\varepsilon_i + 1)$, where $L_0$ is determined from the MRI as the distance between the tibial and femoral insertion sites. This study modeled the PCL as a double bundle (anterior and posterior bundle), the LCL and MCL were modeled with three bundles. The properties of each ligament were adapted from the work of Blankenvoort et al., shown in Table 2 [44].

A fiber orientation dependent transversely isotropic hyperelastic material, defined by a strain energy density, $\psi$, was used to obtain the constitutive equations of the ACL [45]. The strain energy density was defined as,

$$
\psi = C_1 \left( I_1 - 3 \right) + C_2 \left( I_2 - 3 \right) + F_2 \left( \lambda \right) + \frac{K}{2} (\ln(\lambda))^2
$$

where $J$ is the Jacobian of the deformation gradient $F$ defined as $\frac{\partial x}{\partial X}$ where $x$ and $X$ are coordinates of each point in the deformed and undeformed configurations. $C_1$ and $C_2$ are constants representing the Mooney-Rivlin material model and $K$ is the bulk modulus of the material. $I_1$ and $I_2$ are the first and second invariant of the modified Cauchy-Green strain tensor $\mathbf{C} = J^{\frac{2}{3}} F^T F$. The derivative of the fiber strain energy function $F_2$ was defined as

$$
\lambda \frac{\partial F_2}{\partial \lambda} = 0, \quad \lambda \leq 1
$$

$$
\lambda \frac{\partial F_2}{\partial \lambda} = C_3 \left[ e^{C_4 (\lambda - 1)} - 1 \right], \quad 1 < \lambda < \lambda^* \quad (3)
$$

$$
\lambda \frac{\partial F_2}{\partial \lambda} = C_5 \lambda + C_6, \quad \lambda \geq \lambda^*.
$$

The first relationship reflects the inability of ligament structures to support compressive loads. The second relationship is the nonlinear 'toe region' corresponding to the un-crimping of the collagen fibers. $C_3$ scales the exponential stresses, and $C_4$ represents the rate at which the collagen fibers uncrimp. The third relationship corresponds to the linear stress-strain response of the straightened fibers. $C_5$ is the modulus of the straightened fibers and $C_6$ is calculated so that the stress is continuous at $\lambda^*$. $\lambda^*$ is the amount of stretch at which the material transitions from un-straightened to straightened fibers. $\lambda$ is defined as the deviatoric stretch along the fiber direction. The material constants were extracted from the work of Peña et al. [46]. These constants are presented in Table 3. Fiber direction was defined as the vector connecting the centroids of the tibial and femoral insertion sites.

Three ACL size models were constructed based on the average, maximum and minimum ACL graft cross sectional geometries reported in the literature [28-31]. These were created by lofting solid structures from the insertion sites on the femur and tibia to a uniform cross section in the ACL mid substance. The cross sectional area of the average size ACL model was 46.1 mm$^2$, with an anterior-posterior thickness of 9.5 mm and a medial-lateral thickness of 6.2 mm. The cross sectional area of the maximum size ACL model was 56.1 mm$^2$, with an anterior-posterior thickness of 10.5 mm and a medial lateral thickness of 6.8 mm. The cross sectional area of the minimum size ACL model was 37.4 mm$^2$, with an anterior-posterior thickness of 8.5 mm and a medial lateral thickness of 5.6 mm.

Time dependent material properties were not used in this investigation. The transversely isotropic hyperelastic Mooney-Rivlin model has been used extensively to model human ligament [7.41.45.46]. Previous work has validated this material model for predicting the impingement force and contact area of the ACL with the femoral intercondylar notch [7].

To validate the FE model, the kinematic data from Park et al. were applied to the average size ACL model configuration with the original insertion sites [7]. The knee was moved from its initial position with a knee flexion angle of 46.3°, abduction angle of 0°, and 0° of external tibial rotation (ER), to its final position with a knee flexion angle of 44.8°, 10.0° of abduction, and 29.1° of ER. The knee joint axes of rotation are provided in Figure 1b. The final position is shown in Figure 1c, with an isolated view provided in Figure 1d. Figure 2a shows the kinematic boundary conditions applied to the model. The applied kinematic data simulates landing or cutting maneuvers which are known to be associated with high ACL impingement and ACL injury [7.32.47].

Table 2

<table>
<thead>
<tr>
<th>Ligament</th>
<th>Bundle</th>
<th>Stiffness parameter, k[N]</th>
<th>$\varepsilon_i$</th>
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<tbody>
<tr>
<td>PCL</td>
<td>Anterior</td>
<td>9000</td>
<td>-0.24</td>
</tr>
<tr>
<td></td>
<td>Posterior</td>
<td>9000</td>
<td>-0.03</td>
</tr>
<tr>
<td>LCL</td>
<td>Anterior</td>
<td>2000</td>
<td>-0.25</td>
</tr>
<tr>
<td></td>
<td>Superior</td>
<td>2000</td>
<td>-0.05</td>
</tr>
<tr>
<td></td>
<td>Posterior</td>
<td>2000</td>
<td>0.08</td>
</tr>
<tr>
<td>MCL</td>
<td>Anterior</td>
<td>2750</td>
<td>0.04</td>
</tr>
<tr>
<td></td>
<td>Inferior</td>
<td>2750</td>
<td>0.04</td>
</tr>
<tr>
<td></td>
<td>Posterior</td>
<td>2750</td>
<td>0.03</td>
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</tbody>
</table>
Table 3

<table>
<thead>
<tr>
<th>$C_1$ (MPa)</th>
<th>$C_2$</th>
<th>$C_3$ (MPa)</th>
<th>$C_4$</th>
<th>$C_5$ (MPa)</th>
<th>$\lambda$</th>
<th>$K$</th>
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<tbody>
<tr>
<td>1.95</td>
<td>0.0</td>
<td>0.0139</td>
<td>116.22</td>
<td>535.039</td>
<td>1.046</td>
<td>1950</td>
</tr>
</tbody>
</table>

The contact area and impingement force data were compared between the FE model and the experimental data from Park et al. [7].

The cadaver sample from Park et al. exhibited impingement, and had a notch width index (NWI) of 0.22 [7,32]. NWI is defined as the width of the intercondylar notch divided by the distance between the epicondyles [48]. The model developed in the present investigation exhibited a NWI of approximately 0.25. Figure 2b shows the impingement force and contact area data from the average size ACL model closely approximate the experimental data from Park et al. [7]. The peak impingement force from the Park et al. experiment was 36.9 N, and the FE model predicted a peak impingement force of 35.1 N. The contact area at peak impingement from Park et al. was 19.7 mm$^2$, while the model predicted a contact area of 20.7 mm$^2$.

To investigate how ACL graft size and both tibial and femoral insertion site location affect impingement within the intercondylar notch, several configurations of the FE model were developed. The centroids of the tibial and femoral insertion sites were relocated in the $r$ and $\theta$ directions for each of the three ACL size models. Radii of one millimeter and three millimeters were used in a polar array of 45° increments from 0° to 360°. Figure 3 shows the tibial and femoral insertion site configurations used in this investigation. For each configuration, the kinematic data from Park et al. were applied and impingement force and contact area data were obtained [7].

Notchplasty configurations were created using the maximum size ACL model with three millimeter lateral shift of the tibial insertion site as well as a three millimeter antero-proximal shift in the femoral insertion site. To simulate notchplasty, the lateral femoral notch wall surface was isolated, and translated laterally by increments of one millimeter.

Contact interaction between the ACL and PCL could affect the impingement results of this investigation. To determine if interaction existed between the ACL and PCL for the kinematics used in this investigation, a 3D PCL model was incorporated into our model. A worst case configuration was simulated by including the maximum size ACL graft with a three millimeter medial shift of the tibial insertion site and the 3D PCL. For the kinematic boundary conditions applied, no ACL–PCL interaction existed, validating the exclusion of the 3D PCL.

Understanding the effects that graft size and insertion site shifting may have on knee joint kinematics is important, as kinematic changes could further affect impingement. To determine if these variables affect knee joint kinematics, several tests were performed using three configurations of the model: the standard configuration, the three millimeter AP femoral shift model, and the three millimeter L tibial shift model. Each model was subjected to three loading scenarios: a two Newton-meter flexion moment, a three newton-meter ER moment, and a 30 N-m valgus moment. These values correspond to the reaction moments associated with knee joint motions applied in our previous work. The resulting kinematics from each test were compared across the three model configurations. It was observed that minimal differences in kinematics existed between the insertion site shifting models across the three tests. This result supports the method of applying the same kinematic data to each model configuration.

![Boundary Conditions](image1.png)

**Figure 2.** a) Boundary conditions applied to the FE model, extracted from Park et al. [7]. b) Comparison of impingement force and contact area between Park et al. and the model presented in this investigation [7].
3. Results

The contact area and impingement force results from the average size ACL model in the original insertion site configuration compared well with the experimental data from Park et al. [7]. This provided a baseline against which the simulations of the different ACL configurations could be compared.

To understand how ACL graft size alone affects impingement, the simulations of the three ACL size models at their original insertion sites were compared. The results are presented in Figure 4. For the kinematics applied, the maximum size ACL

![Figure 3](image)

**Figure 3.** a) Superior axial view of left tibia showing ACL insertion locations. b) Graphic showing shift of insertion site by radius r and angle θ. c) Sagittal plane cross section view of femur showing femoral insertion locations. d) Graphic showing shift of femoral insertion site by radius r and angle θ. Radii of one millimeter and three millimeters were used in this investigation. Green circle represents the original insertion site. Offset locations are shown as gray circles, equally spaced in 45° increments from 0° to 360°.

![Figure 4](image)

**Figure 4.** Comparison of maximum, average and minimum width ACL models at the original insertion site when the knee is subjected to the kinematics shown in Figure 2a. a) Contact area comparison. b) Impingement force comparison.
increased contact area by 27.4%, and impingements force by 48.1%, compared to the average size ACL. The minimum size ACL decreased contact area by 31.1% and impingement force by 33.5% compared to the average size ACL.

The impingement effects from a one millimeter tibial insertion site shift, seen in Figure 5, were highly dependent on the shift direction. When comparing the average size ACL models with a one millimeter radial shift to the average size ACL model at the original insertion site, the greatest increase in contact area observed was 8.3% in the antero-lateral (AL) direction. The largest decrease in contact area observed was 31.6% in the postero-medial (PM) direction. The largest increase in impingement force observed was 41.7% in the AL direction. The greatest decrease in impingement force observed was 42.6% in the anterior (A) direction. When comparing the maximum size ACL model with a one millimeter radial shift to the average size ACL model at the original insertion site, an increase in contact area was observed in all directions. The largest increases observed were 34.3% in the AL and 32.1% in the L directions. Large increases in impingement force were also observed from this configuration, corresponding to increases of 101.37% in the AL, and 92.0% in the L directions. When comparing the minimum size ACL model with a one millimeter radial shift to the average size ACL model at the original insertion site, decreases in contact area were observed in all directions. The largest decreases were observed in the PM direction. Impingement force also decreased in all directions, with the largest observed as a 60.7% decrease in the PM direction.

For the tibial insertion site simulations with three millimeters of shift produced the most pronounced effects on ACL-intercondylar notch impingement. When comparing the average size ACL model with a three millimeter radial shift to the same model at the original insertion site, the greatest increases in contact area observed were 70.7% in the L and 54.4% in the AL directions. The greatest decrease in contact area observed was 70.0% in the PM direction. The largest increases in impingement force observed were 155.4% in the AL and 132.8% in the L directions. The largest decrease in impingement force observed was 85.0%.

### Tibial Insertion Site Shift Results

**Contact Area**

- Max ACL Size
- Avg ACL Size
- Min ACL Size

**Impingement Force**

- Max ACL Size
- Avg ACL Size
- Min ACL Size

**Figure 5.** Tibial insertion site shift results. Comparison of the contact area and impingement force data at the end of the time step with a knee flexion angle of 44.8°, valgus angle of 10.0°, and an ER angle of 29.1°. Data is shown for the tibial insertion site configurations with radii \( r \) of 1 mm, and 3 mm for the three ACL graft size models.
in the PM direction. When comparing the maximum size ACL model with a three millimeter radial shift to the average size ACL model at the original insertion site, the largest increases in contact area observed were 100.4% in the L and 88.5% in the AL directions. The largest decrease in contact area observed was 45.4% in the PM direction. The largest increases in impingement force observed were 242.9% in the AL, and 201.2% in the L directions. When comparing the minimum size ACL model with a three millimeter radial shift to the average size ACL model at the original insertion site, the largest decrease in contact area observed was 97.4% in the PM direction. The greatest decrease in impingement force observed was 99.8% also in the PM direction. A large increase in impingement force of 90.8% was observed in the AL direction for the minimum size ACL model compared to the average size model at the original insertion site. This is interesting as it shows that significant impingement increases may be possible even with a small ACL graft if it is not properly located on the tibial plateau.

The impingement effects from a one millimeter shift in the femoral insertion site, seen in Figure 6, were also highly dependent on the shift direction. Comparing the average size ACL models with a one millimeter radial femoral insertion site shift to the average size ACL model at the original insertion site, the greatest increase in contact area observed was 22.0% in the antero-proximal (APr) direction. The largest decrease in contact area was 23.3% in the postero-distal (PD) direction. The largest increase in impingement force observed was 31.3% in the APr direction. The greatest decrease in impingement force observed was 27.8% in the PD direction. Comparing the maximum size ACL model with a one millimeter radial shift to the average size ACL model at the original insertion site, an increase in contact area was observed in all directions. The largest increase was 67.2% in the APr direction. This configuration also produced the greatest increase in impingement force, reported at 97.3%. When comparing the minimum size ACL model with a one millimeter radial shift to the average size ACL model at the original insertion site, decreases in contact area were observed in all directions. The largest decrease was observed in the PD direction.

**Femoral Insertion Site Shift Results**

![Femoral Insertion Site Shift Results](image)

Figure 6. Femoral insertion site shift results. Comparison of the contact area and impingement force data at the end of the time step with a knee flexion angle of 44.8°, valgus angle of 10.0°, and an ER angle of 29.1°. Data is shown for the femoral insertion site configurations with radii (r) of r = 1 mm, and r = 3 mm for the three ACL graft size models.
Compared to the one millimeter shift, simulations with three millimeters of shift in the femoral insertion site enhanced the impingement effects between the ACL and the intercondylar notch. Comparing the average size ACL model with a three millimeter radial shift to the same model at the original insertion site, the greatest increase in contact area observed was 112.0% in the AP direction. The greatest decrease in contact area observed was 69.2% in the PD direction. The largest increase in impingement force observed was 292.6% in the AP direction, and the largest decrease in impingement force was 73.3% in the PD direction. When comparing the maximum size ACL model with a three millimeter radial shift to the average size ACL model at the original insertion site, the largest increase in contact area observed was 168.0% in the AP direction. The largest decrease in contact area observed was 100.0% in the Distal (D) direction. The largest increase in impingement force observed was 346.2% in the AP direction. When comparing the minimum size ACL model with a three millimeter radial shift to the average size ACL model at the original insertion site, decreases in contact area and impingement force were seen in all but the AP direction. A three millimeter shift in the AP direction increased contact area by 27.8%, and impingement force by 113.8%. The largest decrease in contact area observed was 69.9% in the PD direction. The greatest decrease in impingement force observed was 81.6% also in the PD direction.

The same kinematic data were applied to the notchplasty models, and impingement force and contact area were monitored. It was determined that five millimeters of notchplasty eliminated all impingement for the maximum size ACL model with a three millimeter lateral tibial insertion site shift, while four millimeters of notchplasty eliminated all impingement for the maximum ACL size with a three millimeter antero-proximal femoral insertion site shift.

The effects of combined femoral and tibial insertion site shifts were also of interest. One simulation was performed using the maximum size ACL graft with the tibial and femoral insertion sites exhibiting the greatest impingement (femur = 3 mm AP, tibia = 3 mm L). This simulation produced a contact area of 62.1 mm², which is 245.0% greater than the average size ACL model at the original insertion site locations. The impingement force was an astounding 273.0N, which is 678.3% greater than the average size ACL graft at the original insertion site locations. Combined tibial and femoral insertion shifting does not result in cumulative impingement. In fact, the contact area is less than the cumulative amount (femoral = 48.22 mm², tibial = 41.41 mm², combined = 62.1 mm²), while the impingement force was greater than the cumulative amount (femoral = 156.5 N, tibial = 105.7 N, combined = 273 N). Notchplasty was also performed for this configuration, and five millimeters of notchplasty was sufficient for eliminating all impingement between the ACL and the intercondylar notch.

4. Discussion

After sufficient validity of the FE model had been established, different configurations of ACL graft size and both tibial and femoral insertion site locations were introduced. It was expected that larger ACL grafts as well as lateral tibial and anterior femoral insertion site shifting would lead to increased contact area and impingement forces, as these shifts would move the ACL closer to the intercondylar notch wall.

The impingement on the intercondylar wall is of interest as this is thought to be a major factor influencing ACL injury [8,12,32,34]. In an animal study by Fung et al., cell death was reported to occur at the impingement site of impinged ACL's compared to unimpinged ACL's [49]. A similar study performed by Gyger et al. suggests that apoptosis has a role in ligament rupture [50]. This evidence supports the notion that intercondylar notch impingement may weaken the ACL and could be a leading cause of ACL tears.

What is most interesting about the results for the average size ACL is that a three millimeter shift of the tibial insertion site in both the AL and I directions increased impingement force by over 100%, and increased contact area by over 70%. Shifting the tibial insertion site in the PM and M directions had the opposite effect, decreasing both the impingement force as well as the contact area. Also, for the average size ACL model, femoral insertion site shifts of three millimeters in the anterior and antero-proximal directions increased impingement force over 200%.

Compared to the average size ACL, the maximum size ACL model increased the impingement force as well as the contact area between the ligament and the intercondylar notch wall of the femur. The greatest change in impingement force due to tibial insertion site shifting was seen in the maximum size ACL model with a three millimeter AL insertion site shift. This configuration increased the impingement force by over 200%, and increased the contact area by over 100% compared to the average size ACL model with no insertion site shifting. The greatest change in impingement force due to femoral insertion site shifting was seen in the maximum size ACL with a three millimeter antero-proximal shift. This configuration produced an increase of 346.2% in impingement force compared to the average size ACL model at the original insertion site location.

The results indicate that impingement force and contact area are both highly sensitive to minor differences in graft size and both tibial and femoral insertion site location. The results may suggest that notchplasty be performed to reduce the possibility of adverse interactions between the ACL and the intercondylar notch as insertion site variations are inevitable. As an attempt to quantify the amount of notchplasty required to eliminate impingement, notchplasty was simulated on several high impingement configurations. For the knee joint kinematics used in this investigation, it was determined that five millimeters of notchplasty removed all impingement for the highest impingement configurations.

In some simulations with relatively high levels of impingement, the ACL was in contact with the femoral intercondylar notch at the start of the simulation. This result may be useful for orthopedic surgeons performing ACLR. If the ligament is already in contact with the intercondylar notch after surgery, an applied external tibial rotation and/or valgus loading will only increase this interaction.
A limitation of this study is that only one knee kinematic scenario was explored on one subject. However, the kinematic data used simulate high impingement motion associated with athletic cutting maneuvers and ACL injury [7,32,47]. Other kinematic conditions such as hyperextension are of extreme interest in terms of ACL impingement. Future studies should investigate knee joint hyperextension to determine how different kinematic conditions affect ACL impingement and how variation in the ACL attachment sites could affect the post ACL reconstruction injury. Along with this, ligament prestrain was not included in this model, unlike the group’s previous work [39–41]. This was done in order to compare results with the model from Park et al., which did not include ligament prestrain [7]. However, to understand how the inclusion of prestrain would affect the results, a five percent prestrain was incorporated into the average ACL at the original insertion site. This led to a 2.8% difference increase in contact area compared to the simulation without prestrain. Because the five percent prestrain did not affect the results significantly, excluding prestrain was considered valid in this investigation.

Three millimeters of variation during ACL insertion site location should be considered minimal, yet probable. The results show that these minor variations may enhance the contact interactions between the ACL and the femoral intercondylar notch. However, the reported contact area values are relatively small. Because of this, minor changes in contact area between the configurations investigated may produce large percent differences. This should be considered to prevent a misinterpretation of the results. However, the results show that certain configurations do lead to increased impingement between the ACL and the intercondylar notch wall. Because of this, several recommendations can be made for this subject. If notchplasty is to be performed, four millimeters to five millimeters may sufficiently eliminate ACL-intercondylar notch impingement. Also, if the surgeons are to err on the side of caution while locating the ACL graft insertion sites, avoiding anterior and proximal femoral insertion site locations, as well avoiding anterior and lateral tibial insertion site locations may be recommended.

This study provides a methodology for analyzing how variations in ACLR affect ACL-intercondylar notch impingement for an individual subject. In the future, these methods could be applied to other individuals to determine the optimal configuration of ACL graft size and insertion location to minimize ACL impingement during ACLR. Future studies may also investigate the effects of other surgical variations, such as ACLR technique on ACL impingement. Understanding how the differences in ACL graft size and insertion site placement affect the ACL-intercondylar notch interactions is extremely important. The data provided from this investigation, and future similar studies, have the potential to improve outcomes of surgical procedures, reduce medical costs, and improve patient satisfaction.

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References


