The Effect of Mechanical Varus on Anterior Cruciate Ligament and Lateral Collateral Ligament Stress: Finite Element Analyses

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abstract

The current study analyzed changes in anterior cruciate ligament (ACL) and lateral collateral ligament stress as a result of mechanical varus. In an exploratory pilot study, progressive mechanical varus was introduced to a male finite element model of the lower limb at different knee flexion angles. Nine situations were analyzed (combinations of 0°, 30°, and 60° knee flexion and 0°, 5°, and 10° varus). The ACL stress was measured via changes in section force, von Mises stress, and fiber stress. Lateral collateral ligament stress was measured via changes in section force. For all 3 measures of the ACL, maximum stress values were found in extension, stress decreased with flexion, and the effect of varus introduction was most significant at 30° flexion. With 60° flexion, varus introduction produced a decrease in section force and von Mises stress and a small increase in fiber stress. In all situations and stress measures except fiber stress at 60° flexion, stress was concentrated at the posterolateral bundle. For the lateral collateral ligament, the introduction of 5° and 10° varus caused an increase in section force at all degrees of flexion. Stress in the ligament decreased with flexion. Mechanical varus of less than 10° was responsible for increased ACL stress, particularly at 0° and 30° knee flexion, and for increased lateral collateral ligament stress at all degrees of flexion. Stress was mostly concentrated on the posterolateral bundle of the ACL. [Orthopedics. 2016; 39(4):e729-e736.]

Anterior cruciate ligament (ACL) injury is the most common knee ligament injury. The ACL provides primary restraint to anteriorization forces and secondary restraint to varus forces acting together with the lateral collateral ligament. Varus and ACL injury may occur because of primary varus (bone), physiologic ligament laxity, acute traumatic injury of the lateral ligaments, or chronic lateral instability as a result of loosening of the lateral and posterolateral ligamentous structures (double or triple varus). Correction of lateral and posterolateral instability can be achieved through ligament repair or reconstruction that may be associated with tibial valgus osteotomy.

The magnitude of the relationship between varus and ACL load is not well known. No studies have investigated in detail how different degrees of varus and knee flexion increase ACL load that can lead to increased risk of acute injury or progressive loosening.

In an in vivo biomechanical study, tension on the anteromedial band of the ACL increased in the presence of varus or valgus and axial load at 20° flexion in patients without ligament injuries. Chaudhari and Andriacchi used a finite element model to evaluate the effect of varus on the knee in extension that is capable of causing lateral opening of the joint and subsequent ACL injury. However, load exerted directly on the ACL was not assessed. Understanding the effect of varus on the ACL may help

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the clinician to determine the need for additional procedures (reconstruction of the posterolateral corner and/or tibial osteotomy) combined with ACL reconstruction.

In the finite element method, the object of study is segmented into small units known as finite elements, each of which has individual biomechanical characteristics. A major advantage of the finite element method is that it offers a noninvasive method to simulate different situations and repeat them as many times as desired.\(^5\)

The use of finite elements to study biologic structures has increased in recent decades, with solutions that include the nonlinearity and complexity of the deformation of such structures.\(^6\) Accurate models can be created from magnetic resonance imaging, computed tomography, electromyography, and radioscopic data.\(^7\text{–}^{11}\)

The current study used the finite element model to evaluate the effect of mechanical varus on ACL stress in a limb subjected to a static load at different limb flexion angles during gait. The study also evaluated concomitant changes in lateral collateral ligament stress.

**Materials and Methods**

A 3-dimensional finite element model was created of a right lower limb (pelvis, thigh, leg, and foot) with 3 joints (hip, knee, and ankle) of a man 1.75 m tall and weighing 75 kg. The explicit solver of the Virtual Performance Solution software package (ESI Group, Paris, France) was used for all simulations.

The knee was composed of a bony structure (distal femur, proximal tibia and fibula, patella), a cartilaginous structure (femur, tibia, and patella), menisci (medial and lateral), ligaments (anterior and posterior cruciate, medial and lateral collateral, and medial and lateral patellofemoral), and tendons (quadriceps, patellar, and hamstring).

The anatomy of the model was created from magnetic resonance images, and the biomechanical characteristics used in the model were based on experimental data available in the literature.\(^12\) A previously developed model in the European project “Knee-up” was adapted for use in the current study (Figures 1-2).\(^12\text{–}^{13}\)

Bone structures were meshed with rigid surface elements because of their small strain compared with soft structures and the insignificant difference in results shown by considering the nonlinear behavior (2%) of the model structure.\(^14\)

The articular cartilage was characterized by a linear isotropic elastic behavior.

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**Figure 1:** Anterior (A) and posterior (B) views of the knee model.

**Figure 2:** Anterior (A) and posterior (B) cruciate ligaments on posterior view.
with Young’s modulus \( E = 0.012 \text{ GPa} \) and Poisson’s ratio \( \nu = 0.45 \) and was firmly attached to the bone.\(^{1,12,15,16}\) This characteristic was chosen because of the elastic response of the cartilage to activities such as walking and climbing stairs, during which the load frequency is greater than 1 Hz.\(^{17}\)

The menisci were composed of extracellular matrix and circumferential fibers. The extracellular matrix was considered to have an isotropic linear elastic stiffness of 0.008 GPa and Poisson’s ratio of 0.45.\(^{15}\) A 15% proportion of fibers was considered, and the fibers were composed entirely of collagen, with a linear elastic modulus of 1 to 2 GPa in tension.\(^{18}\) Thus, menisci with stiffness of 0.194 GPa in the direction of fibers were obtained.\(^{19}\) These were attached to their tibial insertions.

The quadriceps, patellar, and hamstring tendons were represented by parallel bars with linear elastic material. They were characterized by a tensile stiffness of 0.660 GPa.\(^{19}\)

The medial and lateral collateral ligaments were composed of a mesh of longitudinal and transverse fibers with linear elastic properties (stiffness of 0.122 kN/mm for the medial collateral ligament and 0.0581 kN/mm for the lateral collateral ligament).\(^{12,20,21}\) The medial and lateral patellofemoral ligaments were composed of 2 bands each, and their stiffness was 0.016 kN/mm and was evenly distributed.\(^{22}\)

The ACL and the posterior cruciate ligament were modeled in more detail because they constituted the focus of the study. They were composed of an extracellular matrix (82%) and fibers (collagen type I, 18%).\(^{12}\) The matrix was modeled as a linear isotropic elastic material with a linear elastic modulus of 0.008 GPa in tension and compression and Poisson’s ratio of 0.45.\(^{11,15}\) The fibers had a Young’s modulus (elastic modulus) of 1.3 GPa.\(^{23-26}\) The same methods and biomechanical characteristics used to construct the ACL were used for the posterior cruciate ligament. The ACL was represented by its 2 bundles (anteromedial and posterolateral), and the posterior cruciate ligament was represented by a single bundle. The ligaments were meshed with hexahedra: the ACL was composed of 620 nodes and 369 elements, and the posterior cruciate ligament was composed of 280 nodes and 156 elements. All materials were considered to be free of initial strain in the 30° position.

An axial load was applied to the limb with the knee in 3 degrees of flexion (0°, 30°, and 60°) and 3 degrees of varus (0°, 5°, and 10°). Thus, 9 situations were evaluated. The condition 10° varus was selected for the study because the current indication for surgical treatment is a grade III lesion of lateral instability (which corresponds to 10° mechanical varus or a 10-mm lateral opening).\(^{27-29}\)

For the study, the following simplifying assumptions were made: (1) 0° varus (ie, neutral axis) was defined by linear alignment between the center of the femoral head, the center of the knee, and the center of the ankle; (2) 30° flexion and 0° varus corresponded to the unstressed ACL; and (3) only passive forces were evaluated. No muscle loads opposing varus forces were applied.

The axial load applied was the weight of the model minus the weight of the right lower limb (ie, 640 N).

At first, for 30° flexion and 0° varus, the minimal static forces in the patellar-tendon couple were calibrated in 3 steps to maintain equilibrium under body weight loading.

Then 0° flexion and 60° flexion were applied in the initial 30° position with a quasi-static simulation. During flexion, the tibia and the foot remained fixed in the floor and the femur and the pelvis were pulled into the defined location (moving freely in the sagittal plane, with the x- and z-axes allowing a possible rollback motion) while the knee joint was constrained in its other rotations (internal-external and varus-valgus). That way, the leg model was positioned in space to align the hip and ankle joints vertically (Figure 3).
The axial load was applied under the head of the femur with the tibia and foot fixed. The pelvis and femur could move freely in the sagittal plane (x- and z-axes).

The detailed ACL response was assessed in 3 ways: through section force, von Mises stress, and fiber stress, as described later. Lateral collateral ligament stress was assessed only through section force.

The ACL was studied with 3 different measures because they provide slightly different assessments of the energy applied to the ligament. Section force is the total transmission force that results from the contribution of the elements intersected by a cutting plane, meaning the total energy applied to the ligament. The concept of von Mises stress arises from distortion energy failure theory. According to distortion energy theory, failure occurs when the distortion energy in an actual case is greater than the distortion energy in the simple tension case at the time of failure. In this way, von Mises stress is the distribution of the total energy in that specific ligament, with its own biomechanical characteristics, including fibers and matrix. It can be correlated to the failure load. Fiber stress corresponds to the stress in the fiber component alone as a result of deformation in the fiber direction, which represents the collagen fibers without consideration of the component matrix. It represents the stress that is influenced by the biomechanical properties of the fibers.

Then ACL stress was compared with ACL failure load. With the use of an average cross-sectional area of 30 mm² for the ACL, the ultimate failure load of the ACL reported by Woo et al. in 1991 was 2160±157 N for young donors (22-35 years old) and 1503±83 N for middle-aged donors (40-50 years old).

RESULTS
Table 1 shows section force on the ACL and lateral collateral ligament. These forces are larger at 0° flexion. For the ACL, introduction of varus caused an increase in section force at 0° and 30° flexion and a decrease at 60° flexion. The increase was more significant at 30° flexion. There was no significant difference between the values of 5° and 10° varus at 60° flexion.

For the lateral collateral ligament, introduction of 5° and 10° varus caused an increase in section force at all degrees of flexion. The stress in the ligament decreased with flexion. Determination of von Mises stress yielded results similar to those for section force (Table 2).

Table 3 shows fiber stress. At 30° flexion, the results were similar to those for von Mises stress. Unlike section force and von Mises stress, fiber stress increased with the degree of varus at 60° flexion.

The maximum von Mises stress occurred in the posterolateral band. The maximum fiber stress occurred in the posterolateral band at 0° and 30° flexion, similar to von Mises stress, and it occurred in the anteromedial band at 60° flexion (Figure 5).

None of the 9 configurations tested resulted in ACL stress that was greater than the ultimate failure loads reported for young donors (22-35 years old), 2160±157 N, or middle-aged donors (40-50 years old), 1503±83 N.

DISCUSSION
Varus thrust as a result of progressive stretching of lateral structures in chronic ACL lesions and acute grade III lesions of the posterior cruciate ligament structures has been recognized as a cause of persistent instability and/or ACL graft failure. However, the appropriate treatment of less severe varus instability, together with other causes of varus, physiologic laxity and primary varus, remains unclear. Understanding the effect of mechanical varus subjected to axial load at different degrees of knee flexion can help clinicians to understand these conditions.
Finite element analyses were used to study ACL stress because it is difficult to apply controlled simultaneous axial and varus load at various degrees of knee flexion in the entire lower limb (including the hip and foot) using biomechanical tests. Varus and valgus load would only provoke ACL stress when concomitant axial load is applied. The feasibility of achieving and controlling the desired degrees of varus required for the test is also questionable. This could be approached by applying progressive lateral lesions until the desired degree of varus is obtained; however, this method is imprecise. In addition, load cells have limited application because they may alter the biomechanical characteristics of ligaments during their implantation and they are inadequate to allow measurement of different loads inside the ligament. Several studies, including the current study, have developed and validated finite element knee models for ligament analysis. This study had some limitations. One was the lack of validation of the results with biomechanical tests. However, based on the numerous studies that validated the finite element method and the similarity of the current results to previous biomechanical and clinical findings, the authors believe that the current data are valid. For the knee model, some structures were not simulated with respect to all physiologic characteristics, but the results of other studies suggested that failure to consider the linear behavior of cartilage and the rigidity of bone is unlikely to interfere with the results. The effect of higher axial loads should also be insignificant. As shown by Homyk et al, maximum deviation of results in the varus

### Table 1

<table>
<thead>
<tr>
<th>Ligament</th>
<th>Flexion</th>
<th>Varus 0°</th>
<th>Varus 5°</th>
<th>Varus 10°</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anterior cruciate ligament</td>
<td>0°</td>
<td>512.262</td>
<td>860.865 (+68%)</td>
<td>1029.400 (+101%)</td>
</tr>
<tr>
<td></td>
<td>30°</td>
<td>154.211</td>
<td>560.758 (+264%)</td>
<td>822.706 (+433%)</td>
</tr>
<tr>
<td></td>
<td>60°</td>
<td>47.834</td>
<td>33.436 (-30%)</td>
<td>32.593 (-32%)</td>
</tr>
<tr>
<td>Lateral collateral ligament</td>
<td>0°</td>
<td>14.061</td>
<td>281.655 (+1903%)</td>
<td>565.916 (+100%)</td>
</tr>
<tr>
<td></td>
<td>30°</td>
<td>0.005</td>
<td>193.712 (+38x10^5%)</td>
<td>485.497 (+150%)</td>
</tr>
<tr>
<td></td>
<td>60°</td>
<td>0.008</td>
<td>180.774 (+22x10^5%)</td>
<td>436.670 (+141%)</td>
</tr>
</tbody>
</table>

### Table 2

<table>
<thead>
<tr>
<th>Flexion</th>
<th>Varus 0° Minimum, MPa</th>
<th>Maximum, MPa</th>
<th>Varus 5° Minimum, MPa</th>
<th>Maximum, MPa</th>
<th>Varus 10° Minimum, MPa</th>
<th>Maximum, MPa</th>
</tr>
</thead>
<tbody>
<tr>
<td>0°</td>
<td>0.091</td>
<td>29.795</td>
<td>0.384</td>
<td>40.366 (+35%)</td>
<td>0.509</td>
<td>48.431 (+63%)</td>
</tr>
<tr>
<td>30°</td>
<td>0.285</td>
<td>7.929</td>
<td>0.625</td>
<td>22.797 (+188%)</td>
<td>0.864</td>
<td>33.201 (+319%)</td>
</tr>
<tr>
<td>60°</td>
<td>0.130</td>
<td>3.448</td>
<td>0.149</td>
<td>2.622 (-24%)</td>
<td>0.158</td>
<td>3.259 (-5%)</td>
</tr>
</tbody>
</table>

### Table 3

<table>
<thead>
<tr>
<th>Flexion</th>
<th>Varus 0° Minimum, MPa</th>
<th>Maximum, MPa</th>
<th>Varus 5° Minimum, MPa</th>
<th>Maximum, MPa</th>
<th>Varus 10° Minimum, MPa</th>
<th>Maximum, MPa</th>
</tr>
</thead>
<tbody>
<tr>
<td>0°</td>
<td>-0.044</td>
<td>163.611</td>
<td>-0.065</td>
<td>212.29 (+30%)</td>
<td>-0.089</td>
<td>336.810 (+106%)</td>
</tr>
<tr>
<td>30°</td>
<td>-0.023</td>
<td>43.268</td>
<td>-0.083</td>
<td>125.354 (+190%)</td>
<td>-0.000</td>
<td>182.446 (+322)</td>
</tr>
<tr>
<td>60°</td>
<td>-0.451</td>
<td>16.950</td>
<td>-0.349</td>
<td>18.268 (+8%)</td>
<td>-0.346</td>
<td>20.157 (+19%)</td>
</tr>
</tbody>
</table>
and valgus moments was 2% between 725 N and 2610 N axial load, representing 1×body weight and 3.6×body weight, respectively. Hence, the current data can be considered valid for activities of daily living up to 3.6×body weight, such as walking. Although the main movement during flexion-extension is in the sagittal plane, it is accompanied by longitudinal axial rotation of the femur in relation to the tibia and the rollback mechanisms. This rotation is a result of complex interactions of the tibiofemoral articular anatomy and muscle activity that are dependent on the subject and were not adequately defined to be included in this study. Instability of external rotation and physiologic posterolateral laxity have been shown to be associated with worse outcomes in ACL reconstruction. Consequently, the significant increase in ACL stress found with mechanical varus may be underestimated in this study because the test does not simulate the typical failure mechanism of ACL and simultaneous external rotation torque was not applied, although rotational instability as a result of ACL instability was considered. Finally, the results of the current study apply only to mechanical varus as a result of lateral instability and cannot be assumed to represent changes caused by bone deformities, such as primary varus.

Stresses in the ACL, including section force, von Mises stress, and fiber stress, were maximum in extension (512, 30, and 164 MPa, respectively), similar to those reported in the biomechanical study of Markolf et al.33

Introduction of mechanical varus increased stress on the ACL in both the extended knee and at 30° flexion. Although the increase in stress was greater at 30° knee flexion, the maximum absolute values were still found for the extended knee. The effect of varus was significant at both 5° and 10° varus. Thus, ACL forces increased with varus loading and decreased with flexion between 0° and 30°. Similar findings were obtained in a biomechanical study of cadaveric knees by Markolf et al43; these authors found that the ACL was stressed mostly in extension and hyperextension. The current study also analyzed the ACL at 60° flexion; under these conditions, varus load led to a decrease in section force and von Mises stress and a small increase in fiber stress. Furthermore, stress on the lateral collateral ligament peaks in extension, when it is most taut, and decreases as the ligament slackens during flexion, primarily within the first 30° flexion, as shown in the current study and in previous studies. Therefore, lateral instability at high degrees of flexion may be more negligible relative to ACL stress than at 0° to 30° flexion. Restoration of lateral instability in early degrees of flexion should be prioritized.

Chaudhari and Andriacchi4 showed that the threshold for ACL injury is very sensitive to small changes in alignment, decreasing by 1 body weight load threshold for every 2° varus. Marouane et al45 showed that ACL force increases to 144 N at 20 Nm varus but decreases to 102 N at 20 Nm valgus. The ACL force was not related to the degree of varus, which is the parameter that is used in clinical practice. The magnitude of the increase in ACL stress in the current study, which for 10° varus with 30° flexion reached 433%, 319%, and 322%, respectively, for section force, von Mises stress, and fiber stress, emphasizes and, most importantly,
which quantifies the effect of small increases in mechanical varus.

The distribution of stress inside ligaments, including the ACL, is nonuniform and changes depending on flexion and the load applied.\textsuperscript{37,46} With respect to location, von Mises stress was located in the posterolateral bundle in all situations, whereas fiber stress at 60° flexion was located at the anteromedial bundle. Because in most situations the posterolateral bundle was subjected to higher stress, the effect of graft position on reconstruction outcomes can be discussed.

The current findings show that none of the situations studied overcame the threshold for ACL failure. These findings are similar to those of Homyek et al.,\textsuperscript{39} who showed that 27.2° varus was required for ACL failure with 25° flexion and neutral rotation. Although 10° varus is insufficient to induce acute ACL failure, this degree of varus results in a substantial increase in ligament stress that can promote failure in the long term.

Although a complete lesion of the lateral ligamentous structures increased the lateral opening by 4 mm (corresponding to 4° varus), operative treatment is usually indicated in acute cases for grade III lesions (>10 mm lateral opening, corresponding to 10° mechanical varus).\textsuperscript{27,29} For chronic lateral instability with the same amount of lateral opening and/or varus thrust, the operative indications are similar.

The findings of the current study suggest that in concomitant ACL and lateral instability, surgery should be considered for smaller degrees of lateral instability than current recommendations suggest.\textsuperscript{27,31} Less restrictive criteria for valgus osteotomy or lateral ligament reconstruction could prevent some failures of isolated ACL reconstruction.

**Conclusion**

This exploratory pilot study showed that mechanical varus as a result of ligamentous deformity of less than 10° was responsible for increased ACL stress, particularly at 0° and 30° knee flexion and for the lateral collateral ligament at all degrees of flexion. Stress was concentrated on the posterolateral bundle of the ACL. These results suggest that even mild mechanical varus negatively affects the ACL and lateral collateral ligament. Surgeons should consider corrective valgus osteotomy or lateral ligament reconstruction during ACL reconstruction, even with smaller degrees of lateral instability than current recommendations suggest.

**References**


