INTRODUCTION:
Meniscectomy is a well-known risk factor for osteoarthritis (OA) in humans. Loss of meniscal function can produce both load bearing changes as well as kinematic changes to the knee. Previous work has shown that during normal walking and stair ascent, subjects with partial medial meniscectomized (PMM) knees have the average tibial position shifted towards external rotation by approximately 5° throughout the stance phase of gait. Additionally, it has been shown that patients who have undergone arthroscopic partial meniscectomy experience an increase in peak knee adduction moment during normal walking. These studies suggest the possibility that the kinematic and kinetic changes following partial meniscectomy can lead to subsequent premature degeneration of the menisci following partial meniscectomy and subsequent repeat surgeries.

Thus, the purpose of this study was to test the hypothesis that the cyclic meniscal strains during walking are increased when compared to normal when the rotational and loading changes associated with the partial meniscectomy are introduced.

METHODS:
Meniscal strains were calculated using a three dimensional finite element model (FEM). The cartilage and menisci were each discretized into a mesh of brick elements. The bone-cartilage interface was fixed with having shared nodes between the 2D bone and 3D cartilage elements. The cartilage was assumed to behave as a linear elastic, isotropic solid with a modulus of 10 MPa and a Poisson’s ratio of 0.46. The menisci were modeled as transversely isotropic with radial and axial moduli of 20 MPa and a circumferential modulus of 140 MPa as per previous finite element models. The in-plane Poisson’s ratio was 0.2, while the out of plane Poisson’s ratio was 0.3. The meniscal horn attachments were modeled as previously described.

The geometry for the femur, tibia, femoral and tibial articular cartilage, and menisci were derived from segmented MR images to build 3-D models of the bones, cartilage, and menisci at both HS and MS for model validation. Solid meshes of the meniscal and tibial cartilage were created using MSC Patran (MSC Software, Santa Ana, CA). Solid meshes of the menisci were created using an in-house algorithm written in MATLAB (Mathworks, Inc., Natick, MA). The model consisted of 13,980 nodes and 11,655 elements. Bone was assumed to be rigid relative to the soft tissue and modeled using rigid 2D elements.

The anterior (ACL) and posterior (PCL) cruciate ligaments and meniscal (MCL) and lateral (LCL) collateral ligaments were modeled using nonlinear tension-only springs with a constant, k, of 1500. Contact was modeled between the femoral cartilage and the meniscus, the meniscus and the tibial cartilage, and the femoral and tibial cartilage for both the medial and lateral sides resulting in six contact pairs.

Model validation was conducted using a test bed that allowed prescribed displacements (flexion angles, anterior-posterior (AP) displacements, and internal-external (IE) rotations) to be applied to the tibia relative to a femoral reference frame. A human cadaveric knee (male, 57 yrs) was harvested and used for the experiment.

Two specific time points: heel-strike (HS) and mid-stance (MS) during the gait cycle were used to prescribe the primary and secondary motion of the knee (flexion) with using the cadaveric test bed, and the FEM of the knee was placed in positions representing HS and MS and imaged using magnetic resonance (MR) for model validation. A compressive load of 700N was distributed through the medial and lateral femoral condyles along the axial direction during the entire loading cycle to represent a load of approximately 1 body weight.

The hypothesis was tested by comparing the effects of a kinematic shift on the strains within the meniscus after a 5° external rotation (Rot) was applied to the tibia to replicate the kinematic change reported following meniscectomy. To look at the consequence of an increased adduction moment, the ratio of medial/lateral compressive load was varied from 50:50 to 70:30. Finally, maximum strains in the lateral and medial menisci were computed for the 5° rotation with the 50:50 load and 70:30 load distributions.

RESULTS:
Spatial comparison of the predicted displacement field of the model with experimental measurements indicate a close match between the FE predicted field and the experimental field found in the MRI of the cadaver (Figure 1).

The peak strains were higher in the rotational shift case (Table 1). In each case, the peak strains were highest in both menisci at the anterior and posterior horns. The adduction moment change did not appear to change the peak logarithmic strains in either meniscus.

<table>
<thead>
<tr>
<th>Ext Rotation</th>
<th>Med:Lat Load</th>
<th>0º</th>
<th>5º</th>
<th>0º</th>
<th>5º</th>
</tr>
</thead>
<tbody>
<tr>
<td>Max Strain (Lateral)</td>
<td>0.19</td>
<td>0.48</td>
<td>0.19</td>
<td>0.47</td>
<td></td>
</tr>
<tr>
<td>Max Strain (Medial)</td>
<td>0.83</td>
<td>0.88</td>
<td>0.82</td>
<td>0.88</td>
<td></td>
</tr>
</tbody>
</table>

Table 1. Peak maximum principal logarithmic strains at mid-stance.

DISCUSSION:
The results support the hypothesis that increased external tibial rotation during the gait cycle following meniscectomy can lead to increased strains within the menisci. Strains in the meniscus appear to be more sensitive to rotational changes than an increased adduction moment. Combining the kinematic and kinetic changes leads to nearly the same increases in strains as just the kinematic change by itself. Although the increase in the peak maximum principal logarithmic strain is higher in the lateral meniscus than the medial meniscus in both cases of rotation, it is important to note that the overall peak strains are still about twice as high in the medial meniscus than the lateral meniscus. While the assumptions of this study will likely influence the magnitude of the strains predicted by this model, the conclusions regarding the relative increases in meniscal strains after introducing kinematic changes predicted by the model seem consistent with clinical results that show further meniscal damage after meniscectomy is common.

In conclusion, a kinematic change as small as 5° of external tibial rotation following a partial meniscectomy can be the catalyst that leads to meniscal degradation, loss of meniscal load bearing, and ultimately knee OA. Although an increased peak knee adduction moment during walking has been shown to be present in subjects with medial compartment knee OA and several studies have shown that with increasing age and levels of osteoarthritis, meniscal tears also increase in frequency, these results suggest that it may not be an increased adduction moment that leads to meniscal tears but rather that the adduction moment may increase as a result of the subsequent cartilage degeneration.

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REFERENCES: