DEVELOPMENT OF A FINITE ELEMENT TOOL FOR STRESS ANALYSIS OF A SURGICALLY ALTERED FEMUR

Anthony Au*, Adrian Liggins, James Raso**, and Alidad Amirfazli***
***Department of Mechanical Engineering, University of Alberta, Edmonton, AB, Canada**
****Glenrose Rehabilitation Hospital, Edmonton, AB, Canada**

This paper reports the development of a tool for analyzing stresses in a surgically altered femur. The three-dimensional FE model incorporates a novel approach to implementing orthotropic and heterogeneous bone properties and non-uniform distributed loading. The model contains cortical, cancellous, and subchondral bone incorporating experimentally determined mechanical properties to characterize the anisotropy and heterogeneity of the bone. Ligaments and muscles of the joint are represented to more fully describe the loading condition. Use of the tool is demonstrated for an anterior cruciate ligament (ACL) reconstruction with button-type fixation and the knee joint at full extension. The stresses at the tunnel aperture on the femoral cortex produced from the fixation were noticeable for low levels of loading. Forces from the ligaments and muscles had only slight influence on the stresses at the aperture. Repeated compression of the femoral cortex at these stress levels may cause microdamage to the cortex eventually resulting in fatigue failure.

KEY WORDS: femur, knee reconstruction, finite element analysis, button, stress.

INTRODUCTION: A tool was recently developed enabling the creation of finite element (FE) models of the knee bones which improved upon the state of the art (Au, 2003). This tool has potential application for a better understanding of post-operative anterior cruciate ligament (ACL) reconstruction bone stresses. Surgically created tunnels from ACL reconstruction cause stress deprivation in the surrounding bone and lead to bone weakening. In addition, the potential mechanical risks of using peripheral button-type graft fixation, characterized by repetitive compression of the femoral cortex, has not been examined with regards to possible bone fracture at the fixation site. This tool addresses deficiencies of past FE models by giving special consideration to the incorporation of realistic geometry, material properties, and loading to provide an improved analysis of the bone stress states. Predicted bone resorption patterns from stress analysis of a femur FE model for an ACL reconstruction is discussed.

METHODS: A 3D reconstruction of a composite femur (Sawbones, Pacific Research Labs, Vashon Island, WA) was used to model the bone geometry (Greer, 1999); the model is available on the Internet at the International Society of Biomechanics Finite Element Repository (ISB, 2001). For computational efficiency, only the distal 9 cm of the femur were used; longer lengths do not affect stress distributions (Au, 2003). The model was exported to ANSYS (Swanson Inc., Houston, PA) for stress analysis.

Cortical and cancellous bone structures were represented as linearly elastic, orthotropic, and heterogeneous, with material properties assigned based on comprehensive experimental data (Rho, 1992). Heterogeneous bone was implemented by dividing the femur into 25 volumes (4 cortical and 21 cancellous) and mapping the moduli of elasticity directly from the experimental data (Fig 1). Orthotropy was implemented by assigning 9 elastic constants to each volume: 3 elastic moduli, 3 shear moduli, and 3 Poisson's ratios. Mapped shear moduli of cancellous bone were not available, therefore the shear moduli for each volume were determined from empirical relationships (Ashman et al., 1989). Orthotropic Poisson's ratios of cancellous bone also were not available, therefore transversely isotropic Poisson's ratios were assumed (Williams and Lewis, 1982). A layer of subchondral bone was modeled at the distal femur (Fig 1); its properties were assumed as isotropic and homogeneous due to scarcity of experimental data. The bone was characterized as having the mechanical properties of 45-year-old bone.

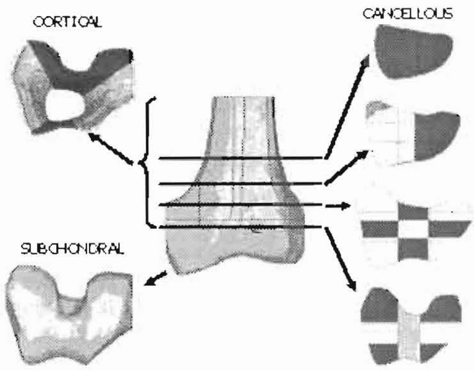

Figure 1: The bone divisions within the femur FE model. There are four transverse sections of cancellous bone, four quadrants of cortical bone, and one section of subchondral bone. Colours are for graphical clarity to distinguish various regions of the bone with different properties, but are not representative of material property values.

The model was meshed with 2 mm tetrahedral elements resulting in approximately 800,000 nodal degrees of freedom. A convergence analysis validated the adequacy of the mesh (Au, 2003). A loading case modeling the knee in full extension was studied. The joint surface was loaded with a non-uniformly distributed compressive force of approximately three times body weight, reflecting experimentally determined contact patterns (Fukubayashi and Kurosawa, 1980). The contact patterns were incorporated by digitizing and matching the experimental and model geometries; a program written in MATLAB (Mathworks, Natick, MA) was used to detect the nodes in each pressure region and to assign the distributed loading. Ligament and muscle forces were incorporated using similar methods. Tensile forces in the posterior cruciate ligament (PCL) and medial and lateral collateral ligaments (MCL and LCL, respectively) were uniformly distributed over experimentally determined attachment areas (Harner et al., 1995; Meister et al., 2000). The PCL was assumed to have anterolateral and posteromedial bands. The MCL was assumed to contain superficial and deep bands. In such cases of multi-banded ligaments, forces were divided among the bands and uniformly distributed within the insertion area of each band. Forces were similarly modeled in the gastrocnemius (medial and lateral), popliteus, and adductor magnus muscles. Figure 2 shows a schematic of the load conditions assigned to the femur; Table 1 summarizes the magnitudes of the muscle and ligament forces assigned. This realistic approach to incorporating loading conditions is rarely done in FE models (Crowninshield et al., 1976). To approximate physiological ligament and muscle forces occurring during full extension, tensile forces acting during contralateral heel strike of the gait cycle (Shelburne et al., 2004) were simulated.

Table 1 Summary of the ligament and muscle forces assigned to the femur FE model.

Ligament/Muscle	Force (N)
PCL	5
LCL	5
MCL	5
Gastrocnemius (medial)	478
Gastrocnemius (lateral)	478
Popliteus	5
Adductor Magnus	46

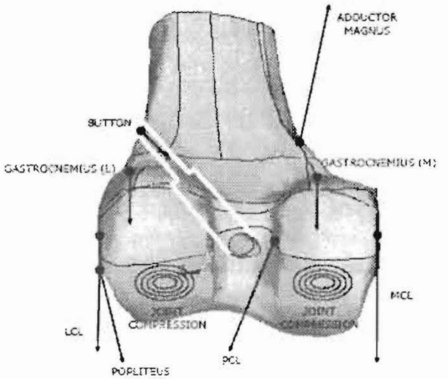

Figure 2: Schematic of loading condition for femur FE model. Dots illustrate approximate attachment area of forces. Three ligaments were represented (LCL, MCL, PCL); 3 muscles were represented (gastrocnemius, adductor magnus, popliteus). The solid white line depicts the bone tunnel; peripheral fixation button forces (BUTTON) were placed around the aperture of the tunnel. Non-uniform distributed loading was placed on the femoral condyles; the ellipses represent the pressure contour borders.

The ACL reconstruction was represented using two coaxial tunnels: a guide tunnel of 4 mm diameter and a distally enlarged tunnel of 8.5 mm diameter (Fig 2). The tunnels were rotated 35 degrees relative to the coronal plane and 30 degrees relative to the sagittal plane. A peripheral fixation button was assumed to contact the femoral cortex at 4 surface points and pulled on by a 200 N force directly along the tunnel axis.

RESULTS AND DISCUSSION: Stresses slightly above 100 MPa were observed at the tunnel aperture as a result of button-type fixation loading on the cortex (Fig 3a). At 100 MPa, the fatigue cycle of cortical bone is 106 cycles; over a 10-year period, approximately 107 cycles are applied to bone tissue (Swanson et al., 1971). The high stress levels resulting from a 200 N load typically seen in gait are expected to increase in more physical activities. Thus, repeated cortical compression at these stresses over a 10-year period may fracture the bone underneath the button and compromise the reconstruction. However, one must also consider that microdamage in the bone is constantly being repaired by the body. In addition, the assumed 200 N graft force occurring at full extension will vary during gait and affect the results.

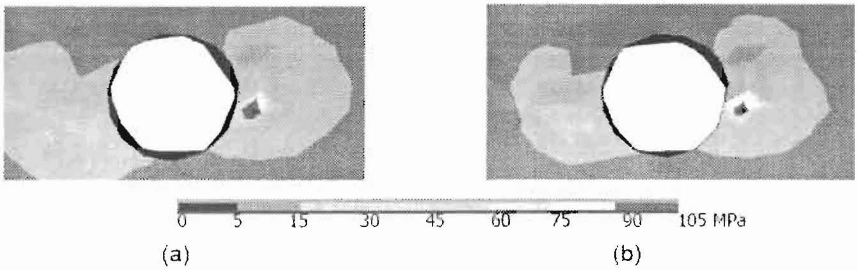

Figure 3: A contour plot of von Mises stresses at the cortical bone tunnel aperture is shown. The plot is of a plane view of the stress field. Stress distributions are shown for model (a) with ligament and muscle forces and (b) without ligament and muscle forces.

The effect of including ligament and muscle forces on cortical stresses was also investigated. Figure 3b shows the stress distribution at the tunnel aperture when tensile forces from ligament and muscles were not included and only joint and graft forces were considered. Inclusion of the ligament and muscle forces increased the maximum cortical stress by 5%. The small change

in cortical stresses suggests that, in this case, the bone stresses caused by the ligaments and muscles are fairly localized, and the cortical stresses at the tunnel aperture arise mainly from the button-type fixation and joint compression. In a recent investigation, we predicted that the cortical stresses at the tunnel aperture were very similar for 45- and 65-year-old bone without accounting for the influence of ligament and muscle forces (but with otherwise identical load conditions) (Au et al., 2004). The results presented here confirm the results of that investigation; stress distribution around the tunnel aperture in the lateral cortex will be very similar for bone between the ages of 45 and 65. As such, the stress contour shown in Fig 3a is representative for that age group.
Exclusion of the mechanical influence of cartilage and menisci of the knee limit the possible clinical value of this model.

CONCLUSION: The tool presented here provides a more comprehensive approach to the analysis of the post-ACL reconstruction stress state of the knee than was previously available. The model combines a realistic 3D geometry of the femur with anisotropic and heterogeneous bone. This tool has the potential to become a valuable assistant in the design of peripheral fixation buttons. In addition, the ease of bone property incorporation enables this model to be used for studies of aging in relation to button fixation.

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