Technical Brief

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INTERNAL FORCES OF THE FEMUR: AN AUTOMATED PROCEDURE FOR APPLYING BOUNDARY CONDITIONS OBTAINED FROM INVERSE DYNAMIC ANALYSIS TO FINITE ELEMENT SIMULATIONS

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Introduction

The capacity for improvement of finite element models to accurately simulate internal body stresses and strains is dependent on several issues that include the accuracy of the underlying geometry models, methods for meshing that geometry, appropriately defining the model material properties, and accurately applying the model boundary conditions. Computed Tomography scans have also been used to extract patient specific material properties associated with the underlying bone geometry [1-3] further increasing the specificity of individualized models.

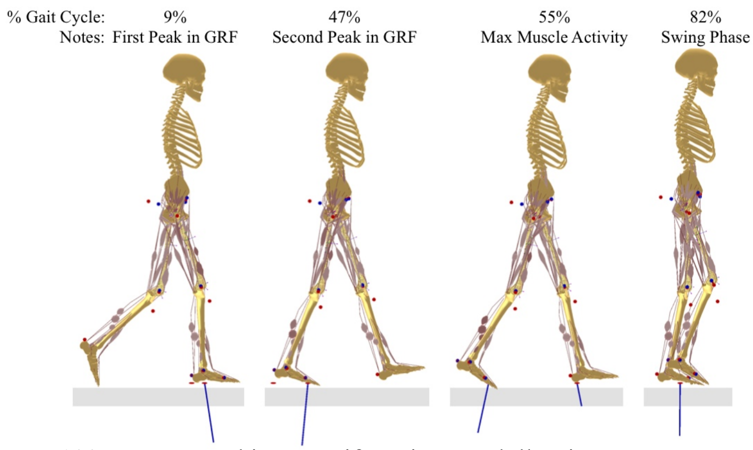
By utilizing validated inverse dynamics based musculoskeletal models for automatically defining the boundary conditions for finite element models, FEA problems can be defined more quickly and for internal body structures undergoing realistic exertions. The realization of such a process has widespread applications, including potential progress toward developing suitably acceptable models for use in the clinical environment. In this paper, a procedure, available in the form of a software program, for automatically extracting the results from a rigid-body based musculoskeletal simulation to define the boundary conditions of a finite element model of a bone segment is presented. The methodology is used to analyze the strains, stresses, and deformations of a femur during a nominal gait cycle.

Materials and Methods

Mesh Generation & Material Property Assignment

A finite element model of a human femur was created using Mimics v12 (Materialise, Leuven, Belgium). A mesh consisting primarily of ~2mm tetrahedral elements with excluded mid-side nodes was imported into ANSYS Workbench v11 (ANSYS Inc, Canonsburg, PA, USA). The complete model of the human femur consisted of 89,891 tetrahedral elements (Ansys Element Type, Solid185 – 4 node linear tet) with 18497 nodes.

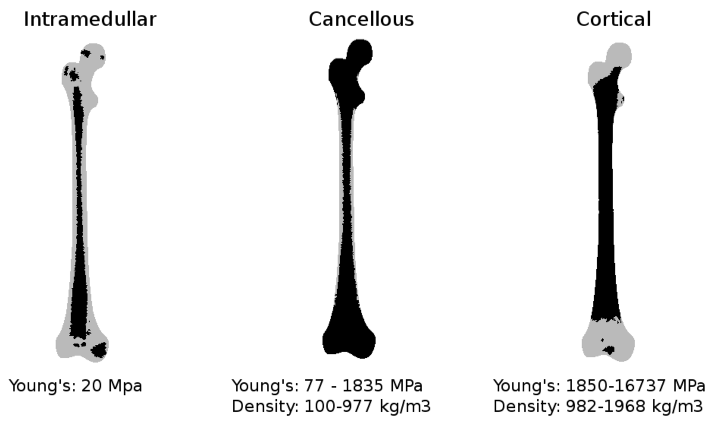
The material property of each tetrahedral element was defined using a procedure similar to that used by Peng et al. [4]. The densities were assigned by linear interpolation between 100 kg/m^3 and 2000 kg/m^3 and correlated with a range of Hounsfield Units (HU) determined from the CT scan. The extracted material properties were modeled as linear elastic and isotropic. Elements with a HU of below



100 were merged into a uniform intramedullar tissue element type and given an elastic modulus of 20 MPa. The remaining elements were assigned elastic moduli calculated from apparent densities using axial loading equations developed by Lotz et al. [5]. A Poisson's ratio of 0.30 was used for all materials.

Musculoskeletal Model

Muscle force magnitudes were determined from the output of the AnyBody System musculoskeletal modeling gait application [6]. To facilitate comparison with previous studies [7,8], 4 stages in the gait cycle were chosen for analysis. The four specific times in the gait cycle used for analysis (9%, 47%, 55% and 82%) were selected using the same criteria as Duda et al. [8]. The rigid body AnyBody femur model consisted of 28 connected 'via-point' muscles and one wrapped muscle that was used to model the Iliopsoas muscle group. All the muscles were modeled as 3 element hill-type muscles with the default parameters defined in the AnyBody Repository v7.0.



Boundary Conditions

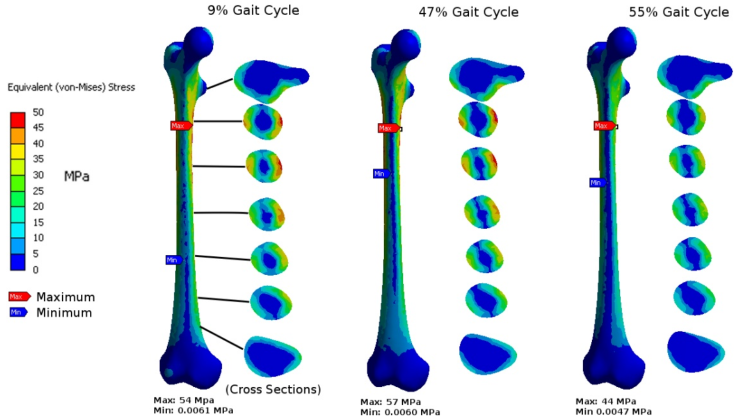
The positions of the muscle forces applied to the FE femur model were calculated using the anatomical representation of the corresponding nodal positions used in the AnyBody musculoskeletal body model. The finite element model of the femur (derived from CT scan data) was scaled and oriented to match the rigid body representation of the AnyBody femur. The automated procedure allowed for the muscle forces to be applied at the same positions in both the AnyBody model and the FE model while maintaining the force equilibrium, calculated during the inverse dynamics analysis used to solve for the AnyBody muscle forces, in the FE model. The joint reaction forces at the hip and knee were applied to the FE model in a similar manner as described for the muscle forces. Since the sum of the muscle forces and joint reactions were in equilibrium, there was no need to artificially constrain the position of any node or geometry feature of the FE model.

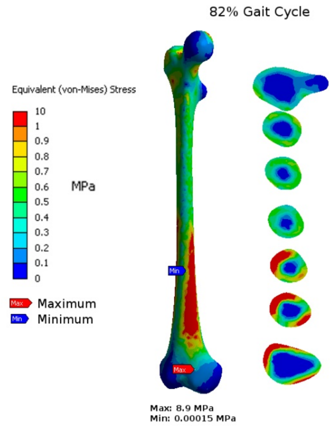
Results

Axial cross sections of the long-axis of femur were analyzed to determine the stress gradients between the different surfaces of the bone. At the second peak in GRF (47% of the gait cycle) the stresses in an axial cross section located 4 cm distal of the lesser trochanter resulted in a gradient with a maximum stress of 54 MPa on the medial surface of the bone, a maximum stress of 47 MPa on the opposing lateral surface, and a minimum stress of 23 kPa in the intermediate intramedullary tissue. Similar trends in the axial stress gradients were observed progressing distally along the bone away from the hip joint with increasing reductions in stress magnitudes for both the medial and lateral stresses at each axial cross-section taken suggesting a higher concentrated loading closer to the hip joint than the knee joint.

Discussion

The automated procedure described in this paper for incorporating simulated muscle forces as boundary conditions to a patient-specific finite element femur model was used to compare the strains, stresses, and deformations along the long axis of the bone with other published results. Although a perfect match was not found between the results presented here and other published findings, the proposed method shows substantial tractability in being able to effectively combine the results from validated musculoskeletal rigid body models with patient specific derived FE models for novel exertions. Future efforts should be directed in better utilizing available methods for defining subject specific geometry used for the FE model as well as assessing the effect of the slight nodal displacements imparted to the rigid body model on the simulated muscle forces. Additionally, an optimization procedure to best match the derived patient bone geometry to the rigid body model to achieve an optimal match would be beneficial. Future applications of the presented methodology should also focus on utilizing the results from the FE model (i.e. displacements) to drive/adapt the inverse dynamics motion for accurate simulations of more flexible internal structures. We hope this tool can be utilized to further develop coupled models utilizing the best aspects from rigid body dynamics and





finite element modeling simulations.

A trial version of the software that implements the methodology described in this paper is available at http://www.ozeninc.com/Any2Ans.

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For more information on the Any2Ans, ANSYS, AnyBody or Mimics softwares used in this paper or for consulting inquiries contact Ozen Engineering:
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