Transactions on Biomedicine and Health vol 2, © 1995 WIT Press, www.witpress.com, ISSN 1743-3525

Finite element analysis of the human tibia

B.V. Mehta, S. Rajani
Department of Mechanical Engineering, Ohio University, Athens, OH 45701, USA

Abstract

A 3-D solid model of the human tibia and the fibula was constructed using Magnetic Resonance Imaging and solid modeling software. A finite element analysis of the tibia was conducted to evaluate stresses developed in the tibia under static loads and to study the effect of varying material properties on these stresses. Loading conditions and material properties used were taken from literature. Two finite element models were taken into consideration. A model of the tibial post upto a length of 130mm was studied to compare results to previous literature and a model of the whole tibia under similar loading conditions was analyzed. Maximum stresses developed for cancellous bone were within ultimate stress values and a tendency of the cancellous bone to distribute stresses to regions of compact or cortical bone was observed.

1 Introduction

Recent research in the field of automotive crash analysis and its effect on the human body, has brought us to a point where injury to the upper human body has been eliminated to an enormous extent. This has been achieved mainly by designing cars incorporated with air bags. In the event of a car crash, it has now been discovered that the human lower leg is another part of the human body which is severely effected. It is the tendency of the driver of the vehicle to depress the brake pedal just before impact. This action results in the lower leg taking on most of the impact. This research was aimed at developing a 3-D model of the tibia and the fibula and studying various loading conditions using varying material properties. Most of the work done so far has been on studying the knee joint (tibio-femoral joint) and as far as the tibia is concerned, modeling of the tibial plateau and tibial post have been reported. A real time 3-D model of the whole tibia and its analysis has not been reported in the literature till this date.

The leg is made up of two bones, namely, the medial tibia and the lateral fibula. An interosseous membrane fills the gap between the tibia and the fibula. The proximal portion of the tibia is made up of the lateral and medial condyles, which articulate with the lateral and medial condyles of the femur. A projection known as the tibial porosity serves as the point of attachment of the patellar ligament to the tibia. As one goes lower, the tibia narrows into what is called the shaft of the tibia, and then expands again towards the proximal end. At the proximal end of the tibia, the medial malleolus articulates with the ankle joint. The fibula is a long slender bone, which has its head articulating with the tibia. It however, does not take any part in the knee joint. The lateral portion of the fibula extends into the lateral malleolus, which extends beyond the medial malleolus of the tibia.

One of the earliest models reported of the tibia, was that developed by Hayes *et al.* in 1977 [1]. They developed an axisymmetric finite element model of the lateral tibial plateau. Three principal regions were defined in the model, namely, articular cartilage, compact bone and trabecular bone. A resultant load of 445N was applied to the region of contact of the femoral condyles with the articular cartilage. Results indicated that most of the load applied to the cartilage surface of the tibia, was distributed to the compact bone of the tibial diaphysis, through the subchondral trabecular bone. In the case of trabecular bone, high compressive stresses developed directly below the region of applied load and shear stresses developed beneath the edge of the region of applied load. Murase *et al.* (1981) developed an axisymmetric model of the proximal tibia and the tibial component [2]. A finite element analysis was performed on the model based on nonaxisymmetric loading conditions.

Little *et al.* (1986) developed a three dimensional finite element model of the upper tibia [3]. This model was developed mainly to provide a base for further analysis involving modeling of prosthetic resurfaced tibiae. The model geometry was developed by taking sections of a tibia, from a 73 year old male cadaver. The average size of the tibia had a mediolateral width of 74 mm and an overall length of 130mm. All materials in the tibia were assumed to be linearly elastic and isotopic. Mechanical properties for cortical bone and cartilage were taken from literature. An indentation test was performed to determine the non-homogeneous characteristics of cancellous bone. This study indicated a factor of safety of between eight and twelve, on occurrence of maximum stresses developed in the cortical and cancellous bone, during normal stance. The metaphyseal shell was assumed to have a uniform wall thickness of 1.5mm with properties of cortical bone.

Transactions on Biomedicine and Health vol 2, © 1995 WIT Press, www.witpress.com, ISSN 1743-3525

The objective of this research was to develop a 3-D solid model of the bones (tibia and fibula) constituting the human lower extremities, using images taken through the Magnetic Resonance Imaging (MRI) Technique. A study of the images obtained using this approach, indicated a clear distinction between different materials constituting the lower leg, particularly the bones when compared to images from ultrasound and CT-scans. This was a basis of achieving accuracy in developing the model, both, in terms of geometry and material properties. In addition, the model of the tibia was to be used to carry out finite element analysis, in order to understand its behavior under various loading conditions. This model, being the only one developed so far, of the whole tibia, would form a basis for future studies (displacement, stress, strain and prosthetic analysis) involving the human lower leg.

2 Computer Modeling of the Tibia and Fibula

Two techniques were looked at to obtain cross sectional images of the lower leg, namely, ultrasound imaging and magnetic resonance imaging. Ultrasound waves are very effective when it comes to penetration of tissues. However, it is not too effective in the penetration of air and bone. As a result of this, although a distinct boundary between regions of compact bone and muscle was observed, a boundary between regions of compact bone and cancellous bone could not be identified. Magnetic resonance images on the other hand proved to be very effective in displaying distinctly the various regions in a cross section of the human leg. The MRI images for this study were obtained from the lower left leg of a male. A total of 23 images were obtained. The concentration of the images obtained was higher at the proximal and the distal ends of the lower leg when compared to the middle region. This was done to capture the major feature changes that are present at the proximal and the distal ends of the tibia. The distance between each section was known and the total length of the lower leg scanned was 443.5 mm.

Two softwares packages were used for the purpose of image processing and data collection. A software known as Aldus Photostyler was used to scan the images and transfer them onto a 3.5" floppy in a Tag Image File (TIF) format. OPTIMAS, an image processing software was used for extraction of the x and y coordinates of the cortical and cancellous regions of the tibia. The distance between the sections provided the required z coordinate for the purpose of 3-D modeling. The 3-D solid model of the tibia and the fibula was developed using Intergraph's Engineering Modeling Software and Parametric Programming Language. The total length of the tibia modeled was 443.5mm from the proximal end to the distal end. The definition of the various regions of bone (compact and cancellous) and geometric characteristics of this model were found to be extremely accurate. The thickness of the compact bone shell for the tibia was found to gradually decrease as one proceeds from the distal

Transactions on Biomedicine and Health vol 2, © 1995 WIT Press, www.witpress.com, ISSN 1743-3525

to the proximal end and diminishes towards the tibial plateau. The 3-D model of the tibia and the fibula is shown in fig.1.

fig.1 3-D Model of the Tibia and the Fibula

3 Finite Element Models of the Human Tibia

Two FE models were developed. The first model took into consideration 130mm length of the tibia from the tibial plateau and the second model was developed for the entire length of the tibia. Each of the above models were analyzed using one-material, three-materials and four-materials for different areas of the bone. Patran3 was used for the purpose of mesh generation, application of material properties and the application of boundary conditions. A loading condition of 2450 N was applied to the articular cartilage region of the model. This loading condition was taken from literature (Harrington *et. al.*)[4] and occurs during normal gait in the stance phase at full extension. This load was distributed equally over the articular cartilage at regions where the femoral condyles came into contact with the tibial plateau. The four-material models constituted of 4 material properties namely, articular cartilage, compact or cortical bone, intramedullary trabecular (cancellous) bone and subchondral trabecular (cancellous) bone. The Young's Modulus of the cartilage layer and compact bone were taken as 11.6 MPa and 17.2 Gpa respectively (Little *et. al.*)[3]. Poisson's ratio for both the regions was taken as

0.3 from the same literature. Material properties for trabecular bone were the same as those used by Hayes *et. al.* in their study[1]. Young's Modulus and Poisson's ratio for subchondral trabecular bone were taken as 700 N/sq.mm and 0.3 respectively and those for intramedullary trabecular bone were taken as 350 N/sq.mm and 0.3 respectively.

3.1 Tibial Model Entirely Constituting Compact Bone

The results of this finite element model of the tibia when analyzed within I/FEM, conformed extremely well with results obtained by previous researchers. This model of the tibia was assumed to be made up entirely of compact bone. The maximum stress developed in the tibia was 17.37 N/mm^2. This compressive stress occurred on the lateral side at a distance of approximately 60 mm from the distal end of the tibia. Studies performed by previous researchers on the upper tibia reported a maximum compressive stress of 24.77 N/mm^2 at the posterior surface. The maximum stress values obtained in the upper tibia in this study was approximately 8 N/mm^2. The stresses developed were comparatively lower since this model was made up of compact bone. The study performed by Little *et. al.* [3] considered the model made of a compact bone shell, and cancellous bone filled the area within the compact bone shell. The maximum displacement that occurred was in the area of the applied load and took place along the longitudinal axis of the tibia. A maximum displacement of approximately 0.6 mm was observed directly beneath the region of applied load. Moreover, a deflection of approximately 3mm of the tibia occurred.

3.2 Model of the 130mm Tibial Post Considering 3 Materials

The materials considered in this case were the cartilage layer, compact bone and subchondral cancellous bone. Analysis of this model revealed the development of higher stresses in regions of compact bone when compared to cancellous bone. A study of the stress patterns of the model indicated a distribution of stresses by cancellous bone to regions of compact bone. A maximum compressive stress of 12.48 N/mm^2 was observed at a distance of approximately 30mm from the distal end of the model on the lateral side of the tibial post and in the region of the neck of the tibia on the lateral side. This occurred in the region of compact bone. Moreover, tensile stresses were observed at the tibial plateau indicating a possibility of tibial plateau fracture which is a common orthopaedic problem. Fig. 2 shows the distribution of stresses at a cross section of the tibial post.

The maximum displacement that occurred in the model was 1.686mm in the region of the cartilage layer just beneath the applied load. This compares to a value of approximately 1.5 - 3 mm reported by Little *et. al.* [3] in their

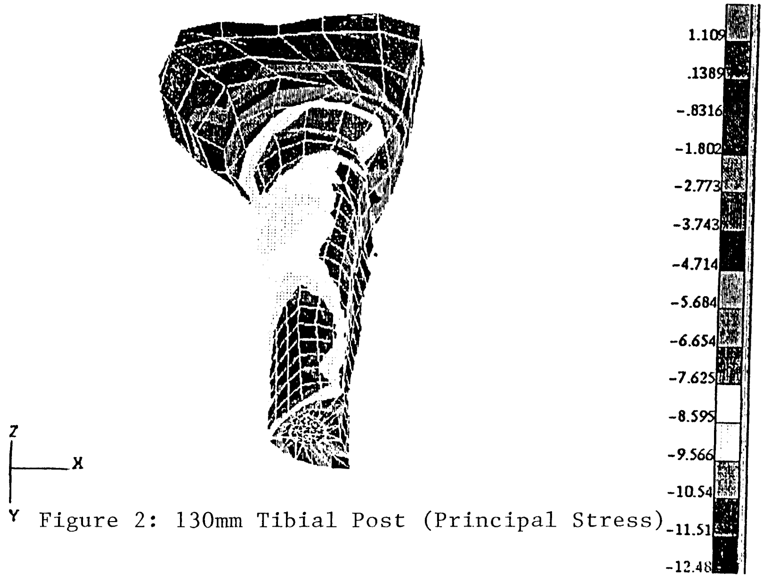

Figure 2: 130mm Tibial Post (Principal Stress)

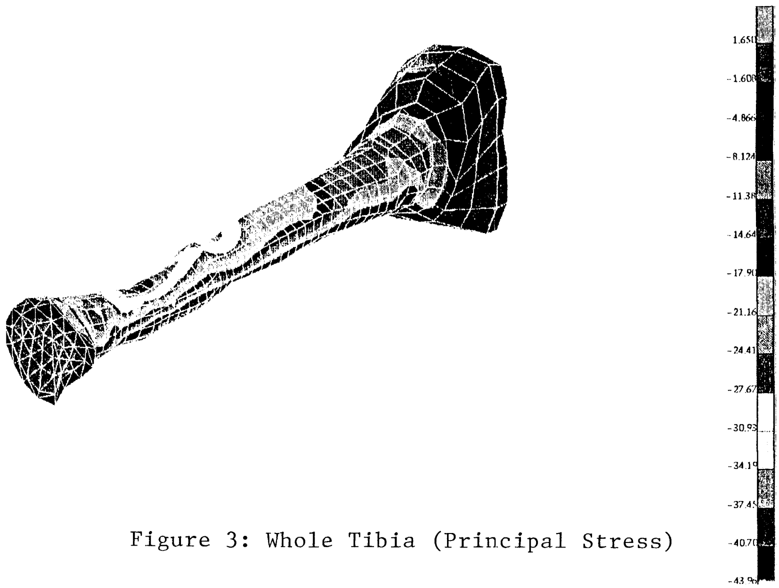

Figure 3: Whole Tibia (Principal Stress)

Transactions on Biomedicine and Health vol 2, © 1995 WIT Press, www.witpress.com, ISSN 1743-3525

study under similar loading conditions. Moreover, a deflection of approximately 0.5mm was observed in addition to a lateral displacement of the cartilage layer.

3.3 Model of the 130mm Tibial Post Considering 4 Materials

The results of the static nonlinear analysis of the tibial post modeled with 4 materials namely the cartilage layer, compact bone, subchondral trabecular (cancellous) bone and intramedullary trabecular bone indicated a slight change in the behavior of the tibia under similar loading conditions.

Maximum compressive principal stress values increased to 12.59 N/mm^2 in the region of compact bone when compared to a value of 12.48 N/mm^2 in the previous model. However, the region of the occurrence of these principal stresses did not change drastically. The change in material properties did not affect the displacement or deflection of the tibia to a significant degree. These values also compare to principal stresses of approximately 4 n/mm^2 developed in the compact region of an axisymmetric model of the tibial plateau developed by Hayes *et. al.* [1]. The maximum load applied in the above case was 445N.

The stresses developed in these models were comparatively higher when compared to the stresses developed in the whole model of the tibia considering compact bone. This shows the importance of considering both compact bone and cancellous bone while modeling. However, a change in material properties of cancellous bone does not seem to make a significant difference to the behavior of the tibia under external forces.

3.4 Results of the Analysis of the whole Tibia

A maximum longitudinal nodal displacement of 1.826 mm took place in the region of the articular cartilage just below the applied load for the model developed using 3 materials. Nodal displacements gradually decreased as one proceeded from the proximal to the distal end of the tibia and were considerably low towards the distal end. However, an interesting observation was a deflection of 7.954 mm of the tibia in the lateral direction. This behavior of the tibia, which is a long bone of the human body can be compared to the similar behavior of another long bone, the femur. A deflection of 2-3mm of the femur was reported by Rohlmann *et. al.* [5]. The loads applied in his study were less when compared to those of the tibia in this study.

The maximum principal stress occurred at approximately 70 mm from the distal end of the tibia as shown in figure 3. A compressive stress value of 43.35 N/mm^2 was observed in the region of compact bone in the lateral region of the tibia. Moreover, higher stresses were observed in the lateral region when

Transactions on Biomedicine and Health vol 2, © 1995 WIT Press, www.witpress.com, ISSN 1743-3525

compared to the medial region of the tibia. The ultimate strength for compact bone is 193 N/mm^2. This provides a factor of safety of 4.5 for compact bone. The maximum principal stress occurring in the region of cancellous bone is approximately 12 N/mm^2. When compared to the ultimate strength of 39 N/mm^2 reported by Little *et. al.* [3], this provides for a factor of safety of 3.3.

The behavior of the tibia under the influence of four materials as compared to three was similar to the model of the tibial post with four and three materials. A slight increase in stress values was observed in the later model. Stress values of 43.96 N/mm^2 occurred as compared to a value of 43.35 N/mm^2 in the earlier model. Deflection results rose to 8.069mm and a maximum nodal displacement of 1.839mm was observed.

4 Conclusion

The deflection and stress values developed for the 130 mm model compared well and was within limits of those found in literature. Considerable stresses developed in the region even well below 130mm from the tibial plateau. This could be attributed to the geometry of the model in addition to loading and boundary conditions. However, this does prove the importance of consideration of the whole model of the tibia when conducting studies related to the lower extremities of the human body.

5 References

1. Hayes, W. C., Swenson L. W. and Schurman D. J., "Axisymmetric Finite Element Analysis Of The Lateral Tibial Plateau", Journal of Biomechanics, Vol. 11, 1978, pp. 21-33.
2. Murase, K., Crowninshield R. D., Pederson D. R. and Chang, T., "An Analysis Of Tibial Component Design In Total Knee Arthroplasty", Journal of Biomechanics", Vol. 16, 1983, pp. 13-22.
3. Little, R. B., Wevers, H. W., Siu, D. and Cooke, T. D. V., "A Three-Dimensional Filite Element Analysis of the Upper Tibia", Journal of Biomechanical Engineering, Vol. 108, 1986, pp. 111-119.
4. Harrington, I. J., " A Bioengineering Analysis of Force Actions at the Knee in Normal and Pathological Gait," J. of Biomedical Engineering, Vol. 11, 1976, pp.167-172.
5. Rohlmann, A., Bergmann, G. and Koelbel, R., "The Relevance of Stress Computation in the Femur with and without Endoprosthesis", Finite Elements in Biomechanics, ed. Gallagher, R. H., Simon, B. R., Johnson, P. C. and Gross, J. F., New York, 1982, pp. 361-375.