MICROMOTION ANALYSIS IN BONE IMPLANTS

Sidi Berri, New York City College of Technology; Gaffar Gailani, New York City College of Technology

Abstract

In the internal fixation of long bones, a plate with a number of screws is attached to the bone in the fracture site. The micro-motion between the screws and the plates affects the stability of the implants as well as the healing process. The magnitude of this motion is believed to have influence on the formation of the fibrous tissue around the implants, which is assumed to be one of the causes of the loosening of implants. Current experimental techniques do not provide the data needed for evaluating this micro-motion. Limited work has been done to develop a full 3D-scale Finite Element (FE) model to study this phenomenon. Imported models from Computer Aided Design (CAD) packages cannot address the non-linearity issues associated with contact, as well as the complex geometry. The best solution would seem to be the development of an FE model within the FE software that can address a complicated issue such as contact. Usually FE software doesn't have good sketching tools such as CAD software. In this paper, a new approach to constructing this model is proposed. The new approach requires discretizing the geometry and the finite element mesh. Results are obtained for different types of loads, including cyclic external loads and static compressive preloads. The model can accommodate Dynamic Compression Plates (DCP), Locked Compression Plates (LCP), and Low Contact Dynamic Compression Plates (LC-DCP). A meshing technique based on the nodes shared in the boundaries was adopted to mesh the large number of surfaces produced by 180 volumes. Rigid Links were created to represent the locking feature of LCP. The FE model has shown many advantages, such as flexibility in geometry and quantifying results that were predicted theoretically. The model can easily be modified to fit many geometrical or other design changes without remeshing, which may aid in design optimization. In addition, the model can analyze the micro-motion between the implants.

Introduction

Internal fixation of long bones using plates has been practiced for more than 100 years. The method has become increasingly popular since World War II. In the early stages of fracture repair the function of an internal fixation procedure was to immobilize the fracture fragments. This allowed bony union to proceed [1]. The internal fixation of fractures, Fig (1), has evolved in recent decades with a change of emphasis from mechanical to biological priorities. Recently, internal fixation underwent a basic evolution. The new objective is to achieve maximum stabilization with minimum damage to the blood supply during fracture repair. Biological internal fixation needs to be improved based on today's understanding of biology, biomechanics and mechanics in order to provide better and safer fracture treatment.

The concept of biological internal fixation is still developing. There is controversy surrounding the rigidity of the plates that are currently in use. In addition, there are incidents of bone refracture, screw and fatigue failures. There is a need for research, as well as the development of structural analysis tools to evaluate current and future internal fixation plate designs. The goal is to produce a plate that has the required strength to promote fracture healing so as to hinder bone remodeling. The contradiction exists because there is a need for extremely rigid fixation during the healing of fractures, and less rigid fixation during later bone remodeling. This remains an enigma in orthopedic surgery.

Numerous models have been used to study the internal fixation of long bones. Simon et al. [2] used very simplified 1D, 2D &3D models, concluding that these models needed improvement to quantify the stresses and strains in critical areas. Ganesh et al. [3] utilized a 2D FE model to introduce a new design of plates based on graded stiffness. Numerous 2D FE models have been developed for the analysis of plated and non-plated long bones. Inherent in all these studies were unknown errors associated with a 2D approximation of a complex 3D problem. Cordey et al. [4] proposed the composite beam-theory approach to analyze the bone and the plate as a composite beam. The composite beam theory has its own limitations, such as deficiency in analyzing the micro-motion between the plate and the bone.

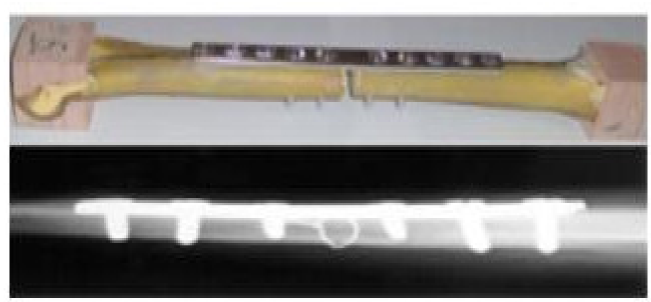

Figure 1. Experimental and real-life models of bone fracture repair.

Many new plate designs were introduced recently such as: Locked Plates (see Fig 2), Point Contact Fixators (PC-Fix), Stiffness Graded; and Biodegradable. Most of these designs, if not all, were not associated with FE analysis in 3D. The 3D analysis was important to quantify the improvement with respect to the conventional plates, as well as sketching guidelines for the direction of the development of new designs.

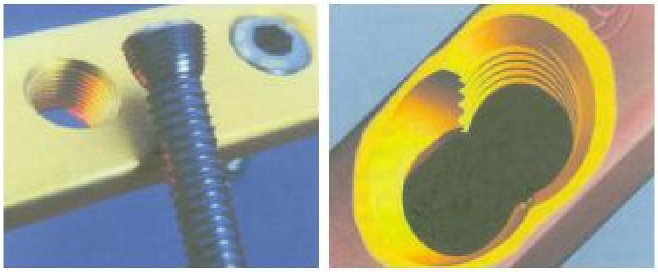

Figure 2. Experimental and real-life models of bone fracture repair, Cordey et al. [4].

Methods

The complexity of the geometry is a fact that the FE model needs to consider. The geometry of the whole assembly has been discretized numerous times (as shown in Fig 3), in order to reduce computational costs, and to create a flexible geometrical model that can be modified to match the design of different implants. The discretization will create more structural mesh that is well distributed over the model, and help in the meshing algorithm, where contact surfaces need to be selected. Automatic Dynamic Incremental Nonlinear Analysis software (ADINA) was used for the FE analysis.

In Fig (3) the whole discretization of the geometry is shown. The prism volumes were used to fit the conical shape of the screw head, as well as the whole screw shaft. Four cylindrical coordinate systems and one Cartesian coordinate system were used to create the Discretized Geometrical Model (DGM).

The material properties for the plate and screws are the same. Steel or titanium may be used to make the plate and screws. It should be noted that there are many published papers on the study of the mechanical properties of bones. The bone is treated as an orthotropic or transverse isotropic material due to the variation of the module of elasticity (E) in the longitudinal, and the transverse transversal directions of the bone. It is important to mention that bone is slightly viscoelastic, under low frequencies, and its measured Young's modulus is to some extent strain-rate dependent [5].

The meshing process was established so that it could accommodate the numerous contact surfaces, as well as the continuity of the model components. A specific meshing technique based on Nodal Coincidence Checking was applied to mesh the model.

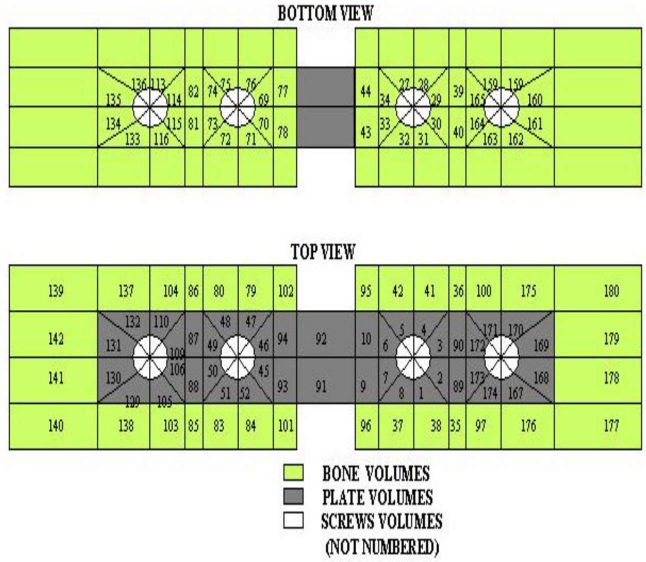

Figure 3. Top and bottom views of the DGM. The numbers show the volume number for each component.

In Fig (4) a brief idea of how the contact surfaces will be meshed in 3D is introduced. The surface of V1, S1, is in contact with the surface of V2, S1. S4 in V1, and S2 in V3 are identical because both of them belong to the screw. Therefore, it is expected that the meshing algorithm should have the two separate surfaces of the screw, S1 and S2, sharing the same nodes. The contact between the plate and the bone is demonstrated by the contact of S4 in V2 and S2 in V4. The meshing algorithm should allow no sharing of nodes between these two surfaces. There will be two nodes at every point to allow contact in the initial configuration. The same procedure will be applied for the contact between the screw and the bone. As a check, after the completion of the mesh, two nodes should be detected at the boundary of V1 and V2. Notice that P1 and P2 of V1 have the same coordinates of P2 and P1 of V2. Three nodes should be detected at P1 and P2 of V3 because these coordinates are shared among the former, P3 and P4 of V2, and P1 and P2 of V4. If the contact between the screw and the bone is ignored then the number of nodes in the boundary will drop from 3 to 2 nodes. The key point is that both S1 in V1 and V2 have exactly the same coordinates, which demonstrate the initial configuration - the screw head is in full contact with the plate's hole.

The results are shown in Fig (5). At the last circular line that represents the boundary between the plate volume and the screw volume, the mesh algorithm has assigned two nodes at one point. There is a duplication of nodes. Furthermore, each node is represented by a green number at each point, and can be clearly seen by selecting the node's number where the number appears to be overlapping.

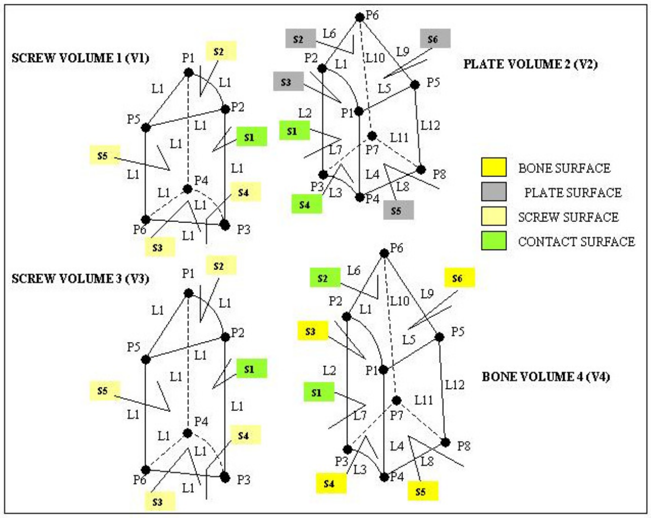

Figure 4. The meshing process with surfaces shown with their normals.

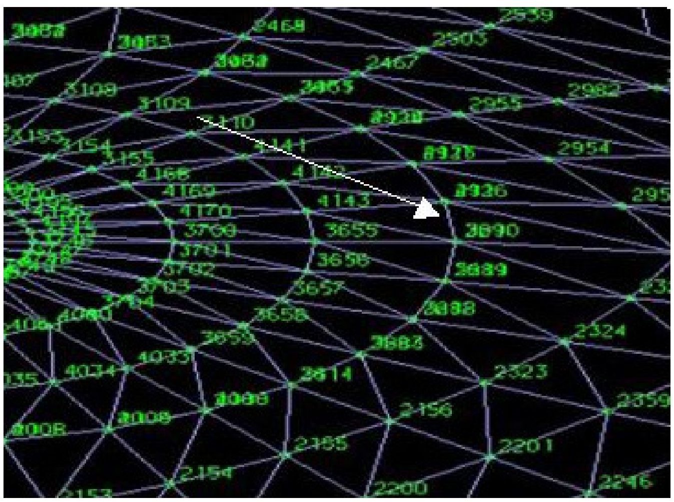

Figure 5. Verification of the final mesh for the interface of plate volumes and screw volumes. The white arrow is pointing towards the boundary of V1 and V2 (the plate volume and the screw volume).

In order to accommodate the locking feature of the LCP as shown in Fig (2), Rigid Links need to be created. As shown in Fig (6) Rigid Links are special constraint equations established between two nodes - a master node and a slave node. As the nodes displace due to deformation, the slave node is constrained to translate and rotate, such that the distance between the master node and the slave node remains constant, and that the rotations at the slave node are the same as the corresponding rotations at the master node, ADINA [6].

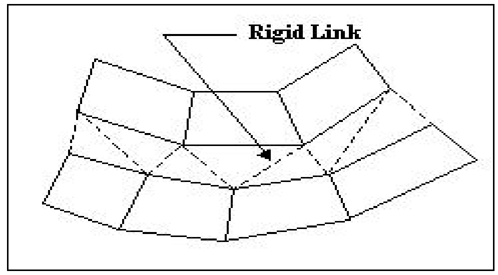

Figure 6. Mechanism of the Rigid Links.

The master and slave nodes should be decided based on the force transmitted from the bone to the implants, or vice versa. Likewise, the DCP should be modeled by choosing the contactor and target surfaces. Contact problems are nonlinear and there is no analytical solution for complex contact. The FE method is frequently used to determine the contact stresses and deformations. Static and cyclic loads might be used in the analysis. The static loads were in the preloading stage. The cyclic loads, for example bending moments and torsions, were considered to be external loads.

Results

In this communication we analyzed only the LCP and DCP plates, which are the most common techniques, though the model can be extended to a variety of plate types. LCP and DCP rely on different mechanical principles to provide fracture fixation. The results indicated that this model could be used to study the biological effects on the bone fracture fixation induced by using plates and screws, such as bone remodeling and formation of fibrous tissue around the implants. Bone remodeling is very sensitive to small changes in cyclic bone stresses that are produced by external cyclic loads.

In bending the stress in LCP plates is shown in Fig (7). It is observed in the analysis that the stresses becomes higher near the hole site. The DCP plates have a higher rate of micro-motion compared to LCP plates, where the micro-motion is constrained. Fig (8) shows the micro-motion of the plate and the screw for DCP. The plate moves more than the screw because the screw is assumed to be well fixed to the bone, while the plate is fixed only by the screw. The model simulations are available online at the ADINA website: http://www.adina.com/newsgD013.shtml .

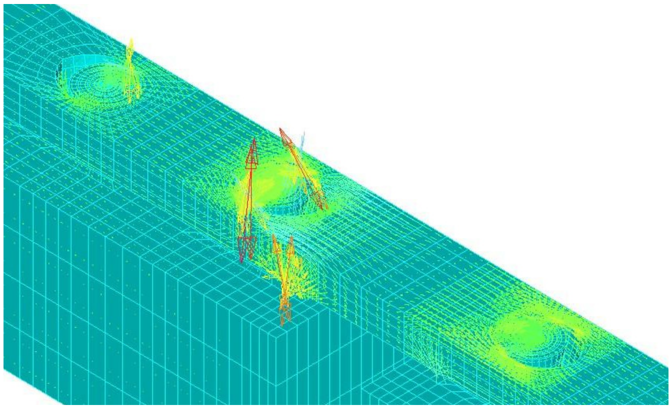

Figure 7. Stress vector in LCP subjected to bending.

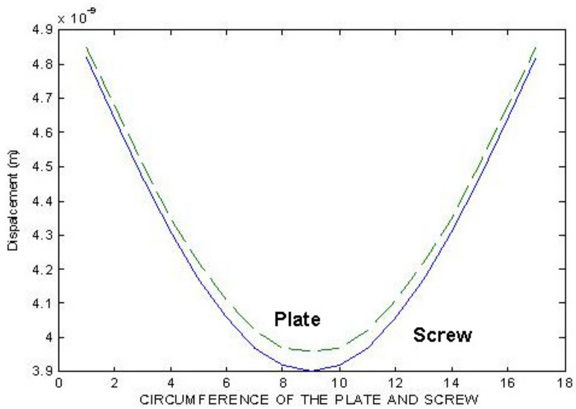

Figure 8. Micro-motion of the plate and the screw. The plot shows that the plate moves more than the screw, which is expected.

Discussion

In the late stages of fracture repair a loss of bone mass may occur. Many researchers attribute this loss to "Stress Shielding", which occurs as a result of structural adaptation of bone to reduced stress with the subsequent danger of re-fracture. One of the assumptions made is that the bone loss is a result of "plate induced osteopenia." Cheal et al. [7] concluded that disuse osteopenia should be limited to the central region between inner screws, Seabeck et al. [8]. High stresses were detected in the middle of the plate. These high stresses between the plate and the bone may affect the cells on the bond surface.

The stability of the implants is a major concern. One of the factors that affect this stability is the formation of fibrous tissue in the interface between the bone and the implants. This fibrous tissue loosens the implants and leads to implant failure. There is some uncertainty as to the reason for this growth of fibrous tissue, but shear stress between the screw and the bone is suspected to be the cause of this formation according to the studies of many authors like Buchler et al. [9]. This shear stress may originate from the bone-implant micro-motions. The in vivo study of Jasty et al. [10] showed that small amplitudes of micro-motion (≤ 20 μm) have no influence on the bone healing, but amplitudes greater than 150 μm produce a fibrous interface at a depth of 1 or 2 mm around the implants during the 6 weeks following the implantation. The FE model can predict the relationship between the micro-motion and the mechanical stimulus. This will be a subject for future communication. Fig (9) shows the loosening of implants and the region of formation of the fibrous tissue. Buchler et al. [9] investigated the formation of the fibrous tissue between the bone and the screw and came up with a numerical solution that predicted the formation of the fibrous tissue, which is stimulated by shear stress in a period of 0 – 8 weeks. Fig (10) shows the stress distribution through the longitudinal axis.

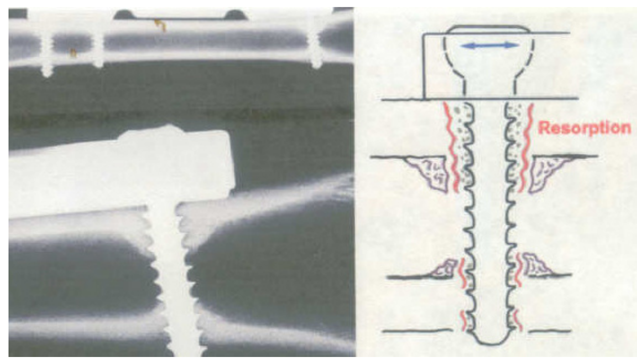

Figure 9. The effect of micro-motion on the stability of implants, Perren [1].

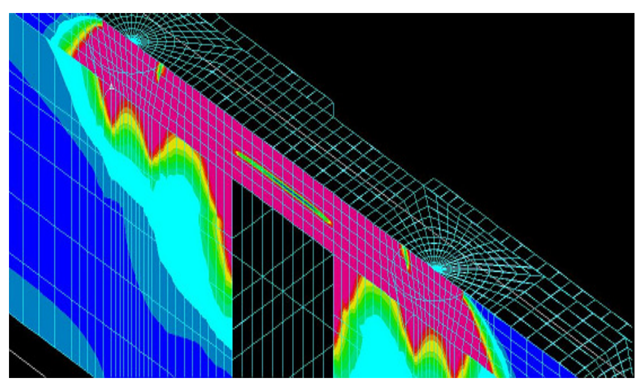

Figure 10. A sectional view through the longitudinal axis shows the shear stress distribution. High shear stress formations are observed near the screw head between the bone and the screw, surrounded by black squares. The shear stress is believed to be responsible for the formation of the fibrous tissue between the screw and the bone.

One of the advantages of this model is that it accommodates different geometries with minimal changes. For example, Fig (2) shows two types of holes for LCP. One hole fits the screw's conical head exactly, and the other hole is bigger than the head of the screw. The latter provides a surgeon with the opportunity to change the location of the screw if necessary. The DGM can easily create the two different types of holes by moving three geometrical points backward and forward. Once these points are moved there should be no need for remeshing or reassigning of boundary conditions, and the FE solution could be obtained in a reasonable time.

Summary

Three-dimensional modeling is necessary in the study of screw stresses and contact stresses between the bone and the plate. The FE model has demonstrated the complex nature of the structural analysis and design of internal fixation plates. The results of this study indicate the validity of the numerical modeling approach. The results also demonstrate an agreement with the mechanics of materials approach. This FE model is useful for a better understanding of the stability of implants. A typical application of the FE model will be important for designing new implants that can provide better and safer fracture treatment.

References

[1] Stephan M. Perren, "Evolution of the Internal Fixation of Long Bone Fractures," British Journal of Bone Joint Surgery, 2002, 84-B: 1093-110.

[2] B. R. Simon, S. L-Y Woo, G. M. Stanley, et al, "Evaluation of One-, Two-, and Three –Dimensional Finite Element and Experimental Models of Internal Fixation Plates," Journal of Biomechanics, 1977, Vol. 10, pp.79-86.

[3] VK Ganesh, K Ramakrishna and Dhanjoo N. Ghista, "Biomechanics of Bone-Fracture Fixation by Stiffness-Graded Plates in Comparison with Stainless-Steel Plates," Biomedical Engineering Online 2005, 4:46.

[4] J. Cordey, S. M. Perrren, S.G. Steinemann, "Stress Protection Due to Plates: Myth or Reality? A Parametric Analysis Made Using the Composite Beam Theory," Injury, Int. J. Care Injured 31, 2000, S-C1-13.

[5] K. J. Faran, N. Ichioka, et al., "Effect of Bone quality on the Forces Generated by Compression Screws," Journal of Biomechanics, 1992, Vol. 32, 861-864.

[6] ADINA R & D. ADINA Theory and Modeling Manual. Volume 1, 2001.

[7] Edward J. Cheal, Wilson C. Hayes, et al., "Stress Analysis of Compression Plate Fixation and Its Effects on Long Bone Remodeling," Journal of Biomechanics, 1985, Vol. 18, No. 2, 141-150.

[8] J. Seebeck, J. Goldhahn, et al., "Effect of Cortical Thickness and Cancellous Bone Density on the Holding Strength of Internal Fixator Screws," Journal of Orthopedic Research, 2004, Vol. 22, 1237-1242.

[9] P. Buchler, D. P. Pioletti, and L. R. Rakotomanana, "Biphasic Constitutive Laws for Biological Interface Evolution," Biomechan. Model Mechanobiology 1, 2003, 239-249.

[10] M. Jasty, C. Bragdon, et al. In Vivo Skeletal Responses to Porous-Surfaced Implants Subjected to Small Induced Motions. J. Bone Joint Surgery AM 79, 1997, 707 – 714.

Biographies

SIDI BERRI is an associate professor in the Mechanical Engineering Technology department at New York City College of Technology. He received his PhD from NYU Polytechnic, NY. His interests involve biomedical engineering, composite materials and nanotechnology. He can be reached at sberri@citytech.cuny.edu

GAFFAR GAILANI is an assistant professor in the Mechanical Engineering Technology Department at New York City College of Technology. He received his PhD from the City University of New York. His research work is focused on biomechanics and poroelasticity. His can be reached at ggailani@citytech.cuny.edu