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Numerical modeling of the effect of reaming on pressure and temperature in orthopaedic procedures of long bones

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**NUMERICAL MODELING OF THE EFFECT OF REAMING ON PRESSURE
AND TEMPERATURE IN ORTHOPAEDIC PROCEDURES OF LONG BONES**

By

Jonathan Bahen, B.Eng

Ryerson University, 2004

A thesis

presented to Ryerson University

in partial fulfillment of the requirements for the degree of

Master of Applied Science

In the program of

Mechanical Engineering

Toronto, Ontario, Canada, 2006

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ABSTRACT

Numerical Modeling of the Effect of Reaming on Pressure and Temperature during Orthopaedic Procedures Involving Long Bones

Master of Applied Science 2006, Jonathan Bahen

School of Graduate Studies, Ryerson University

Orthopaedic procedures often involve reaming of the intramedullary canal of the long bones to facilitate the insertion of an intramedullary device. The effect of this reaming on the pressure and temperature within the bone was investigated numerically by solving the full three-dimensional momentum equations, together with the continuity equation and energy equation, using the finite element technique. It was found that once the flutes of the reamer become clogged and congealed, the rotational rate had little effect on the pressure distribution. A temperature model for the reamer head during reaming procedures, which took into account parameters of both the bone and reamer, was implemented. The numerical simulation was found to accurately mimic the thermal results obtained during experimental procedures. Viscosity variations of marrow substitutes, which are commonly used in experiments, were found to have little effect on temperature distribution.

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NOMENCLATURE

The definition of symbols will be done in alphabetical order:

- a : Darcy coefficient
- c_p : Specific heat [$\text{J/kg} \cdot ^\circ \text{K}$]
- d : Diameter of the Reamer [mm]
- d_i : Inner Diameter of the bone [mm]
- d_a : Outer Diameter of the bone [mm]
- Da : Darcy number
- F : Axial force of the Reamer [N]
- Fr : Froude Number (ratio of inertial forces to gravitational forces)
- g : Gravitational force [m/s^2]
- k : Thermal Conductivity Coefficient [$\text{J/m} \cdot \text{s} \cdot ^\circ \text{K}$]
- L : Reference length [m]
- n : Rotational Rate of the Reamer [1/s]
- p : Pressure [Pa]
- P : Dimensionless pressure
- Pe_T : Thermal Peclet Number
- Pr : Prandtl number (ratio of momentum and thermal diffusivities)
- r : Radial direction (Radial System) [m]
- R : Dimensionless radial direction (Radial System) [m]
- Re : Reynolds number (ratio of inertia to viscous forces)
- t : Time [s]
- T : Temperature [$^\circ \text{K}$]
- T_0 : Initial Temperature [$^\circ \text{K}$]
- ΔT : Change in temperature of the Reamer Face [$^\circ \text{K}$]
- T^* : Dimensionless Temperature
- u_r : Velocity in the r direction (Radial System) [m/s]
- U_r : Dimensionless velocity in the r direction (Radial System)

u_θ : Velocity in the θ direction (Radial System)
 U_θ : Dimensionless velocity in the θ direction (Radial System)
 u_z : Velocity in the z direction (Radial System) [m/s]
 U_z : Dimensionless velocity in the z direction (Radial System)
 u_0 : Reference velocity in axial direction [m/s]
 z : Axial direction [m]
 Z : Dimensionless axial direction (Radial System)

Greek symbols

ζ : Velocity Convergence Tolerance
 λ : Thermal Conductivity Coefficient [J/m · s · °K]
 θ : Angular Direction
 κ_r : Permeability in the r direction [m²]
 κ_θ : Permeability in the θ direction [m²]
 κ_z : Permeability in the z direction [m²]
 μ : Viscosity [Pa · s]
 $\bar{\mu}$: Effective Viscosity [Pa · s]
 μ_f : Friction Coefficient
 ρ : Density [kg/m³]
 φ : Porosity
 τ : Dimensionless time

Chapter 1 : Introduction and Literature Review

This chapter will present the literary review of the current status of the research that has been conducted on Fat Embolism Syndrome (FES) and necrosis of bone tissue due to heat generation of the reamer.

1.1 Fat Embolism Syndrome

Fat embolism has been known to the medical world for almost 150 years. It was first discovered by F.A. Zenker in 1862, who found fat droplets in the lungs of a railroad worker. Ten years after this discovery the first clinical study of fat embolism syndrome was carried out. This study was a description by Von Bergman of a man who had suffered a femoral fracture and had died due to respiratory distress which may have been related to his initial injury [1]. One of the first papers that was published on the subject was by Peltier [2], who described the pulmonary consequences of fat embolism, as well as detailing the treatments that were available. It is important to distinguish between fat embolism and fat embolism syndrome. Fat embolism refers to the presence of fat globules in the lung and peripheral circulation after a long-bone fracture or major trauma. Fat embolism syndrome is a serious manifestation of fat embolism that involves progressive respiratory insufficiency, and deteriorating mental status occurring within 72 hours of injury [3].

In the late 1960's and early 1970's FES was being detected in patients who were receiving common orthopaedic procedures [4]. The overall mortality due to FES was reported to be 10% to 20% during this period [3]. This caught the attention of several

researchers who tried to decipher the causes and signs of FES. One diagnostic criterion that was based on a collection of symptoms and laboratory parameters was presented in Gurd [5]. Using this criterion, at least one major and four minor symptoms needed to be detected to definitively reveal a case of FES [4]. Although there were suggestions that Gurd criteria may under diagnose the syndrome, [6] it would be almost 15 years before a more comprehensive criterion was purposed. Lindeque *et al* [7] published a study which proposed a new criterion for detecting FES based on respiratory parameters which increased the incidence of FES for patients who undergo a variety of orthopaedic procedures that involve intramedullary reaming.

1.1.1 Rate of Incidence

The incidence of FES and mortality directly related to the syndrome vary significantly between studies. Some studies show rates of incidence as low as 0.5% [8] and 2.2% [1] and others to as high as 41% [9]. This variation is accounted for by the different methods and criteria used to detect the fat emboli. What is certain, regardless of the criteria used, is the fact that FES does occur as a results of orthopaedic procedures and it can result in serious side effects including the mortality of the patient [9][10]. Studies conducted in the 1990's reported mortality to be in the range of 5% to 10% [3][11]. This is cause for alarm due to the fact that there were 44,792 total hip and total knee replacements performed in Canada among Canadian residents in 2001-2002, representing a seven-year increase of 39% from 32,147 procedures in 1994-1995 and a one-year increase of 4% from 42,917 procedures in 2000-2001 [12]. If these trends continue, and they are expected to, FES could affect over 4,000 Canadians per year.

Although over the past 25 years the use of intramedullary reaming and nailing to repair tibial and femoral fractures has become popular and excellent results for patient recovery have been reported [8], large emboli have still been observed in the right cavities of the heart seconds following the reaming [13]. It is believed that this may be due to the highly elevated pressures caused by the reaming procedure. Intramedullary pressure elevation during reaming can vary greatly depending on a number of parameters, including the presence or absence of a fracture, the speed of the reamer penetration, and the design of the reamer [13]. Other studies have observed that the intramedullary marrow viscosity and the depth of the reamer flutes can have a profound impact on the increased intramedullary pressure [14].

The intramedullary canal contains a highly viscous medullary fat permeated by numerous arterial and venous blood vessels. Reaming allows the canal to be open to implant insertion. During the reaming process, the cutting flutes can become clogged and congealed, stopping the transportation of debris out of the bone [15]. When clogging and congealing of the reamer flutes occurs, penetration of the reamer into the medullary cavity can be compared to a piston entering a cylinder filled with a viscous fluid and the reamer displaces a volume of fluid in the cylinder [13]. This can theoretically lead to very high pressure values, as seen with hydraulic instruments [16]. As the mean blood pressure in the capillaries is only approximately 25 mm Hg, the reaming process can cause the intramedullary pressure to greatly exceed this value, at which point, fat ejection is inevitable [16]. Although the exact pressure when fat ejection occurs is not well

established, it is accepted that it will occur due to increased pressure caused by intramedullary reaming, which at times can exceed 1000 mm Hg [13].

With the problems that are associated with increases in the intramedullary pressure so evident, many researchers have conducted experiments to better understand the causes and effects of the pressure increases. Tronzo *et al* [17] concluded that the pressure increase in the femoral canal resulted from the insertion of a cemented prosthesis, and that this was the direct cause of the detected marrow embolization. In this study, pressure increases were recorded to be in excess of 300 mm Hg. Muller *et al* [18] conducted reaming in five pairs of human femora and found that there were pressure increases in the intramedullary canal of over 800 mm Hg while using blunt reamers, and 400 mm Hg with sharp reamers a clear indication that pressure increases can reach dangerous levels during normal operating procedure.

Several researchers have made attempts to reduce this unwanted pressure build-up in the bone. Schmidt *et al* [19] found that reducing the pressure was possible if a venting hole was inserted into the cortex. Although this did cause the sought after pressure drop in the intramedullary canal, the venting hole has since been proven impractical in clinical trials, as it tends to clog with the fatty marrow and bone debris caused by the reamer process [20].

Development of a procedure that does not cause elevated medullary pressure exceeding the threshold for fat embolism has not yet been obtained, although several

recommendations have been made by researchers. Muller *et al* [18] recommended that the reamers used should be handled carefully and must be changed regularly to insure that they maintain their optimal sharpness. Blunt reamers should never be used as they cause significantly higher pressures than those of sharp reamers. Ries *et al* [21] showed that slow insertion and the use of a fluted rod could greatly reduce the intramedullary pressure in the femoral canal. Even slight modifications to the instruments, such as decreasing the shaft diameter of the reamers, can have a profound effect on decreasing the pressures [23].

1.2 Thermal Injury

During orthopaedic procedures that involve reaming, surgeons must not only be aware of the risks of elevated pressures within the intramedullary canal, they also need to be concerned with the heat that is being generated by the reaming process and its possible traumatic effect on the bone. It is known that orthopaedic procedures involving reamers cause an increase in temperature inside the medullary cavity. Reaming has been known to cause adverse effects for centuries. Hippocrates himself said that reaming should be done slowly and the reamer frequently withdrawn and cooled to avoid excessive heat generation [24]. This temperature increase damages cell enzymes which are essential to adequate functioning of the individual cells and consequently for fracture healing [25]. It can also cause damage to the cortical bone, as was seen in several animal studies [26][27]. Although the exact temperature of human cortical bone death due to overheating is not known explicitly, the average temperature which has been reported to cause bone necrosis is approximately 50°C [28].

Ochsner *et al* [24][29] attempted to classify the thermal damage associated with reaming in orthopaedic procedures in an attempt to standardize the cataloging of these injuries. To simplify the classification, heat-induced damage was divided into four degrees of severity (0-3). The recommendations on the classification procedure are displayed in Table 1-1.

There are many parameters that effect the rise in temperature due to reaming including the size of the reamer in relation to the size of the intramedullary cavity, the speed of the reamer, the force applied to conduct the reaming procedure and the physical properties of the bone and reamer used [15]. Studies have given recommendations to ensure that thermal damage to the bone is avoided. Since it is known that the size of the reamer in relation to the size of the intramedullary cavity has a strong effect on the heat generation of the procedure, it has been recommended that an assessment of the smallest diameters of the medullary cavity in two planes be conducted prior to reaming [29]. No more than 1 mm of bone should be drilled away in one procedure. The smallest commercially available reamers for power-driven use have a diameter of 9 mm. Should the diameter of the medullary cavity be smaller than 8mm, then the use of AO hand reamers, which are available in smaller diameters, or an alternative fixation procedure, would be recommend [24].

Table 1-1: Severity of Heat Damage [24]

Severity	Effects	Description
Grade 0	No heat induced damage or devascularization	No devascularization or heat damage.
Grade 1	Devascularization	The heat damaged zone is cut away by subsequent reaming procedures. Damage to the vascular system remains.
Grade 2	Heat Damage with conservation of an intact periosteum	The heat damaged areas are not cut away by subsequent procedures. The bone is damaged by heat and devascularized. The damage is so severe that healing will be retarded, but periosteal regeneration potential remains. Potentially delayed fracture healing is the result.
Grade 3	Heat-induced damage to bone, periosteum, and possibly the soft tissues	The entire bone cross section, including the periosteum, is devitalized. In the central area of the diaphysis the heat is such that the damage it causes extends beyond the bone itself. A completely necrotic bone segment forms without any residual local regeneration capacity. If the skin damage promotes infection, severe sequestration will occur. Depending on the extent of damage to the soft tissue, grave consequences are to be expected and further operations will be unavoidable.

Other notable recommendations have been made including avoiding the use of a tourniquet [30], development of improved cutting tools [31], treating reamers gently when using them, as well as replacing them often to ensure the tools are consistently sharp [32], checking the reamer temperature throughout the procedure and allow time for cooling [15].

1.3 Numerical Modeling

Viscous fluid transport in porous media has been investigated by many authors for a variety of applications. Bernsdorf *et al* [33] used a three-dimensional numerical model using the tomography technique in order to study the origins of pressure drop in porous media flow. Their study confirmed that important factors effecting pressure loss in porous media included the elongation and contraction of fluid elements, as well as the absence of shear forces in the model.

Sun *et al* [34] used an explicit finite element method for solving a dual phase, immiscible, incompressible flow in porous medium in two dimensions. A mixed element method was used to approximate the velocity and pressure. The resulting method was shown to converge to an exact solution.

A numerical scheme was created by Holstad [35] in an attempt to show fluid velocity conservation, pressure and temperature fields in a porous medium. The primal dual mixed finite element method was used to calculate the velocity, pressure and temperature. The author showed that the numerical simulation matched experimental results.

A finite element method was presented by Juanes [36] to find a solution of a multiphase flow in porous media based on a multi-scale decomposition of pressures and fluid saturations into resolved scales and unresolved scales. The results compared quite favorably with the classical Galerkin method.

Viscous fluid transport through porous media has a vast range of applications but numerical attempts in the literature to model fluid flow in bones specifically have been limited. There are some notable exceptions including the work of Dunne *et al* [37], whose model describes cement flow in the intramedullary cavity. In their work, a simple cylinder was first modeled and later, the authors progressed to an actual femoral prosthesis. Their theoretical model was designed to allow the prosthesis bone geometry on an individual patient to be evaluated in terms of probable pressure distributions in the medullary cavity during cemented fixation and in turn, can guide the stem design with reference to preparation of the medullary canal. Their predicted pressures from the theoretical model were found to be close to those measured experimentally. The authors were able to obtain the pressure gradient within the bone with a predicted maximum pressure in the range of 70-160 kPa.

Studies were also conducted by Gaber [38] with the modeling of the insertion of a reamer without rotation into the intramedullary canal using a simple co-axial cylinder model. The results obtained were able to closely imitate the behavior of experimental results by Johnson *et al* [39], which used human cadaver femora. Modifications of this model to coincide with the experimental properties used in the research of Dobrjanski *et al* [40] also yielded quality results.

Some studies have been conducted to explain how heat would be produced during reaming of the intramedullary canal during orthopaedic procedures. There have been numerical and analytical studies concerning modeling the thermal effects of drilling and

reaming in bone. Davidson *et al* [41] modeled the effects of drilling on bone by creating an FEM model of two concentric cylinders of a specified height and calculating the thermal penetration of heat from the drill using the thermo-mechanical equations developed from machining theory. Although the results compared only satisfactory with experimental results, the conclusion, through the parametric study that was conducted, was that drill speed and feed rate were significant factors in determining the thermal penetration into the bone.

Another notable analytical model was the work done by Baumgart *et al* [15]. They developed a model that was based on the solution for heat conduction in a semi-infinite rod. Taking into account the heat generation as a result of friction between the reamer and the bone, they were able to construct an analytical model. The model described the heat generated on the reamer face during reaming of the intramedullary canal in orthopaedic procedures. The result was a time dependent equation that would predict the temperature of the reamer face when all the known properties of the bone, reamer and procedure are known.

The present study attempts to model intramedullary reaming and the procedure's subsequent effects on the bone by the use of the finite element method. Initially, the governing equations used to solve this procedure will be presented, along with the solution technique and the mesh sensitivity analysis. Chapter 3 will explain the effects of rotational speed on the intramedullary pressure. Chapter 4 and 5 will examine and compare thermal effects of reaming with experimental data in the literature. Chapter 6

will give predictions on the effects of viscosity to thermal penetration into the bone. Chapter 7 will highlight the conclusion and Chapter 8 will discuss future studies for this research.

Chapter 2 : Governing Equations and Numerical Solutions

2.1 Momentum Equations

The three-dimensional differential equations of fluid flow through a porous medium for a laminar, transient, incompressible, Newtonian flow can be written as follows:

r-Component

$$\frac{\rho}{\phi} \frac{\partial u_r}{\partial t} + \left(\frac{\mu}{\kappa_r} \right) u_r = -\frac{\partial p}{\partial r} + \mu \left[\frac{\partial}{\partial r} \frac{1}{r} \left(\frac{\partial r u_r}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2 u_r}{\partial \theta^2} - \frac{2}{r^2} \frac{\partial u_\theta}{\partial \theta} + \frac{\partial^2 u_r}{\partial z^2} \right] \quad (2-1)$$

θ -Component:

$$\frac{\rho}{\phi} \frac{\partial u_\theta}{\partial t} + \left(\frac{\mu}{\kappa_\theta} \right) u_\theta = -\frac{1}{r} \frac{\partial p}{\partial \theta} + \mu \left[\frac{\partial}{\partial r} \frac{1}{r} \left(\frac{\partial r u_\theta}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2 u_\theta}{\partial \theta^2} + \frac{2}{r^2} \frac{\partial u_r}{\partial \theta} + \frac{\partial^2 u_\theta}{\partial z^2} \right] \quad (2-2)$$

z-Component

$$\frac{\rho}{\phi} \frac{\partial u_z}{\partial t} + \left(\frac{\mu}{\kappa_z} \right) u_z = -\frac{\partial p}{\partial z} + \mu \left[\frac{1}{r} \frac{\partial}{\partial r} \left(\frac{r \partial u_z}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2 u_z}{\partial \theta^2} + \frac{\partial^2 u_z}{\partial z^2} \right] + \rho g \quad (2-3)$$

where r is the bone radius, ρ is the density in kg/m^3 and κ is the bone permeability in m^2 .

Equations (2-1) to (2-3) describe the momentum equation for the fluid flow in a porous medium, where u_r , u_θ and u_z represent the fluid velocity in the radial (r), angular (θ) and axial (z) directions, respectively. Finally, μ represents the viscosity of the fluid within the pores.

2.2 Continuity Equation

In numerical modeling, the continuity equation must be satisfied.

$$\frac{1}{r} \frac{\partial}{\partial r} (ru_r) + \frac{1}{r} \frac{\partial u_\theta}{\partial \theta} + \frac{\partial u_z}{\partial z} = 0 \quad (2-4)$$

Since the flow is incompressible, the term that specifies the change of density with time was eliminated.

2.3 Energy Equation

For the work conducted in Chapters 4-6 the three dimensional energy equation needs to be solved, to complete the solution.

$$(\rho c_p)_{\text{eff}} \left(\frac{\partial T}{\partial t} + u_r \frac{\partial T}{\partial r} + u_\theta \frac{1}{r} \frac{\partial T}{\partial \theta} + u_z \frac{\partial T}{\partial z} \right) = k \left(\frac{1}{r} \frac{\partial}{\partial r} \left(r \frac{\partial T}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2 T}{\partial \theta^2} + \frac{\partial^2 T}{\partial z^2} \right) \quad (2-5)$$

Where ρ and c_p are density kg/m^3 and specific heat at a constant pressure in $\text{J/kg} \cdot ^\circ \text{K}$ of the fluid, k is the thermal conductivity in $\text{J/m} \cdot \text{s} \cdot ^\circ \text{K}$, T is the temperature in $^\circ \text{K}$.

Except in the almost trivial case where only a conduction problem ($u_i = 0$) is being solved, the energy equation is coupled with the continuity and momentum equations through the velocity field u_i , and density, ρ . Because the equations are coupled, it is essential that scaling factors be used to non-dimensionalize them. The energy equation should be consistent in the scaling factors used to non-dimensionalize the continuity and momentum equation.

2.4 Non-Dimensionalization

In order to better understand the results, equations (2-1) to (2-5) are converted to a dimensionless form. This helps in choosing relevant data and assists in presenting the

data in a suitable way. The dimensionless variables used in the momentum, continuity and energy equations are as follows:

$$R = \frac{r}{L}, Z = \frac{z}{L}, U_r = \frac{u_r}{u_0}, U_\theta = \frac{u_\theta}{u_0}, U_z = \frac{u_z}{u_0}, u_0 = \sqrt{g \times L}, P = \frac{pL}{\mu u_0},$$

$$T^* = \frac{T - T_0}{\Delta T}, \tau = \frac{tu_0}{L}, a = \frac{\mu}{\mu}, \quad (2-6)$$

$$Re = \frac{\rho u_0 L}{\mu}, Pr = \frac{\mu c_p}{k}, Pe_T = Re Pr, Da = \frac{\kappa}{L^2}, Fr = \frac{u_0^2}{gL}$$

where L is the characteristic length of the cylindrical bone. $u_0 = \sqrt{gL}$ is the characteristic velocity adopted for non-dimensional calculations, t is the time in seconds, p is pressure, Re is the Reynolds number, Pr is the Prandtl number, Da is the Darcy number, Pe_T is the thermal Peclet number and Fr is the Froude Number. U_r , U_θ and U_z represent the dimensionless fluid velocities in the radial (r), angular (θ) and axial (z) directions, respectively. The non-dimensional variable a is defined as the ratio of the fluid viscosity to the effective viscosity and τ is the dimensionless time. Steps for this non-dimensionalizing process are shown in Appendix A.

2.4.1 Dimensionless Momentum Equations

The three-dimensional differential equations of fluid flow through a porous medium for a laminar, transient, incompressible, Newtonian flow can be written in dimensionless form as follows:

R-Component

$$\frac{\text{Re}}{\varphi} \frac{\partial U_r}{\partial \tau} + \left(\frac{1}{\text{Da}} \right) U_r = -\frac{\partial P}{\partial R} + \frac{1}{a} \left[\frac{\partial}{\partial R} \frac{1}{R} \left(\frac{\partial R U_r}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 U_r}{\partial \theta^2} - \frac{2}{R^2} \frac{\partial U_\theta u_\theta}{\partial \theta} + \frac{\partial^2 U_r}{\partial Z^2} \right] \quad (2-7)$$

θ -Component:

$$\frac{\text{Re}}{\varphi} \frac{\partial U_\theta}{\partial \tau} + \frac{1}{\text{Da}} U_\theta = -\frac{1}{R} \frac{\partial P}{\partial \theta} + \frac{1}{a} \left[\frac{\partial}{\partial R} \left(\frac{\partial (R U_\theta)}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 U_\theta}{\partial \theta^2} + \frac{2}{R^2} \frac{\partial U_r}{\partial \theta} + \frac{\partial^2 U_\theta}{\partial Z^2} \right] \quad (2-8)$$

Z-Component

$$\frac{\text{Re}}{\varphi} \frac{\partial U_z}{\partial \tau} + \frac{1}{\text{Da}} U_z = -\frac{\partial P}{\partial Z} + \frac{1}{a} \left[\frac{1}{R} \frac{\partial}{\partial R} \left(\frac{R \partial U_z}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 U_z}{\partial \theta^2} + \frac{\partial^2 U_z}{\partial Z^2} \right] + \text{Re Fr}^{-1} \quad (2-9)$$

2.4.2 Dimensionless Continuity Equation

$$\frac{1}{R} \frac{\partial}{\partial R} (R U_r) + \frac{1}{R} \frac{\partial U_\theta}{\partial \theta} + \frac{\partial U_z}{\partial Z} = 0 \quad (2-10)$$

2.4.3 Dimensionless Energy Equation

$$\text{Pe}_r \left(\frac{\partial T^*}{\partial \tau} + U_r \frac{\partial T^*}{\partial R} + U_\theta \frac{1}{R} \frac{\partial T^*}{\partial \theta} + U_z \frac{\partial T^*}{\partial Z} \right) = \left(\frac{1}{R} \frac{\partial}{\partial R} \left(R \frac{\partial T^*}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 T^*}{\partial \theta^2} + \frac{\partial^2 T^*}{\partial Z^2} \right) \quad (2-11)$$

2.5 Solution Technique

FIDAP is a general-purpose software package that is used in the simulation of incompressible or compressible flows. FIDAP offers complete mesh flexibility, and has robust calculations. FIDAP's capabilities include fluid flow, heat and free surface

modeling. FIDAP has the capacity to perform all steps of a fluid flow simulation which include model building, mesh generation, data input, solution, and post processing. FIDAP uses the finite element method to solve its problems. When three-dimensional meshes are needed, they are usually exported from the complementary FLUENT software called GAMBIT. GAMBIT is used to generate detailed meshes for complex geometries.

When defining the problem, care was exercised when specifying the allowable errors for the solution convergence. Too large of an error allowance would result in false convergence, and too small of an allowable error would result in either very lengthy calculation times, or no convergence at all. The two main errors that the finite element program took into consideration were the velocity convergence tolerance and the residual vector convergence tolerance. The velocity convergence tolerance is defined by the equation:

$$\left\| \frac{\zeta_i - \zeta_{i-1}}{\zeta_i} \right\| \leq \zeta_{\text{error}} \quad (2-12)$$

where ζ is the solution velocity vector. The error for this study was specified as 1×10^{-3} .

The residual error on the other hand is defined as:

$$\left\| \frac{R_i}{R_0} \right\| \leq R_{\text{error}} \quad (2-13)$$

The error for this case was also specified to be 1×10^{-3} . The norm $\| \|$ is a root mean square norm summed over all the equations for the model.

2.6 Finite Element Method

The non-linear equations (2-1) to (2-5) were solved simultaneously using the segregate solver which creates a separate system of equations for each degree of freedom. An eight node hexahedral element was used for this simulation, where five unknowns, radial velocity, angular velocity, axial velocity, pressure and temperature were evaluated at each element, as shown in Figure 2-1. It was found that a hexahedral element mesh would result in a smoother and faster transient simulation [38].

Figure 2-1 : Hexahedral Elements Mapped in Gambit [38]

In FEM, the partial differential equations of fluid mechanics (the Navier-Stokes equations) that describe the flow throughout its domain are replaced by a set of algebraic equations. This linear system of equations is then solved to determine the velocity components, pressures and temperatures at each node in every element [42]. The time integration scheme that was chosen was Backward Euler, this gave unconditional stability and allowed for large time steps while maintaining convergence.

2.7 Mesh Sensitivity

2.7.1 Mesh Sensitivity Pressure Model

To ensure the correctness of the model, mesh sensitivity was considered to reduce the number of elements and nodes in the mesh while maintaining accurate results. This is usually done to minimize computational time and memory, while keeping the model accurate.

Gaber *et al* [38] provided a grid refinement study for the number of axial nodes in the model. However, due to the present additional boundary conditions and their direct effect on the nodes on the inner cylinder edge, refined mesh sensitivity has been performed also for these radial nodes. A non-uniform boundary layer grid structure starting at the edge of the inner cylinder has been applied. A large aspect ratio for the node location starting from the edge of the inner cylinder has been initially chosen. This aspect ratio has been gradually decreased until the optimum mesh was found. Due to the transient nature of the experiment, the average pressure taken at 3 different times 0.68 s, 3.39 s and 6.80 s (which represents the

initial, middle and final time steps) has been considered as a key parameter for the refinement study.

As the aspect ratio has been decreased, the error value has been computed. When the percentage error of the average pressure values between successively finer meshes was less than 5%, the second last mesh was deemed adequate. In the present case using an initial distance of 5% of the inner radius the optimum aspect ratio for the radial mesh is 1.4.

Figure 2-2: Percentage Error for the Non-Dimensional Pressure Values at the mid-section vs. the Aspect Ratio

2.7.2 Mesh Sensitivity Thermal Models

The mesh sensitivity for the thermal studies done in Chapters 4-6 was conducted to insure optimal computing time without the accuracy of the results being compromised.

For the minipig model, a model was initially selected with the maximum number of nodes possible for the model within the allowable memory confines. This was used as the base to which other models were tested against. It consisted of 100 axial nodes, 10 angular nodes and 50 radial nodes. Initially, the angular and radial nodes on the 'control' model were kept consistent and the radial nodes were varied. Since a transient model was developed, the non-dimensional temperature within the bone was taken at four different times, 6 seconds, 12 seconds, 18 seconds and 24 seconds, which represent a full range of times over the course of the reaming. As the node number was increased, the non-dimensional temperature in the middle of the long bone, 1 mm from the reamer radius, was calculated and compared to the control mesh. This comparison revealed a percentage error which is displayed in Figure 2-2. When the percentage error of the non-dimensional temperature was less than 5%, the mesh was deemed accurate. The number of axial nodes is therefore 50. The optimal angular and radial nodes were calculated in a similar fashion and the optimal nodes for them were 6 and 40, respectively.

Figure 2-3: Percentage Error for the Non-Dimensional Temperature Values 1 mm from the Reamer Face vs. the Number of Axial Nodes

Mesh sensitivity for the other models used in chapter 4 and chapter 6 were conducted in a similar fashion.

2.8 Boundary Conditions

Each chapter will describe a different problem and will incorporate a different model.

The boundary conditions for each model will be summarized in the associated chapter.

Chapter 3 : Effect of Rotation on Pressure Build-up with a Clogged Reamer

3.1 Introduction

A numerical model was created to discover the effect of rotation rate on the pressure build-up in the intramedullary cavity. The model assumes that the flutes of the reamer are clogged with marrow and bone debris and the reamer is acting as a spinning hydraulic piston. The results of the experiments conducted by Johnson *et al* [39] and the numerical study done by Gaber [38] are compared to the numerical results obtained in this study. The need to accurately compare to both Johnson *et al* [39] and Gaber [38] was essential, so the same concentric cylinder setup was used, as in Gaber [38]. This configuration is seen in Figure 3-1. Figure 3-1 also shows the location of the compact bone layer, as well as the intramedullary section in the femur.

The inner diameter of the cylinder is used to mimic the intramedullary section of the bone and was adjusted for each reamer size to make a tight sealed fit. The rotation components were added to the bottom surface of the inner cylinder to simulate the rotation of the reamer during insertion. The cadaveric femur bone used in Johnson *et al* [39] was wrapped in Teflon so that no fluid was allowed to escape through the pores of the bone, a condition reflected in the choice of the zero velocity boundary condition on the outer surface of the cylinder. Two cases were solved, to compare to both Johnson *et al* [39] and Gaber [38]. The first case was to examine the effect of an insertion of a 9 mm reamer, and the second case an insertion of a 10 mm reamer.

3.2 Model Description

Following the work of Gaber [38], the femur was modeled as two co-axial cylinders with a length of 350 mm and an outer diameter of 32 mm. This is displayed in Figure 3-1. The inner diameter of the cylinders was varied from 9 mm to 10 mm, to reflect two differently sized reamers. The bottom surface of the inner cylinder was defined so that it was able to simulate the rotation and insertion of the reamer into the bone marrow cavity.

Figure 3-1: Numerical Model of the Femur with Femur profile [43]

An important parameter when discussing the causes of FES is bone porosity. The porosity of bone can vary with age and gender and an approximation of this important parameter must be obtained for the simulation. It has been shown [44] that use of a linear isotropic description of cortical bone as a two-level porous medium is an appropriate

approximation. The average bone porosity ϕ varies between studies [44][45] but can be approximated to be 5%. The permeability κ of bone, like the porosity, is also seen to be quite varied [45]; an approximate value that was seen in [46] and used by Gaber [38] and will be utilized is in the range of $3 \times 10^{-9} - 16 \times 10^{-9} \text{m}^2$. By using the theoretical relationships described in depth in [44], poroelastic parameters can also be described for other bone types by using their experimental data sets.

Another important input parameter in numerical modeling of reaming is the viscosity of the bone marrow. It is known that bovine marrow characteristics greatly resemble those of humans [47] and is commonly used in biomechanical and numerical studies. Known values of bovine marrow have been experimentally calculated to be in the range of 0.04-0.4 Pa·s [47]. It is also known that bone and bone marrow both have similar densities. These values are approximately $1810 \text{ kg/m}^3 - 2100 \text{ kg/m}^3$ [43][48]. The material properties used in this study are summarized in Table 3-1. It should be noted that, in the current model, the simulation is assumed to be isothermal. This allows the energy equation to be omitted from the calculations, and pressure will be the dominant force driving the simulation.

Table 3-1 : Physical Properties of the Bone and Marrow

Physical properties of Compact Bone section		Physical properties of Medullary section	
Symbol	Value	Symbol	Value
ϕ	0.05	ϕ	0.95
κ	$3 \times 10^{-9} - 16 \times 10^{-9} \text{m}^2$	κ	$3 \times 10^{-7} - 16 \times 10^{-7} \text{m}^2$
ρ	1810 kg/m^3	ρ	1810 kg/m^3

3.3 Boundary Conditions

Predictions of the present model will be compared to the results of Johnson *et al* [39], who monitored the pressure distribution within a femur during intramedullary reaming, and Gaber [38], who used a numerical model to simulate the experiment without rotation of the reamer. In the experiments of Johnson *et al* [39] a frozen cadaveric femur was cleaned of all flesh and instrumented with pressure taps mounted at mid-shaft. Prior to power reaming, there was initial canal broaching with an awl and guide wire placement. Power reaming then began with a 9 mm reamer which produced the initial cavity. The reaming speed was 2.58 cm/s (± 0.6 cm/s). Since there was no living tissue surrounding the bone, the cavity stayed empty after the initial ream. In reality, the cavity would fill up with blood and fat instantaneously. This was simulated by refilling the bone cavity with albumin. The second reaming was done using a 10 mm reamer. Pressure values at the mid-point were obtained for both reaming cases and were plotted against the experimental results.

The boundary conditions used in the model were chosen to simulate experiments performed by Johnson *et al* [39] and the simulation performed by Gaber [38]. The bottom face of the inner cylinder will act as a moving boundary to mimic the insertion of the reamer into the femur bone. In Johnson *et al* [39], Teflon was wrapped around the bone to prevent the escape of intramedullary canal contents and to maintain a constant temperature of 37°C (thus maintaining the assumption that the process was isothermal and that the energy equation could be omitted for the sake of simplicity). The boundary conditions adopted in this model were thus:

- Velocity was specified as zero at the surface of the outer cylinder, as well as the top and bottom of the outer cylinder and the top of the inner cylinder. Therefore, no fluid could escape through the side walls or the top and bottom of both cylinders, The velocity boundary conditions at $r = 1.6$ cm and at $z = 0$ and 35 cm therefore are :

$$U_r = 0, \quad U_\theta = 0, \quad U_z = 0 \quad (3-1)$$

- At the bottom of the inner cylinder, the surface velocity is set to be equal to the insertion rate to mimic the movement of the reamer as it advances at a constant velocity.

$$U_z = \text{insertion velocity} \quad (3-2)$$

- The bottom face of the inner cylinder is also given the boundary conditions to imitate the rotation of the reamer.

$$U_\theta = \text{rotation speed} \quad (3-3)$$

Figure 3-2: Boundary Condition Used in the Femur Model

3.4 Results and Discussion

3.4.1 Case 1: Insertion with 9 mm Reamer

The first reamer size that Johnson *et al* [39] used was the 9 mm reamer. Pressure values were recorded as the maximum value seen at the mid-shaft as the reamer was inserted into the bone. Gaber [38] had investigated the effect of insertion without rotation and his results are depicted in the Figure 3-2 and Figure 3-6 as 0 rpm value. For this simulation rotational speeds ranging from 1 rpm to 450 rpm were chosen. These values were used to simulate actual rotational speeds present during orthopaedic procedures. Mousavi *et al* [49][50] investigated intramedullary pressures in sheep during reaming and

showed rotational speeds of 150 rpm and 450 rpm. In the present simulation, 450 rpm was taken as the high rotational speed and 100 rpm the low speed. 1 rpm was used simply to ensure the model was operating properly. With the 9 mm reaming case two scenarios were analyzed, the case of varying viscosity and the case of varied insertion rate.

3.4.1.1 Varying Insertion Rate

In Figure 3-2(a) and 3-2(b) the variation of pressures over the reamer location is presented for an insertion speed of 2.58 cm/s. In this figure, the viscosity of the marrow was assumed to be equal to $\mu = 0.22 \text{ Pa}\cdot\text{s}$. What is evident in this simulation, as well as in Gaber [38], is, when the reamer advances, the pressure increases continuously in an exponential fashion and compares well with the experimental results.

Figure 3-3: Pressure Variation during Insertion of a 9mm Reamer at a Constant Velocity of 2.58 cm/s

Top (a) Full View

Bottom (b) Magnified View of Center

Looking at Figure 3-2(b), one can see more clearly the small impact the rotational component has on the overall pressure distribution. This shows that, once the flutes are clogged, the speed of the rotation is insignificant to the pressure increase because no bone and marrow debris is being transported out of the bone. Figure 3-3 displays the exact pressure increases due to the added energy that the rotational component supplies. Clearly, the rotational component has little effect as all pressure increases were less than 1 %. Johnston *et al* [39] gives the mean rate of insertion to be $2.58 (\pm 0.6)$ cm/s. Trying to determine if rotation would have any effect at the minimum and maximum extremes of the insertion, simulations were run for 1.98 cm/s and 3.18 cm/s. The base line comparison is

made to Gaber [38] whose results are the control 0 rpm. Although Figure 3-4 shows larger percentage increases for the slower insertion rates, all values are in the range of less than 1% and can be deemed insignificant to the overall pressure increase.

Figure 3-4: Percentage Increase in Pressure due to the Added Rotation with 9 mm Reamer with 2.58 cm/s Insertion

Figure 3-5: Percentage Increase in Pressure of 100 rpm due to Varying Insertion rates

3.4.1.2 Varying Viscosity

The present model made the same assumptions as Gaber [38] in that back flow fluid was not allowed to escape from the interface between the bone and reamer, which is an idealized condition. Nevertheless, it should be noted that Johnston *et al* [39] did not report any fluid escaping. To follow the results of Gaber [38], the simulation was repeated with the initial insertion rate of 2.58 cm/s and varying viscosities to examine the effect of rotation with varying viscosities. The results of Gaber [38] showed significant changes in intramedullary canal pressure at the mid-shaft during insertion, and he found that the marrow viscosity played an important role in determining the pressure variation as a function of the insertion rate. The effect of the rotation that was added was minimal, regardless of the viscosity value, as is shown in Figure 3-5, which displays the percentage increase of intramedullary canal

pressure at the mid-shaft of 100 rpm over 0 rpm at varying viscosities. The added rotation is not fulfilling its purpose of extracting the debris at a faster rate because the flutes are clogged, and with the rotational speeds used in the reaming procedure, an insignificant amount of energy is added to elevate the intramedullary pressure.

Figure 3-6: Percentage Pressure Increase of 100 rpm due to Varying Viscosity

3.4.2 Case 2: Insertion with 10 mm Reamer

In the experiments of Johnson *et al* [39], the same bone that was used in the initial reaming using the 9 mm reamer was subsequently used for the reaming using the 10 mm reamer. To simulate the refilling of the reamed portion of the intramedullary canal with blood and marrow the bone was filled with albumin before reaming the bone again using the 10mm reamer. In Johnson *et al* [39], because there is no mention of the specific type

of albumin used, there was an uncertainty in the viscosity values to be used in the simulation. Gaber [38] used varying viscosity values of albumin between $0.0077 \text{ Pa} \cdot \text{s}$ and $0.077 \text{ Pa} \cdot \text{s}$. This analysis will also incorporate these values. There would likely have been some human marrow left from the previous reaming due to the fact that the bone was not cleaned prior to the albumin being inserted, a consistent viscosity will be assumed for the specified albumin. Using the same assumption as Gaber [38], the present works assumes that the fluid wetting the compact bone did not change viscosity between the two cases.

3.4.2.1 Alternative Reamer Diameter

Using the new viscosity values, the simulation was repeated using the 10 mm reamer with an insertion rate of 2.58 cm/s , with varying rotation rates. This was to determine if the rotational speed had an impact on different diameter reamers. As was the case for the 9 mm reamer, it can be seen for the 10 mm reamer the minimal effect that the rotation of the reamer has on the overall effect of the intramedullary pressure. Figure 3-6(a) and 3-6(b) display the intramedullary canal pressure at mid-shaft with viscosity equal to $\mu=0.0385 \text{ Pa} \cdot \text{s}$.

Figure 3-7: Pressure Variation during Insertion of a 10 mm Reamer at a Constant Velocity of 2.58 cm/s
Top (a) Full View   **Bottom (b) Close View of Center**

Figure 3-8: Percentage Increase in Intramedullary Pressure at the Mid-Shaft of 100 rpm during insertion of 10mm Reamer with Varying Viscosities

3.4.2.2 Varying Viscosity

The viscosity of albumin over a range of values was used to see the effect of changing viscosities in different reamer diameters. The viscosity was varied from 0.0077 Pa·s and 0.077 Pa·s to see if rotation rate would produce an effect on the intramedullary pressure. What is evident after the simulations are analyzed is the rotation rate has little effect on the overall intramedullary pressure build-up. Figure 3-7 shows this insignificant effect very clearly in that the percentage that the pressure increases due to the increased rotation is less than 1%.

3.5 Conclusion

When reaming of the long bones is conducted, it is known that the flutes of the reamer become clogged with debris and the reaming procedure becomes similar to that of a hydraulic piston, a cylinder filled with viscous fluid [16]. From the present analysis it can be seen that when this occurs, the pressure increase in the intramedullary cavity is unaffected by the rotation rate of the reamer. This has been shown to be true regardless of the insertion speed or viscosity of the marrow. Based on the present findings the surgeon need not be concerned with the reamer rotation rate if the flutes of the reamer have become clogged. This will have a minimal effect on intramedullary pressure increase.

It should be noted that although the rotation rate of the reamer has been found to have no significant effect on the pressure increase in the intramedullary cavity, it is possible that the rotation rate affects the temperature increase of the bone tissue during reaming. This temperature change due to reaming is a topic that requires further investigation, as it has been shown that if the reaming temperature exceeds 44.1°C, cell enzymes are damaged and the bone healing process is negatively influenced [15].

Chapter 4 : Verification of the Temperature Distribution in the Porous Model to Cortical Bone during Reaming

4.1 Introduction

The study on the heat effects on cortical bone conducted by Frölke *et al* [48] involved using an artificial bone substitute rather than a cadaveric bone. The implementation of a reamer, and the associated heat generation into the bone, was orchestrated by the use of a controlled heat source instead of an actual reamer. The heat source was kept at a constant temperature of 47°C through the use of an electrical heat inductor to insure a reliable result. The setup used in his experimentation can be seen in Figure 4-1. The artificial bone that was employed to act as cortical bone in the study was polycarbonate [48]. The measurements were carried out on artificial bone to rule out the confounding effects of the heterogenic structure of cortical bone [48]. The properties that were required to conduct an accurate simulation model of the experiment were readily available and are displayed in Table 4-1. The comparative properties of both polycarbonate and human cortical bone are seen to be relatively close and it is an accepted material to be used in the modeling of human bone [51]. The properties of polycarbonate will be implemented into the porous model to confirm its accuracy.

Because the artificial reamer that was applied in the study was not moving and fitted exactly in the polycarbonate cylinder, it was able to be removed and simulated in the model by the use of a constant heat boundary condition. This allowed the model to consist simply of the outer hollow cylinder. Symmetry was again utilized to ensure that the maximum accuracy was attained with the minimum amount of computing time. The model used in the simulation is displayed in Figure 4-2.

Frölke *et al* [48] also used an approved mathematical model in their study, the results of which will be used in tandem with the experimental to confirm the accuracy of the current porous model. Once the porous model is able to be verified, it will allow for its use in more complex simulations which more accurately represent the actual reaming during orthopedic procedures.

Table 4-1: Material Properties of Bone and Bone Substitute [48]

Material	Density [kg/m ³]	Conductivity Coefficient J/m·s·°K	Specific Heat J/kg·°K
Bone	2100	0.4	1285
Polycarbonite	1200	0.2	1170

4.2 Model Description

4.2.1 Porous Model

The porous model that was used was created to ensure that the porous media specifications that were being applied in the model would accurately describe the temperature distribution that would occur in cortical bone due to the traumatic effects of a heated reamer. The porous model was a 10 cm long hollow tube with an outer diameter of 24 mm and an inner diameter of 12 mm. This porous model would be validated against the results obtained by Frölke *et al* [48]. The properties of the artificial bone substitute used in [48] were implemented in the porous model.

Figure 4-1: Experimental Model [48]

Figure 4-2: Porous Model

4.2.2 Boundary and Initial Conditions of the Porous Model

The predictions of the present model will be compared to the results of Frölke *et al* [48], who monitored the temperature distribution through cortical bone when one side is heated. Frölke *et al* [48] used both an experimental and an approved numerical model to display his findings. In his study a controlled heat source of 47°C was used instead of a reamer. The heat source fitted exactly into the polycarbonate cylinder with no gap between the two.

The boundary conditions used in the model were chosen to simulate the experiment of Frölke *et al* [48] mentioned above. The experimental models were placed in a 37°C water bath and the boundary conditions were chosen to be consistent with this:

- Velocity was specified as zero at all of the boundary locations throughout the model as there was no movement of the reamer.

$$U_r = 0, \quad U_\theta = 0, \quad U_z = 0 \text{ at boundaries} \quad (4-1)$$

- The inner wall of the cylinder was set at a constant value of 47°C

$$T_r = 47^\circ C \quad \text{at} \quad r = 6\text{mm} \quad (4-2)$$

- The outer wall of the cylinder was set at a constant value of 37°C

$$T_r = 37^\circ C \quad \text{at} \quad r = 12\text{mm} \quad (4-3)$$

4.3 Results

The temperature distribution that was observed in the model closely resembled both the numerical and experimental studies. After 15 seconds with a source temperature of 47°C, the porous model temperature was 42.8°C. This compares well with the measured temperature of Frölke *et al* [48] experiment of 43.1°C (± 0.25)°C and the measured values from the numerical analysis of Frölke *et al* [48] which was 43°C. The thermocouples that were used in the experiment had a measurement accuracy of approximately 0.25°C, this is accounted for in the error bars seen in Figure 4-3. Contour Plots at varying steps in the simulation are presented in Figure 4-4.

Figure 4-3: Comparison of Temperature Distribution of the Porous Model to the Experimental Results

Figure 4-4: Temperature Contour Plots of Cross-Section and Different Times.

4.4 Discussion

Temperature measurement has been focused on the cortical heat development during intramedullary nailing in both cadaveric and experimental models since the technique was first implemented [31]. It is a standard procedure to ream the medullary canal at least one millimeter more than the nail diameter to allow the nail to be inserted without risk of additional iatrogenic fracture [48]. If the intramedullary canal is thin, then this can only be completed with multiple passes of the reamer in small step-up increments which can

add to temperatures in excess of the allowed limits [18]. Having a numerical model that can predict the temperature gradients in the cortical bone due to the temperature increases of the reamer will be of great assistance to both the surgeon and the patient.

The experimental and numerical studies that were conducted by Frölke *et al* [48] used an artificial bone substitute to replicate the temperature distribution in the cortical bone during heating, a phenomenon that is seen during the reaming of the intramedullary cavity during both hip and knee replacement surgery. Comparison to both the experimental and numerical procedures would allow the porous media model for cortical bone to be verified. Having an accurate numerical model can allow the patient to avoid the complications that arise during the elevated temperatures in the cortical bone. This provides faster patient recovery and shorter hospital stays. As is displayed in Figure 4-3, one can see that the temperature profile of the bone at 1 mm away from the heat source in the porous media model compared very well to the published experimental results. The clinical implications are limited for this specific study because the elevated temperatures are dependent on numerous parameters [15], including the widths of the intramedullary canal, which can be extremely narrow causing significant heating effects [52]. The reamer design and sharpness also play an important role [32].

Although the temperature of the reamer is far from constant over the course of the procedure, this model is a first verification of temperature distribution in the bone. Using an artificial bone allows there to be a certainty of consistency of properties throughout the

bone. This ensures confidence that the properties inputted into the porous model coincide correctly with those in the experimental procedure.

The effect of elevated temperatures during intramedullary reaming procedures can produce post operative complications. This can cause the patient receiving the operation to have a lengthy recovery time resulting in longer hospital stays. This study was performed to validate the porous model's ability to replicate the effects in cortical bone due to the heat generated by the reamer during orthopaedic surgery. Knowing that the porous model's properties are consistent with that of the artificial bone used, and seeing how closely the porous model mimics the experimental and numerical results of Frölke *et al* [48], it can be ensured with a high level of certainty that the porous media model is in fact a good replication of human cortical bone. This porous model can now be used in a more sophisticated model that will replicate the actual reaming of the intramedullary canal.

Chapter 5 : Verification of Minipig Model to the Experimental Results Obtained *in vivo*

5.1 Introduction

A numerical model was created to compare predictions for thermal penetration into the cortical bone with experimental data. This study used the porous bone model that was developed and verified by comparison with experimental results of Frölke *et al* [48], and the reamer temperature equation that was developed by Baumgart *et al* [15] to construct a model that describes the heat effects on the cortical bone during reaming of the intramedullary canal in orthopaedic procedures. Verification of this model to experimental results is to be conducted to ensure accurate future predictions.

The numerical results were compared to an experimental *in vivo* procedure conducted on minipigs in Garcia *et al* [53]. Many of the required parameters were provided in the paper, unfortunately some vital components were not reported. The physical properties of swine bone are not given, nor are the length of the minipig tibia, the age of the minipigs nor the axial force used. It was assumed that the minipigs were approximately one year old as is consistently seen in other minipig experiments requiring analysis of bone [54][55][56]. The length of the average minipig tibia was determined to be 12 cm, and the outer diameter was determined to be approximately 14.5 mm. This value was determined from data obtained from [55]. See Appendix B for a more comprehensive analysis.

Animal models are commonly used in medicine for *in vivo* testing of techniques and procedures in orthopaedics. They offer excellent comparisons to human bone and are used to improve techniques and examine shortcomings that can be involved with the operational procedure. Although swine, including minipigs, have a higher bone mass than humans, they are large enough to receive prosthetic implants, and bone removal and decomposition occur at the same rate as in humans [57]. These attributes make them widely used for *in vivo* testing. Aerssens *et al* [58] stated that of the six domestic animals that were examined in their study, swine and dog bone most resemble the human bone. For this reason, the physical property values that were used in the model were those of human bone and can be assumed to compare reasonably well with the bone of swine.

It is known that reaming procedures are conducted using axial forces of varying values between 20 and 60N, as can be seen in [39],[59]. Three values of 20N, 40N and 60N were implemented to see if this range of axial force had a profound effect on the temperature distribution. Although values varied, the variations were slight and all temperatures fell within the margin of error. The results are displayed in Appendix C. Therefore an axial force value of 40 N was chosen to act as an average value throughout the rest of the study, due to the fact that the reaming procedure most certainly had a variable force.

It was not specified whether the procedure was conducted with the tibia in the horizontal or vertical position. Therefore, the effect of a directional gravitational force was investigated to determine any variations in the temperature increase.

5.2 Model Description

5.2.1 Minipig Model

For simplicity, the minipig model consists of three 12 cm long, co-axial cylinders with an outer diameter of 14.5 mm. The inner diameter is 6.5 and 8 mm, and the middle cylinder is the reamer diameter of 8 mm and 9 mm respectively. The innermost cylinder was describing the marrow while the middle cylinder is the bone that the reamer is removing. The bottom of the inner and middle cylinders surface is defined in such a way that it moved with time to simulate the reamer advancing through the bone. This model was created to mimic the procedure performed in Garcia *et al* [53] which recorded temperature variations *in vivo* during the intramedullary reaming of the tibia in 14 minipigs. Symmetry was utilized to ensure that the maximum accuracy was attained with the minimum amount of computing time. The model mesh used in the calculations appears in Figure 5-1.

Figure 5-1: Minipig Model

The temperature profile that was used in the minipig model to imitate the reaming procedure was based on the work of Baumgart *et al* [15]. He derived an equation for temperature based on the theory of heat conduction in a three-rod cylinder. The parameters and the values that are required for the model to accurately describe the reaming procedure following the equation in [15] are: the axial force of reaming procedure, the rotational speed of the reamer, the coefficient of friction, the physical properties of the bone and reamer, the inner and outer diameter of the bone and the reamer diameter. The full list of properties and the values that were implemented are listed below in Table 5-1.

Table 5-1: Parameters used in the Minipig Model

Parameter	Symbol	Unit	Value
Axial Force	F	N	40
Rotational Speed	n	rpm	800 [53]
Coefficient of Friction	μ_f		0.1[15]
Bone			
Density	ρ	kg/m ³	2100[48]
Specific Heat	c_p	J/kg·° K	1285[60][48]
Thermal Conductivity Coefficient	λ	J/m·s·° K	0.40[48]
Reamer Material (Steel)			
Density	ρ	kg/m ³	7850[15]
Specific Heat	c_p	J/kg·° K	510[15]
Thermal Conductivity Coefficient	λ	J/m·s·° K	14[15]
Bone Measurements			
Outer Diameter	d_a	mm	14.5[55]
Inner Diameter	d_i	mm	6.5,8
Reamer Measurements			
Diameter of the Head	d	mm	8,9

5.2.2 Initial and Boundary Conditions of the Minipig Model

The predictions of the present model were compared to the results of Garcia *et al* [53], who recorded the temperature within the tibia during intramedullary reaming. In the experiments of Garcia *et al* [53] minipig tibia were reamed initially with an 8 mm reamer then with a 9 mm reamer. The average time of each reaming was 31.7 ± 4.3 seconds. The time used in the present simulation was 30 seconds. The temperature was recorded with a thermal probe placed 1 mm away from the endosteum, perpendicular to the long axis of the medullary canal and in the middle third diaphysis, where the canal is narrowest and therefore where more reaming takes place [53]. The location of the thermocouple can be seen in Figure 5-2. The procedures were conducted *in vivo* leaving most of the bone

covered by tissue allowing the use of constant temperature boundary condition on the outside of the bone [61].

Figure 5-2: The Location of the Temperature Probe [53]

The boundary conditions used in the model were chosen to simulate the study of Garcia *et al* [53] mentioned above.

- Velocity was specified as zero at the surface of the outer cylinder, as well as the top and bottom of the outer cylinder and the top of the inner cylinder. Therefore, no fluid could escape through the side walls or the top and bottom of both cylinders. The velocity boundary conditions at $r = 7.25$ mm and at $z = 0$ and 12 cm therefore are :

$$U_r = 0, \quad U_\theta = 0, \quad U_z = 0 \quad (5.1)$$

- At the bottom face of the innermost and middle cylinder, the surface velocity is set to be equal to the insertion rate of the reamer to mimic the movement of the reamer as it advances at a constant velocity.

$$U_z = \text{insertion velocity} \quad (5.2)$$

- The initial temperature of the entire model for the 8 mm reamer was set as the average initial temperature given in [53].

$$T_i = 34.7^\circ\text{C} \quad (5.3)$$

- The initial temperature for the 9 mm reaming was slightly different. To account for the 8mm reamer insertion. The initial value of bone and marrow was given the final temperature of the 8mm pass.

$$T_{\text{9mm initial}} = 36.2^\circ\text{C} \quad (5.4)$$

- The temperature of the bottom face of the innermost and middle cylinder was set to change with time as specified in the function [15].

$$T_{\text{Bottom Face Inner Cylinder}} = \Phi \frac{16}{3\sqrt{\pi}} \frac{\mu_f F_n}{d_i \sqrt{\rho_1 c_{p1} \lambda_1}} \sqrt{t} \quad (5.5)$$

where

$$\Phi = \frac{1 + \frac{d}{d_i} + \left(\frac{d}{d_i}\right)^2}{\left(1 + \frac{d}{d_i}\right) \left[2 \left(\frac{d_a}{d_i}\right)^2 - \left(\frac{d}{d_i}\right)^2 - 1 + \left(\frac{d}{d_i}\right) \sqrt{\frac{\rho_3 c_{p3} \lambda_3}{\rho_1 c_{p1} \lambda_1}} \right]} \quad (5.6)$$

- The wall of the outer cylinder was set to mimic the actions of the tissue surrounding the bone which is circulating with blood and is set to the constant of the average body temperature.

$$T_{\text{Outer Cylinder Wall}} = 34.7^\circ\text{C} \quad (5.7)$$

5.3 Results

5.3.1 Effect of Reaming

All pertinent information was given or was able to be collected from data associated with the specified animals. This allows for the use of the temperature equation that Baumgart *et al* [15] derived. Two passes were conducted, one with the 8 mm reamer and one with the 9 mm reamer.

Figure 5-3: Temperature at Midsection 5.5mm from the Center

Figure 5-4: Contour Plot of Temperature Penetration into the Bone with the 8mm Insertion

Garcia *et al* [53] conducted *in vivo* the intramedullary reaming of tibia in 14 minipig specimens and found that there was temperature increases associated with this procedure. Peak temperatures found ranged from 34.9°C to 49.4°C. The average maximum bone temperature when reaming was conducted with 8 mm and 9 mm reamers and were found to range from 41.1°C $\pm$ 4.4°C. The mean average increase in the temperature in the second group of animals (8.2°C $\pm$ 4.3°C) was greater than the first group (5.3°C $\pm$ 3.5°C). This would be due to the extended use of the reamers which in turn causes some blunting of the edges.

Temperature measurements were taken from the center of the bone length-wise and 5.5 mm from radial centre of the bone or 1 mm away from the 9 mm reamer, to coincide with the location description presented in [53]. The recorded temperatures maximum temperature increase was 7.65°C. As is observed in Figure 5-5 it can be seen that the porous model's maximum temperatures fall within the margin of error of both the blunt and sharp reamers, with the maximum temperature much closer to the average of the blunt reamer.

Figure 5-5: Temperature Increase Comparison

5.4 Discussion

Comparing the recorded maximum temperature from the minipig model it can be seen that the results compare favorably to the experimental values obtained by Garcia *et al* [53], within the margin of error.

The effects of heat on tissue depend on the temperature achieved and the amount of time that the tissue is exposed. Immediate cellular necrosis attributed to thermal energy has been reported with temperatures above 70°C [62]. This is the extreme case, but studies

have found that irreversible protoplasmic changes occur with moderate temperature elevations [52][62]. Heating to temperatures of 60°C in rabbits found a permanent cessation of blood flow and obvious bone tissue necrosis. This damage showed no signs of repair over the follow-up period of 100 days or more [26]. Exposure to a temperature of 47°C for one minute was seen as the threshold level for bone survival [26]. Frölke *et al* [48] assumed that the time of inserting and removing a reamer from the medullary canal should not exceed 40 seconds under normal conditions. The time of true contact between the reamer and the inner cortical wall may be even less than 15 seconds, implying that it may take a temperature as high as 77°C to cause irreversible damage [48]. This may be true for one reaming pass but as is seen in [18][52] the total reaming time to obtain a large enough hole to facilitate the insertion of the implant requires multiple passes. This increases the required time to complete the procedure from 40 seconds to approximately 400-750 seconds and can include upwards of 12 passes, depending on the incremental increases used and the initial and final cavity size required.

Muller *et al* [18] reamed five pairs of cadaveric femora and found an increase in the temperature of cortical bone of 44.1°C. This study was done *in vitro* and would produce different results than *in vivo* testing due to the specific heat of *in vitro* bone being lower than that of *in vivo* bone due to the lower water content [62]. Henry *et al* [31] conducted reaming on 12 cadaveric tibia and 5 cadaveric femora also found temperature increases ranging 42°C to 67°C. From these results it was concluded that there is a direct correlation between the reamer size and the temperature increase, but factors other than reamer size were not considered. Peyton [63] found evidence that temperature increase

was directly related to the mass of bone removed by rotating instruments. A wide range of temperature increases are found between studies, some of which reach the threshold of thermal necrosis, and some fall below. This impedes the creation of an accurate procedure that will allow decreased reaming time and still maintain a margin of safety to insure that thermal necrosis is not an issue during the reaming process.

Baumgart *et al* [15], in a theoretical model of intramedullary reaming, established that it is possible to describe the heat flow that exists between the bone and the reamer in terms of a modified solution to heat conduction in a semi-infinite rod. The equation and parameters needed are shown in Equation 5-5 and Table 5-1 respectively. For a detailed account of the derivation of Equation 5-5 consult [15]. In the reaming procedure conducted by Garcia *et al* [53] the parameters needed for the theoretical thermal conductivity model were given or able to be approximated within a high level of certainty to allow for a verification of the numerical model using experimental results.

This is the first study to be done to incorporate the reamer head temperature profile from [15] into a numerical model that describes the reamer's progress through the bone while accurately tracking the heat flow in the bone at all locations. What is evident by examining Figure 5-4 is that the minipig model is an accurate representation of the temperature increases that are present during reaming. The model followed two reaming passes through the bone with an 8 mm and 9 mm reamer. The maximum temperature was reached using the 9 mm reamer, and the temperature value falls within the error range of both the sharp and blunt reamer data that was found in [53]. The minipig model

shows that temperature profiles in the bone during reaming can be accurately modeled numerically. The maximum temperature in this study did not reach a level where thermal necrosis became an issue. The verification of the minipig model to accurately approximate the temperature increases associated with reaming when all the required parameters are known allows for modification to the operational technique by the surgeon. By increasing rotational speed, axial force, or another parameter, it is possible to decrease the time required to complete a successful operation. The minipig model can be used to predict the possible extremes that the procedure can be conducted in. This leads to maximizing the number of patients that can be operated on by the surgeon in a given time frame, while minimizing the surgical trauma associated with such invasive procedures.

Chapter 6 : Prediction of Viscosity Effects on Temperature in Experimental Reaming Procedures

6.1 Introduction

Experimental studies focused on correcting the adverse effects displayed in patients following orthopaedic procedures are revealing compelling results and further work is currently underway [40][64]. Studies requiring a bone marrow substitute have in the past used a mixture of paraffin/Vaseline [40][65]. Dobrjanski *et al* [40] used a paraffin/Vaseline mixture of 45/55 at 20°C to produce a close approximation to actual bone marrow viscosity. That work yielded beneficial results concerning pressure buildup in the intramedullary canal due to implant insertion. Further study on these experimental setups is currently being conducted to investigate the effects of reaming on pressure buildup in the intramedullary canal and the associated thermal effects. It is known that paraffin viscosity is a function of temperature and decreases over elevated temperatures [66]. This could cause undesirable effects in experiments using a paraffin mixture that conduct research on temperature distribution during reaming. The purpose of this study is to judge the effects of viscosity on the temperature distribution during experimentation.

Current work is being conducted that expands on the studies of [64]. These studies use a plastic outer shell as a base and a foam middle ring to simulate the bone removed due to reaming. The bone marrow substitute that is used is the paraffin/Vaseline mixture of 45/55. The viscosity of the mixture compares well to human bone marrow at 20°C but the temperature of the reamer face will increase due to friction between the reamer and the bone substitute. This heat will flow to the marrow substitute increasing the

temperature and subsequently affect the viscosity. To find the exact viscosity of the mixture at any given temperatures, the mixture was heated to 60°C to remove any air bubbles, and then cooled to varying temperatures. The mixture's viscosity was measured using a viscometer. The results were calculated by Daniel Dobrjanski at Ryerson University. They are seen in Figure 6-1.

Figure 6-1: Viscosity Profile of the Paraffin / Vaseline Mixture

6.2 Model Description

6.2.1 Viscosity Model

The model was constructed to compare to the experimental setup under construction. It consists of four 35 cm long, co-axial cylinders with an outer diameter of 32 mm. The inner diameter is 10 mm and the second cylinder has the reamer diameter of 11 mm. The

third cylinder has a diameter of 16mm. The innermost cylinder is describing the marrow while the second cylinder is the bone that the reamer is removing. The bottom of the inner and second cylinders surface is defined in such a way that it moved with time to simulate the reamer advancing through the bone. The third cylinder was the same porous structure as the second and was the bone that is reamed away in subsequent reaming passes. The outer cylinder acted as the bone that does not get reamed. It was a porous structure of a lower porosity than the second and third cylinder. The same symmetrical cut was used as in the minipig model to ensure that the maximum accuracy was attained with the minimum amount of computing time. The model mesh used in the calculations appears in Figure 6-2.

The same temperature profile that is detailed [15] and was used in the minipig model was implemented here. The list of physical properties of the bone and marrow substitutes is presented in Table 6-1. The thermo-physical properties of the porous structures were those of human bone due to the nature of the study which is to determine any effects the viscosity change has on temperature, not to predict the temperature. Paraffin's thermo-conductivity and specific heat vary only slightly over the temperature range that is used in this study and were assumed constant [66],[67]. The values used are displayed in Table 6-1.

Table 6-1: Physical Properties of the Bone and Marrow Substitutes

Physical Properties of the Plastic Section		Physical Properties of the Foam Section		Physical Properties of the Paraffin / Vaseline Section	
Symbol	Value	Symbol	Value	Symbol	Value
ϕ	0.225	ϕ	0.85	μ	Variable
ρ	720.75 kg/m ³	ρ	270 kg/m ³	ρ	861.5 kg/m ³
κ	9.5 x 10 ⁻⁹ m ²	κ	9.5 x 10 ⁻⁷ m ²	N/A	

Figure 6-2: Viscosity Model

6.2.2 Initial and Boundary Conditions of the Viscosity Model

The boundary conditions used in the model were chosen to simulate an experimental setup that could be used to conduct temperature predictions for a reaming procedure.

- Velocity was specified as zero at the surface of the outer cylinder, as well as the top and bottom of the outer and third cylinder. The top of the inner and second

cylinder were also specified as zero. Therefore, no fluid could escape through the side walls or the top and bottom of the cylinders. The velocity boundary is therefore at these locations.

$$U_r = 0, \quad U_\theta = 0, \quad U_z = 0 \quad (6.1)$$

- At the bottom face of the innermost and second cylinder, the surface velocity is set to be equal to the insertion rate of the reamer to mimic the movement of the reamer as it advances at a constant velocity.

$$U_z = \text{insertion velocity} \quad (6.2)$$

- The initial temperature of the entire model for the reaming procedure was set as the initial temperature which was 20°C.

$$T_i = 20^\circ\text{C} \quad (6.3)$$

- The bottom face of the innermost and middle cylinder was set to change with time as the function presented in [15].

$$T_{\text{Bottom Reamer Face}} = \Phi \frac{16}{3\sqrt{\pi}} \frac{\mu_f F_n}{d_i \sqrt{\rho_l c_{pl} \lambda_l}} \sqrt{t} \quad (5.6)$$

- The wall of the outer cylinder was set as a constant of the initial temperature.

$$T_{\text{Outer Cylinder Wall}} = 20^\circ\text{C} \quad (6.4)$$

6.3 Results

6.3.1 Effect of Viscosity

The viscosity values were changed to coincide with the change of viscosity in the paraffin/Vaseline mixture when subjected to increased temperatures, as seen in Figure 6-1. The initial temperature of the model was specified to be 20°C. What is evident from Figure 6-3 is that any change in viscosity of the marrow substitute will have little effect on the overall temperature distribution in the bone. There is a slight increase in temperature of approximately 0.01°C, apparent from Figure 6-4, which presents a magnified view of the temperature difference. This temperature elevation is nominal. It would be less than the accuracy of many thermocouples. Figure 6-5 reveals that the maximum difference in the non-dimensional temperature is approximately 1.6%. During the reaming procedure a temperature increase of 20°C would be quite high and a 1.6% difference would not be noticeable.

Figure 6-3: Temperature of the Bone in the Middle of the Diaphysis 1 mm from the Reamer

Figure 6-4: Magnified view of the Temperature in the Bone in the Middle of the Diaphysis 1 mm from the Reamer Face

Figure 6-5: Maximum Percentage Change for Varying Viscosities

6.4 Discussion

Experimental procedures are being conducted to simulate the reaming of bone that occurs commonly in orthopaedic procedures. Porous plastics and hardened porous foams are known to simulate the porous structure that is found in human long bones [68]. Mixtures of different liquids have been used to simulate the properties of human bone marrow at body temperature. One of the most prevalent of these mixtures is the paraffin/Vaseline mixture that was used in studies focused on pressure build-up in the intramedullary cavity [65][40].

The investigation of pressure increases is an important topic in reaming for orthopaedic procedures, but there are other concerns as well. One of these is the temperature elevations recorded during these procedures and the unfavorable effect that they impose on the patient. If both of these concerns could be addressed in a single experimental setup, time and money would be saved as a result.

As mentioned before, a common proposed bone marrow substitute is a paraffin/Vaseline mixture. One of the concerns with using this mixture to study temperature distribution is the viscosity variation that occurs in the mixture with temperature change, as is seen in Figure 6-1. This study was designed to investigate the issues that would occur with this viscosity change and its subsequent effect on the temperature distribution. An extreme case was used and the viscosity of the marrow was changed throughout the model to the viscosity at the specified temperature. This was done to analyze the effect of viscosity change on temperature distribution in the bone in a worst case scenario.

Figures 6-3 and the magnified view in Figure 6-4 present the temperature of the bone substitute at the center of the long bone, 6.5 mm from the centre or exactly 1 mm from the edge of the reamer. Looking at these Figures, it becomes evident that the viscosity decrease is not an issue. There are slight increases in the temperature but it is clear that these increases are extremely minute.

The actual viscosity would not change this drastically throughout the intramedullary cavity only close to the reamer face, and 20°C would be an extreme temperature increase.

It can therefore be stated that the viscosity variation would have no discernable effect in an experiment with the restricted temperature ranges that are seen while reaming in orthopaedic procedures. It can also be recommend that should a 45/55 mixture of paraffin/Vaseline be used to conduct studies on the thermal effect of reaming; the viscosity variation that occurs due to temperature increase would not become a complication that could affect the obtained results.

Chapter 7 : Conclusion

Several different types of problems were looked at in this study. First, the effect of rotation on the pressure buildup in the intramedullary cavity was analyzed. This study produced results that revealed that the rate of rotation of the reamer had little bearing on the elevated pressure once the reamer flutes become clogged and congealed. Next, the thermal effects on the cortical bone that are associated with reaming in orthopedic procedures were analyzed. It was also determined through this research that the temperature profile in the cortical bone can be accurately modeled through the use of numerical tools such as FIDAP. This will lead to the implementation of the most effective and efficient reaming procedure without affecting the safety of the patient. The use of a paraffin/Vaseline mixture to model bone marrow was shown to have no adverse affects on calculating the temperature distribution throughout the bone even though the mixture has viscosity variations over the temperature ranges that would be seen during the experiment.

Chapter 8 : Future Work

Implementing a more complex and realistic model of the femur or tibia bone into the numerical simulation would be beneficial to the accuracy of the results of both the pressure and temperature calculations. The implementation of a reaming procedure that incorporated the amount of debris that is removed during the procedure would be beneficial in modeling a more accurate pressure rise for cases when the flutes of the reamer are not clogged. This, along with simulating different reamer designs, would allow calculations of changes in pressure due to different reamer heads to be modeled accurately and result in the discovery of the most acute reamer. The calculation of a more accurate model to describe heat generated by the reamer head, if the reamer is sharp, would allow for an even closer approximation to the actual temperature rise in the bone. This research should be accompanied by experimental data with which it can be compared. This should be done in an attempt to further validate the numerical results. Ultimately, the goal of this research is to minimize or eliminate the occurrence of FES and the damage of cells due to elevated temperatures. This will be accomplished by crafting a superior procedure to ream the intramedullary cavity without the patient sustaining any undue injury.

Appendix A : Non-Dimensionalization

Governing Equations in Radial Form

Momentum Equations:

r-component:

$$\frac{\rho}{\phi} \frac{\partial u_r}{\partial E} + \left(\frac{\mu}{\kappa_r} \right) u_r = -\frac{\partial p}{\partial r} + \mu \left[\frac{\partial}{\partial r} \frac{1}{r} \left(\frac{\partial r u_r}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2 u_r}{\partial \theta^2} - \frac{2}{r^2} \frac{\partial u_\theta}{\partial \theta} + \frac{\partial^2 u_r}{\partial z^2} \right]$$

θ-component:

$$\frac{\rho}{\phi} \frac{\partial u_\theta}{\partial t} + \left(\frac{\mu}{\kappa_\theta} \right) u_\theta = -\frac{1}{r} \frac{\partial p}{\partial \theta} + \mu \left[\frac{\partial}{\partial r} \frac{1}{r} \left(\frac{\partial r u_\theta}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2 u_\theta}{\partial \theta^2} + \frac{2}{r^2} \frac{\partial u_r}{\partial \theta} + \frac{\partial^2 u_\theta}{\partial z^2} \right]$$

z-component:

$$\frac{\rho}{\phi} \frac{\partial u_z}{\partial t} + \left(\frac{\mu}{\kappa_z} \right) u_z = -\frac{\partial p}{\partial z} + \mu \left[\frac{1}{r} \frac{\partial}{\partial r} \left(r \frac{\partial u_z}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2 u_z}{\partial \theta^2} + \frac{\partial^2 u_z}{\partial z^2} \right] + \rho g$$

Continuity Equation:

$$\frac{1}{r} \frac{\partial}{\partial r} (r u_r) + \frac{1}{r} \frac{\partial u_\theta}{\partial \theta} + \frac{\partial u_z}{\partial z} = 0$$

Energy Equation:

$$(\rho c_p)_{eff} \left(\frac{\partial T}{\partial t} + u_r \frac{\partial T}{\partial r} + u_\theta \frac{\partial T}{\partial \theta} + u_z \frac{\partial T}{\partial z} \right) = k \left(\frac{1}{r} \frac{\partial}{\partial r} \left(r \frac{\partial T}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2 T}{\partial \theta^2} + \frac{\partial^2 T}{\partial z^2} \right)$$

Non-dimensionalization:

These are all of the non-dimensional variables that will be used in simplifying and non-dimensionalizing the momentum, energy and continuity equations:

$$R = \frac{r}{L}, Z = \frac{z}{L}, U_r = \frac{u_r}{u_0}, U_\theta = \frac{u_\theta}{u_0}, U_z = \frac{u_z}{u_0}, u_0 = \sqrt{g \times L}, P = \frac{pL}{\mu u_0},$$

$$T^* = \frac{T - T_0}{\Delta T}, \tau = \frac{t u_0}{L}, a = \frac{\mu}{\mu}$$

$$Re = \frac{\rho u_0 L}{\mu}, Pr = \frac{\mu c_p}{k}, Pe_r = Re Pr, Da = \frac{\kappa}{L^2}, Fr = \frac{u_0^2}{gL}$$

Momentum Equations

R-component:

$$\frac{\rho}{\phi} \frac{\partial u_r}{\partial t} + \left(\frac{\mu}{\kappa_r} \right) u_r = - \frac{\partial p}{\partial r} + \mu \left[\frac{\partial}{\partial r} \frac{1}{r} \left(\frac{\partial r u_r}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2 u_r}{\partial \theta^2} - \frac{2}{r^2} \frac{\partial u_\theta}{\partial \theta} + \frac{\partial^2 u_r}{\partial z^2} \right]$$

$$\frac{\rho}{\phi} \frac{\partial (U_r u_0)}{\partial \left(\frac{\tau L}{u_0} \right)} + \left(\frac{\mu}{Da \times L^2} \right) U_r u_0 = - \frac{\partial \left(\frac{P \mu u_0}{L} \right)}{\partial (RL)} + \mu \left[\frac{\partial}{\partial RL} \frac{1}{RL} \left(\frac{\partial RL U_r u_0}{\partial RL} \right) + \frac{1}{(RL)^2} \frac{\partial^2 U_r u_0}{\partial \theta^2} - \frac{2}{(RL)^2} \frac{\partial U_\theta u_0}{\partial \theta} + \frac{\partial^2 U_r u_0}{\partial (ZL)^2} \right]$$

$$\begin{aligned} \frac{\rho u_0^2}{\phi L} \frac{\partial U_r}{\partial \tau} + \left(\frac{\mu}{Da \times L^2} \right) U_r u_0 = & -\frac{\mu u_0}{L^2} \frac{\partial P}{\partial R} \\ & + \bar{\mu} \left[\frac{u_0}{L^2} \frac{\partial}{\partial R} \frac{1}{R} \left(\frac{\partial R U_r}{\partial R} \right) + \frac{u_0}{L^2} \frac{1}{R^2} \frac{\partial^2 U_r}{\partial \theta^2} - \frac{u_0}{L^2} \frac{2}{R^2} \frac{\partial U_\theta}{\partial \theta} + \frac{u_0}{L^2} \frac{\partial^2 U_r}{\partial Z^2} \right] \end{aligned}$$

$$\begin{aligned} \frac{\rho u_0^2}{\phi L} \frac{\partial U_r}{\partial \tau} \frac{L^2}{\mu u_0} + \left(\frac{\mu}{Da \times L^2} \right) U_r u_0 \frac{L^2}{\mu u_0} = & -\frac{\partial P}{\partial R} \\ & + \frac{1}{a} \left[\frac{\partial}{\partial R} \frac{1}{R} \left(\frac{\partial R U_r}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 U_r}{\partial \theta^2} - \frac{2}{R^2} \frac{\partial U_\theta u_0}{\partial \theta} + \frac{\partial^2 U_r}{\partial Z^2} \right] \end{aligned}$$

$$\frac{Re}{\phi} \frac{\partial U_r}{\partial \tau} + \left(\frac{1}{Da} \right) U_r = -\frac{\partial P}{\partial R} + \frac{1}{a} \left[\frac{\partial}{\partial R} \frac{1}{R} \left(\frac{\partial R U_r}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 U_r}{\partial \theta^2} - \frac{2}{R^2} \frac{\partial U_\theta u_0}{\partial \theta} + \frac{\partial^2 U_r}{\partial Z^2} \right]$$

$$\frac{Re}{\phi} \frac{\partial U_r}{\partial \tau} + \left(\frac{1}{Da} \right) U_r = -\frac{\partial P}{\partial R} + \frac{1}{a} \left[\frac{\partial}{\partial R} \frac{1}{R} \left(\frac{\partial R U_r}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 U_r}{\partial \theta^2} - \frac{2}{R^2} \frac{\partial U_\theta u_0}{\partial \theta} + \frac{\partial^2 U_r}{\partial Z^2} \right]$$

0-component:

$$\begin{aligned} \frac{\rho}{\phi} \frac{\partial u_\theta}{\partial t} + \left(\frac{\mu}{\kappa_\theta} \right) u_\theta = & -\frac{1}{r} \frac{\partial p}{\partial \theta} + \bar{\mu} \left[\frac{\partial}{\partial r} \frac{1}{r} \left(\frac{\partial r u_\theta}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2 u_\theta}{\partial \theta^2} + \frac{2}{r^2} \frac{\partial u_r}{\partial \theta} + \frac{\partial^2 u_\theta}{\partial Z^2} \right] \\ \frac{\rho}{\phi} \frac{\partial (U_\theta u_0)}{\partial \left(\frac{\tau L}{u_0} \right)} + \left(\frac{\mu}{Da \times L^2} \right) U_\theta u_0 = & -\frac{1}{RL} \frac{\partial \left(\frac{P \mu u_0}{L} \right)}{\partial \theta} + \bar{\mu} \left[\frac{\partial}{\partial RL} \frac{1}{RL} \left(\frac{\partial RL U_\theta u_0}{\partial RL} \right) \right. \\ & \left. + \frac{1}{(RL)^2} \frac{\partial^2 U_\theta u_0}{\partial \theta^2} + \frac{2}{(RL)^2} \frac{\partial U_r u_0}{\partial \theta} + \frac{\partial^2 U_\theta u_0}{\partial (ZL)^2} \right] \end{aligned}$$

$$\begin{aligned}
& \frac{\rho u_o^2}{\phi L} \frac{\partial U_o}{\partial \tau} + \left(\frac{\mu}{Da \times L^2} \right) U_o u_o = - \frac{u_o \mu}{L^2} \frac{1}{R} \frac{\partial P}{\partial \theta} \\
& \quad + \bar{\mu} \left[\frac{u_o}{L^2} \frac{\partial}{\partial R} \frac{1}{R} \left(\frac{\partial R U_o}{\partial R} \right) + \frac{u_o}{L^2} \frac{1}{R^2} \frac{\partial^2 U_o}{\partial \theta^2} + \frac{u_o}{L^2} \frac{2}{R^2} \frac{\partial U_r}{\partial \theta} + \frac{u_o}{L^2} \frac{\partial^2 U_o}{\partial Z^2} \right] \\
& \frac{\rho u_o^2}{\phi L} \frac{\partial U_o}{\partial \tau} + \left(\frac{\mu}{Da \times L^2} \right) U_o u_o = - \frac{u_o \mu}{L^2} \frac{1}{R} \frac{\partial P}{\partial \theta} + \frac{u_o \bar{\mu} \mu}{\mu L^2} \left[\frac{\partial}{\partial R} \frac{1}{R} \left(\frac{\partial R U_o}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 U_o}{\partial \theta^2} + \frac{2}{R^2} \frac{\partial U_r}{\partial \theta} + \frac{\partial^2 U_o}{\partial Z^2} \right] \\
& \frac{\rho u_o^2}{\phi L} \frac{\partial U_o}{\partial \tau} \frac{L^2}{u_o \mu} + \left(\frac{\mu}{Da \times L^2} \right) U_o u_o \frac{L^2}{u_o \mu} = - \frac{1}{R} \frac{\partial P}{\partial \theta} + \frac{1}{a} \left[\frac{\partial}{\partial R} \frac{1}{R} \left(\frac{\partial R U_o}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 U_o}{\partial \theta^2} + \frac{2}{R^2} \frac{\partial U_r}{\partial \theta} + \frac{\partial^2 U_o}{\partial Z^2} \right]
\end{aligned}$$

$$\frac{Re}{\phi} \frac{\partial U_o}{\partial \tau} + \left(\frac{1}{Da} \right) U_o = - \frac{1}{R} \frac{\partial P}{\partial \theta} + \frac{1}{a} \left[\frac{\partial}{\partial R} \frac{1}{R} \left(\frac{\partial R U_o}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 U_o}{\partial \theta^2} + \frac{2}{R^2} \frac{\partial U_r}{\partial \theta} + \frac{\partial^2 U_o}{\partial Z^2} \right]$$

Z-component:

$$\begin{aligned}
& \frac{\rho}{\phi} \frac{\partial u_o}{\partial t} + \left(\frac{\mu}{\kappa_z} \right) u_z = - \frac{\partial p}{\partial z} + \bar{\mu} \left[\frac{1}{r} \frac{\partial}{\partial r} \left(\frac{r \partial u_z}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2 u_z}{\partial \theta^2} + \frac{\partial^2 u_z}{\partial z^2} \right] + \rho g \\
& \frac{\rho}{\phi} \frac{\partial U_o u_o}{\partial \left(\frac{\tau L}{u_o} \right)} + \left(\frac{\mu}{Da \times L^2} \right) U_z u_o = - \frac{\partial \left(\frac{P \mu u_o}{L} \right)}{\partial ZL} \\
& \quad + \bar{\mu} \left[\frac{1}{RL} \frac{\partial}{\partial RL} \left(\frac{RL \partial U_z u_o}{\partial RL} \right) + \frac{1}{(RL)^2} \frac{\partial^2 U_z u_o}{\partial \theta^2} + \frac{\partial^2 U_z u_o}{\partial (ZL)^2} \right]
\end{aligned}$$

$$\frac{\rho u_o^2}{\phi L} \frac{\partial U_o}{\partial \tau} + \left(\frac{\mu}{Da \times L^2} \right) U_z u_o = - \frac{\partial \left(\frac{P \mu u_o}{L} \right)}{\partial ZL} + \bar{\mu} \left[\frac{u_o}{L^2} \frac{1}{R} \frac{\partial}{\partial R} \left(\frac{R \partial U_z}{\partial R} \right) + \frac{u_o}{L^2} \frac{1}{R^2} \frac{\partial^2 U_z}{\partial \theta^2} + \frac{u_o}{L^2} \frac{\partial^2 U_z}{\partial Z^2} \right] + \frac{\rho u_o L}{\mu} Fr^{-1}$$

$$\frac{\rho u_o^2}{\phi L} \frac{\partial U_o}{\partial \tau} + \left(\frac{\mu}{Da \times L^2} \right) U_z u_o = - \frac{\mu u_o}{L^2} \frac{\partial P}{\partial Z} + \frac{\mu \bar{\mu} u_o}{\mu L^2} \left[\frac{1}{R} \frac{\partial}{\partial R} \left(\frac{R \partial U_z}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 U_z}{\partial \theta^2} + \frac{\partial^2 U_z}{\partial Z^2} \right] + \frac{\rho u_o L}{\mu} Fr^{-1}$$

$$\frac{\rho u_o^2}{\phi L} \frac{\partial U_\theta}{\partial \tau} \frac{L^2}{\mu u_o} + \left(\frac{\mu}{Da \times L^2} \right) U_z u_o \frac{L^2}{\mu u_o} = -\frac{\partial P}{\partial Z} + \frac{1}{a} \left[\frac{1}{R} \frac{\partial}{\partial R} \left(\frac{R \partial U_z}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 U_z}{\partial \theta^2} + \frac{\partial^2 U_z}{\partial Z^2} \right] + \frac{\rho u_o L}{\mu} Fr^{-1}$$

$$\frac{Re}{\phi} \frac{\partial U_\theta}{\partial \tau} + \left(\frac{1}{Da} \right) U_z = -\frac{\partial P}{\partial Z} + \frac{1}{a} \left[\frac{1}{R} \frac{\partial}{\partial R} \left(\frac{R \partial U_z}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 U_z}{\partial \theta^2} + \frac{\partial^2 U_z}{\partial Z^2} \right] + Re Fr^{-1}$$

Continuity Equation:

$$\frac{1}{r} \frac{\partial}{\partial r} (r u_r) + \frac{1}{r} \frac{\partial u_\theta}{\partial \theta} + \frac{\partial u_z}{\partial Z} = 0$$

$$\frac{1}{RL} \frac{\partial}{\partial RL} (RL U_r u_o) + \frac{1}{RL} \frac{\partial U_\theta u_o}{\partial \theta} + \frac{\partial U_z u_o}{\partial ZL} = 0$$

$$\frac{1}{R} \frac{\partial}{\partial R} (R U_r) + \frac{1}{R} \frac{\partial U_\theta}{\partial \theta} + \frac{\partial U_z}{\partial Z} = 0$$

Energy Equation:

$$(\rho c_p)_{eff} \left(\frac{\partial T}{\partial t} + u_r \frac{\partial T}{\partial r} + u_\theta \frac{1}{r} \frac{\partial T}{\partial \theta} + u_z \frac{\partial T}{\partial z} \right) = k \left(\frac{1}{r} \frac{\partial}{\partial r} \left(r \frac{\partial T}{\partial r} \right) + \frac{1}{r^2} \frac{\partial^2 T}{\partial \theta^2} + \frac{\partial^2 T}{\partial z^2} \right)$$

$$\begin{aligned} & (\rho c_p)_{eff} \left(\frac{\partial (T_0 + \Delta T T^*)}{\partial \left(\frac{\tau L}{u_o} \right)} + (U_r u_o) \frac{\partial (T_0 + \Delta T T^*)}{\partial (RL)} + (U_\theta u_o) \frac{1}{(RL)} \frac{\partial (T_0 + \Delta T T^*)}{\partial \theta} + (U_z u_o) \frac{\partial (T_0 + \Delta T T^*)}{\partial (ZL)} \right) \\ & = k \left(\frac{1}{(RL)} \frac{\partial}{\partial (RL)} \left((RL) \frac{\partial (T_0 + \Delta T T^*)}{\partial r} \right) + \frac{1}{(RL)^2} \frac{\partial^2 (T_0 + \Delta T T^*)}{\partial \theta^2} + \frac{\partial^2 (T_0 + \Delta T T^*)}{\partial (ZL)^2} \right) \end{aligned}$$

$$\frac{L^2 u_o (\rho c_p)_{eff}}{kL} \left(\frac{\partial T^*}{\partial \tau} + U_r \frac{\partial T^*}{\partial R} + U_\theta \frac{1}{R} \frac{\partial T^*}{\partial \theta} + U_z \frac{\partial T^*}{\partial Z} \right) = \left(\frac{1}{R} \frac{\partial}{\partial R} \left(R \frac{\partial T^*}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 T^*}{\partial \theta^2} + \frac{\partial^2 T^*}{\partial Z^2} \right)$$

$$\text{Re Pr} \left(\frac{\partial T^*}{\partial \tau} + U_r \frac{\partial T^*}{\partial R} + U_\theta \frac{\partial T^*}{\partial \theta} + U_z \frac{\partial T^*}{\partial Z} \right) = \left(\frac{1}{R} \frac{\partial}{\partial R} \left(R \frac{\partial T^*}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 T^*}{\partial \theta^2} + \frac{\partial^2 T^*}{\partial Z^2} \right)$$

$$\text{Pe}_r \left(\frac{\partial T^*}{\partial \tau} + U_r \frac{\partial T^*}{\partial R} + U_\theta \frac{\partial T^*}{\partial \theta} + U_z \frac{\partial T^*}{\partial Z} \right) = \left(\frac{1}{R} \frac{\partial}{\partial R} \left(R \frac{\partial T^*}{\partial R} \right) + \frac{1}{R^2} \frac{\partial^2 T^*}{\partial \theta^2} + \frac{\partial^2 T^*}{\partial Z^2} \right)$$

Appendix B : Minipig Bone Calculations

The work of Garcia *et al* [53] used 14 minipigs to conduct an *in vivo* reaming procedure that measured the maximum temperature caused by the reaming procedure. Unfortunately, the length of the tibia and the outer diameter were not specified in the paper, and several contact attempts with the author's resulted in no response. It was therefore required to make certain assumptions. One of these included assuming that the minipig was approximately one year old, as was seen in other studies conducted using minipig bones [55][54][56]. This enabled the measurement of minipig tibia seen in [55]. The intramedullary nail that is seen in the Figures B-1(a) and (b), was stated to be 5mm in diameter. This allowed the use of ImageJ software to use the nail as a calibration and measure the length and outer diameter of the bone.

(a) Posterior-anterior view

(b) Lateral View.

Figure B-1: X-ray of minipig tibia. [55]

A minipig tibia cross section is not a circle, but is elliptical in shape. The lengths of the ellipse dimensions were calculated and then used to approximate a circular radius.

Appendix C : Effect of Axial Force

The one parameter that was not given in [53] was the axial force that was used during the procedure. As is stated in Chapter 5, varying axial forces are known to be used during reaming procedures. To evaluate these forces and what effect they have on the overall result, three forces were examined: 20N, 40N and 60N. The same procedure as is seen throughout Chapter 5 was conducted with a varying axial force. First the 8 mm reamer was used then the 9 mm. The results for the 9 mm reamer are displayed below in Figure C-1.

Figure C-1: Effect of Axial Force on the Extent of Temperature Elevation.

What can be discerned from Figure A-1 is that the all of the axial forces that are used in the references fall within the acceptable range of either the sharp or blunt reamer

maximum temperatures. Due to the fact that procedure of [53] was conducted in vivo and that there was no experimental apparatus setup, the axial force used over the course of the procedure would be prone to vary. Therefore, due to this variation, a constant median value of 40N was used in this study.

Appendix D : FIDAP Code Used for Model

FIDAP Code Used in Chapter 3

```

/ *****
/ Disclaimer: This file was written by GAMBIT and contains
/ all the continuum and boundary entities and coordinate systems
/ defined in GAMBIT. Additionally, some frequently used FIPREP
/ commands are added. Modify/Add/Uncomment any necessary commands.
/ Refer to FIPREP documentation for complete listing of commands.
/ *****
/
/      CONVERSION OF NEUTRAL FILE TO FIDAP Database
/
FICONV( NEUTRAL )
INPUT( FILE="b9.FDNEUT" )
OUTPUT( DELETE )
END
/
TITLE
b9
/
FIPREP
/
/      PROBLEM SETUP
/
PROBLEM ( 3-D, NONLINEAR, FREE, TRANSIENT )
EXECUTION( NEWJOB )
PRINTOUT( NONE )
DATAPRINT( CONTROL )
CLIPPING ( MINIMUM )
0 1E-20 0 1E-20
/
/
/      CONTINUUM ENTITIES
/
ENTITY ( NAME = "BONE", POROUS, PROPERTY = "BONE", MAPERM = "BONE" )
ENTITY ( NAME = "MARROW", POROUS, PROPERTY = "MARROW", MAPERM = "MARROW" )
/
/
/      BOUNDARY ENTITIES
/
ENTITY ( NAME = "TOP_MARROW", PLOT )
ENTITY ( NAME = "PROFILE_MARROW", PLOT )
ENTITY ( NAME = "TOP_BONE", PLOT )
ENTITY ( NAME = "BOTTOM_BONE", PLOT )
ENTITY ( NAME = "PROFILE_BONE", PLOT )
ENTITY ( NAME = "BOTTOM_MARROW", SURFACE, SPINE, PREF, DEPTH = 0, X = 0, Y = 0, Z = 1 )
/
/
/      SOLUTION PARAMETERS
/
/
SOLUTION( SEGREGATED = 1000, KINEMATIC = 5, VELC = 0.001, SURFACE = 0.001, CGS = 1000, CR = 1500, NCGCONV
= 1.E-5, SCGCONV = 1.E-5 )
OPTIONS( NOJACOBI )
EXTRAPOLATION( NOFREE )
PRESSURE( MIXED = 1.E-12, DISCONTINUOUS )
OPTIONS( UPWINDING )
UPWINDING
1 1 1 0 0 0 0 0
POSTPROCESS( NBLOCKS = 2 )
1 1 1
5 400 5
TIMEINTEGRATION( BACKWARD, FIXED, NSTEPS=400, TSTART=0, TEND=0.003591, DT=0.00017955 )
/
/      MATERIAL PROPERTIES
```

```

/
/ Partial list of Material Properties data
/
DENSITY( SET = "BONE", CONSTANT = 0.533571 )
/
VISCOSITY( SET = "BONE", CONSTANT = 21 )
/
GRAVITY( MAGNITUDE = 1 )
/
PERMEABILITY( SET = "BONE", ACOEF, CONSTANT = 100, X = 77.55102E-9, Y = 77.55102E-9, Z = 77.55102E-9, POROSITY
= 0.05 )
/
DENSITY( SET = "MARROW", CONSTANT = 0.533571 )
/
VISCOSITY( SET = "MARROW", CONSTANT = 0.01 )
/
GRAVITY( MAGNITUDE = 1 )
/
PERMEABILITY( SET = "MARROW", ACOEF, CONSTANT = 100, X = 7755.102E-9, Y = 7755.102E-9, Z = 7755.102E-9,
POROSITY = 0.95 )
/
/ INITIAL AND BOUNDARY CONDITIONS
BCNODE( VELOCITY, CONSTANT = 0, ENTITY = "TOP_MARROW")
BCNODE( VELOCITY, CONSTANT = 0, ENTITY = "TOP_BONE")
/

BCNODE(UX, ENTITY = "BOTTOM_MARROW", POLYNOMIAL = 1, CARTESIAN)
0 9885.03 0 1 0

BCNODE(UY, ENTITY = "BOTTOM_MARROW", POLYNOMIAL = 1, CARTESIAN)
0 -9885.03 1 0 0

/BCNODE(UX, ZERO, ENTITY = "BOTTOM_MARROW")

/BCNODE(UY, ZERO, ENTITY = "BOTTOM_MARROW")

/
BCNODE(UZ, CONSTANT = 1, CURVE = 1, ENTITY = "BOTTOM_MARROW", FACTOR = 1)

BCNODE(SURFACE, CONSTANT = 1, CURVE = 2, ENTITY = "BOTTOM_MARROW", FACTOR = 1)
/

TMFUNCTION(SET=1,NPOINTS=21)
0 , 139.23 0.00017955 , 139.23 0.00035911 , 139.23 0.00053866 , 139.23
0.00071821 , 139.23 0.00089776 , 139.23 0.00107731 , 139.23 0.00125686 , 139.23
0.00143641 , 139.23 0.00161596 , 139.23 0.00179551 , 139.23 0.00197506 , 139.23
0.00215461 , 139.23 0.00233416 , 139.23 0.00251371 , 139.23 0.00269326 , 139.23
0.00287281 , 139.23 0.00305236 , 139.23 0.00323191 , 139.23 0.00341146 , 139.23
0.003591 , 139.23
/
TMFUNCTION(SET=2,NPOINTS= 21)
/ TIME | LOCATION | TIME | LOCATION | TIME | LOCATION | TIME | LOCATION
0 , 0 0.00017955 , 0.025 0.00035911 , 0.05 0.00053866 , 0.075
0.00071821 , 0.1 0.00089776 , 0.125 0.00107731 , 0.15 0.00125686 , 0.175
0.00143641 , 0.2 0.00161596 , 0.225 0.00179551 , 0.25 0.00197506 , 0.275
0.00215461 , 0.3 0.00233416 , 0.325 0.00251371 , 0.35 0.00269326 , 0.375
0.00287281 , 0.4 0.00305236 , 0.425 0.00323191 , 0.45 0.00341146 , 0.475
0.003591 , 0.5
/
END
/
CREATE( FIPREP,DELETE )
PARAMETER( LIST )
CREATE( FISOLV )
/RUN(FISOLV,BACKGROUND,FISOLVMEM=80000000)
/RUN( FISOLV, FOREGROUND )

```

FIDAP Code Used in Chapter 4

```
/ *****
/ Disclaimer: This file was written by GAMBIT and contains
/ all the continuum and boundary entities and coordinate systems
/ defined in GAMBIT. Additionally, some frequently used FIPREP
/ commands are added. Modify/Add/Uncomment any necessary commands.
/ Refer to FIPREP documentation for complete listing of commands.
/ *****
/
/   CONVERSION OF NEUTRAL FILE TO FIDAP Database
/
FICONV( NEUTRAL )
INPUT( FILE="injury3.FDNEUT" )
OUTPUT( DELETE )
END
/
TITLE
injury3
/
FIPREP
/
/   PROBLEM SETUP
/

PROBLEM ( 3-D, NONLINEAR, TRANSIENT, ENERGY )
EXECUTION( NEWJOB )
PRINTOUT( NONE )
DATAPRINT( CONTROL )
/
/   CONTINUUM ENTITIES
/
ENTITY ( NAME = "BONE", POROUS, PROPERTY = "BONE" )

/
/   BOUNDARY ENTITIES
ENTITY ( NAME = "BOTTOM_BONE", PLOT )
ENTITY ( NAME = "FACE_BONE1", PLOT )
ENTITY ( NAME = "PROFILE_BONE", PLOT )
ENTITY ( NAME = "FACE_BONE2", PLOT )
ENTITY ( NAME = "TOP_BONE", PLOT )
ENTITY ( NAME = "PROFILE_MARROW", PLOT )

/
SOLUTION( SEGREGATED = 1000, KINEMATIC = 5, VELC = 0.001, SURFACE = 0.001, CGS = 1000, CR = 1500,
  NCGCONV = 1.E-5, SCGCONV = 1.E-5 )
OPTIONS(NOJACOBI)
EXTRAPOLATION(NOFREE)
PRESSURE( MIXED = 1.E-12, DISCONTINUOUS )
OPTIONS( UPWINDING )
UPWINDING
1 1 1 0 0 0 0
POSTPROCESS( NBLOCKS = 2 )
1 1 1
5 400 5
TIMEINTEGRATION( BACKWARD, FIXED, NSTEPS=400, TSTART=0, TEND=0.029713633235, DT=0.000990454441)

/   MATERIAL PROPERTIES
/
/ Partial list of Material Properties data
/
DENSITY( SET = "BONE", CONSTANT = 0.054024787699 )
/
VISCOSITY( SET = "BONE", CONSTANT = 0.01 )
/
GRAVITY( MAGNITUDE = 1 )
/
```



```

PERMEABILITY( SET = "BONE", ACOEF, CONSTANT=100, X = 77.55102E-9, Y = 77.55102E-9, Z = 77.55102E-9, POROSITY
= 0.3 )
/
CONDUCTIVITY(SET= "BONE",CONSTANT=1)
/
SPECIFICHEAT(SET="BONE",CONSTANT=1287)
/INITIAL AND BOUNDARY CONDITIONS
/
/
ICNODE(TEMPERATURE, CONSTANT = 0 , ALL)

BCNODE( VELOCITY, CONSTANT = 0, ENTITY = "TOP_BONE")

BCNODE(UY, ZERO, ENTITY = "FACE_BONE1")
BCNODE(UY, ZERO, ENTITY = "FACE_BONE2")

BCNODE(TEMPERATURE, CONSTANT = 1, ENTITY = "PROFILE_MARROW", FACTOR =1)
BCNODE(TEMPERATURE, CONSTANT = 0, ENTITY = "PROFILE_BONE", FACTOR =1)

TMFUNCTION(SET=1,NPOINTS=31)
/TIME | TEMPERTAURE | TIME | TEMPERTAURE | TIME | TEMPERTAURE | TIME |
TEMPERTAURE

0.000000000000,0.000000000000 0.000990454441,0.000000000000 0.001980908882,0.000000000000
0.002971363323,1.000000000000
0.003961817765,1.000000000000 0.004952272206,1.000000000000 0.005942726647,1.000000000000
0.006933181088,1.000000000000
0.007923635529,1.000000000000 0.008914089970,1.000000000000 0.009904544412,1.000000000000
0.010894998853,1.000000000000
0.011885453294,1.000000000000 0.012875907735,1.000000000000 0.013866362176,1.000000000000
0.014856816617,1.000000000000
0.015847271058,1.000000000000 0.016837725500,1.000000000000 0.017828179941,1.000000000000
0.018818634382,1.000000000000
0.019809088823,1.000000000000 0.020799543264,1.000000000000 0.021789997705,1.000000000000
0.022780452147,1.000000000000
0.023770906588,1.000000000000 0.024761361029,1.000000000000 0.025751815470,1.000000000000
0.026742269911,1.000000000000
0.027732724352,1.000000000000 0.028723178793,1.000000000000 0.029713633235,1.000000000000
/

END
/
CREATE( FIPREP,DELETE )
PARAMETER( LIST )
CREATE( FISOLV )
/RUN(FISOLV,BACKGROUND,FISOLVMEM=80000000)
/RUN( FISOLV, FOREGROUND )

```

FIDAP Code Used in Chapter 5

```
*****
/ Disclaimer: This file was written by GAMBIT and contains
/ all the continuum and boundary entities and coordinate systems
/ defined in GAMBIT. Additionally, some frequently used FIPREP
/ commands are added. Modify/Add/Uncomment any necessary commands.
/ Refer to FIPREP documentation for complete listing of commands.
/ *****
/
/   CONVERSION OF NEUTRAL FILE TO FIDAP Database
/
FICONV( NEUTRAL )
INPUT( FILE="pig65.FDNEUT" )
OUTPUT( DELETE )
END
/
TITLE
pig65
/
FIPREP
/
/   PROBLEM SETUP
/
PROBLEM ( 3-D, NONLINEAR, FREE, TRANSIENT, ENERGY )
EXECUTION( NEWJOB )
PRINTOUT( NONE )
DATAPRINT( CONTROL )
/
/   CONTINUUM ENTITIES
/
ENTITY ( NAME = "MARROW", POROUS, PROPERTY = "MARROW",MAPERM = "MARROW" )
ENTITY ( NAME = "MID", POROUS, PROPERTY = "MID",MAPERM = "MID" )
ENTITY ( NAME = "BONE", POROUS, PROPERTY = "BONE",MAPERM = "BONE" )
/
/   BOUNDARY ENTITIES
/
ENTITY ( NAME = "TOP_MARROW", PLOT )
ENTITY ( NAME = "TOP_MID", PLOT )
ENTITY ( NAME = "TOP_BONE", PLOT )
ENTITY ( NAME = "BOTTOM_MARROW", SURFACE, SPINE, PREF, DEPTH = 0, X = 0, Y = 0, Z = 1 )
ENTITY ( NAME = "FACE_MARROW1", PLOT )
ENTITY ( NAME = "FACE_MARROW2", PLOT )
ENTITY ( NAME = "PROFILE_MARROW", PLOT )
ENTITY ( NAME = "BOTTOM_MID", SURFACE, SPINE, PREF, DEPTH = 0, X = 0, Y = 0, Z = 1 )
ENTITY ( NAME = "FACE_MID1", PLOT )
ENTITY ( NAME = "PROFILE_MID", PLOT )
ENTITY ( NAME = "FACE_MID2", PLOT )
ENTITY ( NAME = "BOTTOM_BONE", PLOT )
ENTITY ( NAME = "FACE_BONE1", PLOT )
ENTITY ( NAME = "PROFILE_BONE", PLOT )
ENTITY ( NAME = "FACE_BONE2", PLOT )
/
SOLUTION( SEGREGATED = 1000, KINEMATIC = 5, VELC = 0.001, SURFACE = 0.001, CGS = 1000, CR = 1500,
NCGCONV = 1.E-5, SCGCONV = 1.E-5 )
OPTIONS(NOJACOBI)
EXTRAPOLATION(NOFREE)
PRESSURE( MIXED = 1.E-12, DISCONTINUOUS )
OPTIONS( UPWINDING )
UPWINDING
1 1 1 0 0 0 0 0
POSTPROCESS( NBLOCKS = 2 )
1 1 1
5 400 5
TIMEINTEGRATION( BACKWARD, FIXED, NSTEPS=400, TSTART=0, TEND=270.683781456460, DT=1.353418907282)

/   MATERIAL PROPERTIES
/
```

```

/ Partial list of Material Properties data
/
DENSITY( SET = "BONE", CONSTANT = 1250.580601993278)
/
VISCOSITY( SET = "BONE", CONSTANT = 0.01 )
/
GRAVITY( MAGNITUDE = 1 )
/
PERMEABILITY( SET = "BONE", ACOEF, CONSTANT = 100, X = 77.55102E-9, Y = 77.55102E-9, Z = 77.55102E-9, POROSITY
= 0.05 )
/
CONDUCTIVITY(SET="BONE",CONSTANT=1)
/
SPECIFICHEAT(SET="BONE",CONSTANT=706.75)

DENSITY( SET = "MARROW", CONSTANT = 1250.580601993278)
/
VISCOSITY( SET = "MARROW", CONSTANT = 0.01 )
/
GRAVITY( MAGNITUDE = 1 )
/
PERMEABILITY( SET = "MARROW", ACOEF, CONSTANT = 100, X = 77.55102E-9, Y = 77.55102E-9, Z = 77.55102E-9, POROSITY
= 0.95 )
/
CONDUCTIVITY(SET="MARROW",CONSTANT=1)
/
SPECIFICHEAT(SET="MARROW",CONSTANT=706.75)
/
DENSITY( SET = "MID", CONSTANT = 1250.580601993278)
/
VISCOSITY( SET = "MID", CONSTANT = 0.01 )
/
GRAVITY( MAGNITUDE = 1 )
/
PERMEABILITY( SET = "MID", ACOEF, CONSTANT = 100, X = 77.55102E-9, Y = 77.55102E-9, Z = 77.55102E-9, POROSITY
= 0.05 )
/
CONDUCTIVITY(SET="MID",CONSTANT=1)
/
SPECIFICHEAT(SET="MID",CONSTANT=706.75)
/INITIAL AND BOUNDARY CONDITIONS
/
/
ICNODE(TEMPERATURE, CONSTANT = 0 , ALL)

BCNODE( VELOCITY, CONSTANT = 0, ENTITY = "TOP_MARROW")
BCNODE( VELOCITY, CONSTANT = 0, ENTITY = "TOP_BONE")
BCNODE( VELOCITY, CONSTANT = 0, ENTITY = "TOP_MID")
/

BCNODE(UX, ENTITY = "BOTTOM_MARROW", POLYNOMIAL = 1, CARTESIAN)
0 -298000 0 1 0

BCNODE(UY, ENTITY = "BOTTOM_MARROW", POLYNOMIAL = 1, CARTESIAN)
0 298000 1 0 0

BCNODE(UX, ENTITY = "BOTTOM_MID", POLYNOMIAL = 1, CARTESIAN)
0 -298000 0 1 0

BCNODE(UY, ENTITY = "BOTTOM_MID", POLYNOMIAL = 1, CARTESIAN)
0 298000 1 0 0

BCNODE(UX, ZERO, ENTITY = "BOTTOM_MID")
BCNODE(UY, ZERO, ENTITY = "BOTTOM_MID")

BCNODE(UX, ZERO, ENTITY = "BOTTOM_MARROW")
BCNODE(UY, ZERO, ENTITY = "BOTTOM_MARROW")

BCNODE(UY, ZERO, ENTITY = "FACE_BONE1")
BCNODE(UY, ZERO, ENTITY = "FACE_BONE2")

```

```
BCNODE(UY, ZERO, ENTITY = "FACE_MID1")
BCNODE(UY, ZERO, ENTITY = "FACE_MID2")
```

```
/
BCNODE(UZ , CONSTANT = 1, CURVE = 1, ENTITY = "BOTTOM_MARROW", FACTOR =1)
BCNODE(SURFACE, CONSTANT = 1, CURVE = 2, ENTITY = "BOTTOM_MARROW", FACTOR =1)
/
BCNODE(UZ , CONSTANT = 1, CURVE = 3, ENTITY = "BOTTOM_MID", FACTOR =1)
BCNODE(SURFACE, CONSTANT = 1, CURVE = 4, ENTITY = "BOTTOM_MID", FACTOR =1)
```

```
TMFUNCTION(SET=1,NPOINTS=201)
/ TIME | VELOCITY | TIME | VELOCITY | TIME | VELOCITY | TIME | VELOCITY
0.000000000000,0.000000000000 1.353418907282,0.003324912912 2.706837814565,0.003324912912
```

(Intermediate time step location and values removed for paper conservation available upon request)

```
269.330362549177,0.003324912912 270.683781456460,0.003324912912
/
TMFUNCTION(SET=2,NPOINTS= 201)
/ TIME | LOCATION | TIME | LOCATION | TIME | LOCATION | TIME | LOCATION
0.000000000000,0.000000000000 1.353418907282,0.004500000000 2.706837814565,0.009000000000
```

(Intermediate time step location and values removed for paper conservation available upon request)

```
270.683781456460,0.900000000000
```

```
/
TMFUNCTION(SET=3,NPOINTS=201)
/ TIME | VELOCITY | TIME | VELOCITY | TIME | VELOCITY | TIME | VELOCITY
0.000000000000,0.000000000000 1.353418907282,0.003324912912 2.706837814565,0.003324912912
```

(Intermediate time step location and values removed for paper conservation available upon request)

```
270.683781456460,0.003324912912
TMFUNCTION(SET=4,NPOINTS= 201)
/ TIME | LOCATION | TIME | LOCATION | TIME | LOCATION | TIME | LOCATION
0.000000000000,0.000000000000 1.353418907282,0.004500000000 2.706837814565,0.009000000000
```

(Intermediate time step location and values removed for paper conservation available upon request)

```
270.683781456460,0.900000000000
```

```
BCNODE(TEMPERATURE, CONSTANT = 1, CURVE = 5, ENTITY = "BOTTOM_MARROW", FACTOR =1)
BCNODE(TEMPERATURE, CONSTANT = 1, CURVE = 6, ENTITY = "BOTTOM_MID" , FACTOR =1)
BCNODE(TEMPERATURE, CONSTANT = 0, ENTITY = "PROFILE_BONE", FACTOR =1)
/
TMFUNCTION(SET=5,NPOINTS=201)
/ TIME | TEMPERTAURE | TIME | TEMPERTAURE | TIME | TEMPERTAURE | TIME |
TEMPERTAURE
```

```
0.000000000000,0.000000000000 1.353418907282,0.070710678119 2.706837814565,0.100000000000
```

(Intermediate time step location and values removed for paper conservation available upon request)

```
270.683781456460,1.000000000000
```

```
/
TMFUNCTION(SET=6,NPOINTS=201)
```

```
/TIME |TEMPERTAURE | TIME | TEMPERTAURE| TIME | TEMPERTAURE | TIME |  
TEMPERTAURE
```

```
0.000000000000,0.000000000000 1.353418907282,0.070710678119 2.706837814565,0.100000000000
```

(Intermediate time step location and values removed for paper conservation available upon request)

```
270.683781456460,1.000000000000
```

```
END  
/  
CREATE( FIPREP,DELETE )  
PARAMETER( LIST )  
CREATE( FISOLV )  
/RUN(FISOLV,BACKGROUND,FISOLVMEM=80000000)  
/RUN( FISOLV, FOREGROUND )  
/
```

FIDAP Code Used in Chapter 6

```
/ *****
/ Disclaimer: This file was written by GAMBIT and contains
/ all the continuum and boundary entities and coordinate systems
/ defined in GAMBIT. Additionally, some frequently used FIPREP
/ commands are added. Modify/Add/Uncomment any necessary commands.
/ Refer to FIPREP documentation for complete listing of commands.
/ *****
/
/   CONVERSION OF NEUTRAL FILE TO FIDAP Database
/
FICONV( NEUTRAL )
INPUT( FILE="Azar4.FDNEUT" )
OUTPUT( DELETE )
END
/
TITLE
Azar4
/
FIPREP
/
/   PROBLEM SETUP
/
PROBLEM ( 3-D, NONLINEAR, FREE, TRANSIENT, ENERGY )
EXECUTION( NEWJOB )
PRINTOUT( NONE )
DATAPRINT( CONTROL )
/
/   CONTINUUM ENTITIES
/

ENTITY ( NAME = "MARROW", FLUID, PROPERTY = "MARROW" )
ENTITY ( NAME = "REAMER", POROUS, PROPERTY = "REAMER" )
ENTITY ( NAME = "FOAM", POROUS, PROPERTY = "FOAM" )
ENTITY ( NAME = "BONE", POROUS, PROPERTY = "BONE" )
/
/   BOUNDARY ENTITIES

ENTITY ( NAME = "BOTTOM_MARROW", SURFACE, SPINE, PREF, DEPTH = 0, X = 0, Y = 0, Z = 1 )
ENTITY ( NAME = "BOTTOM_REAMER", SURFACE, SPINE, PREF, DEPTH = 0, X = 0, Y = 0, Z = 1 )
ENTITY ( NAME = "BOTTOM_FOAM", PLOT )
ENTITY ( NAME = "BOTTOM_BONE", PLOT )
ENTITY ( NAME = "PROFILE_BONE", PLOT )
ENTITY ( NAME = "PROFILE_FOAM", PLOT )
ENTITY ( NAME = "PROFILE_REAMER", PLOT )
ENTITY ( NAME = "PROFILE_MARROW", PLOT )
ENTITY ( NAME = "TOP_BONE", PLOT )
ENTITY ( NAME = "TOP_FOAM", PLOT )
ENTITY ( NAME = "TOP_REAMER", PLOT )
ENTITY ( NAME = "TOP_MARROW", PLOT )
ENTITY ( NAME = "FACE_MARROW1", PLOT )
ENTITY ( NAME = "FACE_MARROW2", PLOT )
ENTITY ( NAME = "FACE_REAMER1", PLOT )
ENTITY ( NAME = "FACE_REAMER2", PLOT )
ENTITY ( NAME = "FACE_FOAM1", PLOT )
ENTITY ( NAME = "FACE_FOAM2", PLOT )
ENTITY ( NAME = "FACE_BONE1", PLOT )
ENTITY ( NAME = "FACE_BONE2", PLOT )
/

/   SOLUTION PARAMETERS
/
/SOLUTION( SEGREGATED = 50, CR, CGS, VELCONV = .001, NCGC = 1.E-6, SCGC = 1.E-6, SCHANG = .0 )
/PRESSURE( MIXED = 1.E-8, DISCONTINUOUS )
/RELAX( HYBRID )
/OPTIONS( UPWINDING, , , )
```

```

/SCALE( VALUE = 1 )
/TIMEINTEGRATION( BACKWARD,NSTEPS = ,DT = 0,,, )
/POSTPROCESS( NBLOCKS = )
/
SOLUTION( SEGREGATED = 1000, KINEMATIC = 5, VELC = 0.001, SURFACE = 0.001, CGS = 1000, CR = 1500,
NCGCONV = 1.E-5, SCGCONV = 1.E-5 )
OPTIONS(NOJACOBI)
EXTRAPOLATION(NOFREE)
PRESSURE( MIXED = 1.E-12, DISCONTINUOUS )
OPTIONS( UPWINDING )
UPWINDING
1 1 1 0 0 0 0 0
POSTPROCESS( NBLOCKS = 2 )
1 1 1
5 400 5
TIMEINTEGRATION( BACKWARD, FIXED, NSTEPS=400, TSTART=0, TEND=158.826050400160, DT=0.397065126000)

/      MATERIAL PROPERTIES
/
/ Partial list of Material Properties data
/

DENSITY( SET = "BONE", CONSTANT = 7949.574710133027)
/
VISCOSITY( SET = "BONE", CONSTANT = 0.01 )
/
GRAVITY( MAGNITUDE = 1 )
/
PERMEABILITY( SET = "BONE", ACOEF, CONSTANT = 100, X = 77.55102E-9, Y = 77.55102E-9, Z = 77.55102E-9, POROSITY
= 0.225 )
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CONDUCTIVITY(SET= "BONE",CONSTANT=1)
/
SPECIFICHEAT(SET="BONE",CONSTANT=188.8950000000000)

DENSITY( SET = "MARROW", CONSTANT = 9501.989056926263)
/
VISCOSITY( SET = "MARROW", CONSTANT = 0.01 )
/
GRAVITY( MAGNITUDE = 1 )
/
CONDUCTIVITY(SET= "MARROW",CONSTANT=1)
/
SPECIFICHEAT(SET="MARROW",CONSTANT=188.8950000000000)
/
DENSITY( SET = "REAMER", CONSTANT = 2977.988445003007)
/
VISCOSITY( SET = "REAMER", CONSTANT = 0.01 )
/
GRAVITY( MAGNITUDE = 1 )
/
PERMEABILITY( SET = "REAMER", ACOEF, CONSTANT = 100, X = 77.55102E-7, Y = 77.55102E-7, Z = 77.55102E-7,
POROSITY = 0.85 )
/
CONDUCTIVITY(SET= "REAMER",CONSTANT=1)
/
SPECIFICHEAT(SET="REAMER",CONSTANT=188.8950000000000)
/
DENSITY( SET = "FOAM", CONSTANT = 2977.988445003007)
/
VISCOSITY( SET = "FOAM", CONSTANT = 0.01 )
/
GRAVITY( MAGNITUDE = 1 )
/
PERMEABILITY( SET = "FOAM", ACOEF, CONSTANT = 100, X = 77.55102E-7, Y = 77.55102E-7, Z = 77.55102E-7,
POROSITY = 0.85 )
/
CONDUCTIVITY(SET= "FOAM",CONSTANT=1)
/
SPECIFICHEAT(SET="FOAM",CONSTANT=188.8950000000000)

```

/INITIAL AND BOUNDARY CONDITIONS

/

ICNODE(TEMPERATURE, CONSTANT = 0, ALL)

BCNODE(VELOCITY, CONSTANT = 0, ENTITY = "TOP_MARROW")

BCNODE(VELOCITY, CONSTANT = 0, ENTITY = "TOP_BONE")

BCNODE(VELOCITY, CONSTANT = 0, ENTITY = "TOP_FOAM")

BCNODE(VELOCITY, CONSTANT = 0, ENTITY = "TOP_REAMER")

/

BCNODE(UX, ZERO, ENTITY = "BOTTOM_REAMER")

BCNODE(UY, ZERO, ENTITY = "BOTTOM_REAMER")

BCNODE(UX, ZERO, ENTITY = "BOTTOM_MARROW")

BCNODE(UY, ZERO, ENTITY = "BOTTOM_MARROW")

BCNODE(UY, ZERO, ENTITY = "FACE_BONE1")

BCNODE(UY, ZERO, ENTITY = "FACE_BONE2")

BCNODE(UY, ZERO, ENTITY = "FACE_REAMER1")

BCNODE(UY, ZERO, ENTITY = "FACE_REAMER2")

BCNODE(UY, ZERO, ENTITY = "FACE_FOAM1")

BCNODE(UY, ZERO, ENTITY = "FACE_FOAM2")

/

BCNODE(UZ, CONSTANT = 1, CURVE = 1, ENTITY = "BOTTOM_MARROW", FACTOR = 1)

BCNODE(SURFACE, CONSTANT = 1, CURVE = 2, ENTITY = "BOTTOM_MARROW", FACTOR = 1)

/

BCNODE(UZ, CONSTANT = 1, CURVE = 3, ENTITY = "BOTTOM_REAMER", FACTOR = 1)

BCNODE(SURFACE, CONSTANT = 1, CURVE = 4, ENTITY = "BOTTOM_REAMER", FACTOR = 1)

TMFUNCTION(SET=1,NPOINTS=401)

/TIME | VELOCITY | TIME | VELOCITY | TIME | VELOCITY | TIME | VELOCITY

0.000000000000,0.000000000000 0.397065126000,0.005666576722 0.794130252001,0.005666576722

(Intermediate time step location and values removed for paper conservation available upon request)

158.826050400160,0.005666576722

/

TMFUNCTION(SET=2,NPOINTS= 401)

/TIME | LOCATION | TIME | LOCATION | TIME | LOCATION | TIME | LOCATION

0.000000000000,0.000000000000 0.397065126000,0.002250000000 0.794130252001,0.004500000000

(Intermediate time step location and values removed for paper conservation available upon request)

158.826050400160,0.900000000000

/

TMFUNCTION(SET=3,NPOINTS=401)

/TIME | VELOCITY | TIME | VELOCITY | TIME | VELOCITY | TIME | VELOCITY

0.000000000000,0.000000000000 0.397065126000,0.005666576722 0.794130252001,0.005666576722

(Intermediate time step location and values removed for paper conservation available upon request)

158.826050400160,0.005666576722

/

TMFUNCTION(SET=4,NPOINTS= 401)

/TIME | LOCATION | TIME | LOCATION | TIME | LOCATION | TIME | LOCATION

0.000000000000,0.000000000000 0.397065126000,0.002250000000 0.794130252001,0.004500000000

(Intermediate time step location and values removed for paper conservation available upon request)

158.826050400160,0.900000000000

```
BCNODE(TEMPERATURE, CONSTANT = 1, CURVE = 5, ENTITY = "BOTTOM_MARROW", FACTOR = 1)
BCNODE(TEMPERATURE, CONSTANT = 1, CURVE = 6, ENTITY = "BOTTOM_REAMER", FACTOR = 1)
BCNODE(TEMPERATURE, CONSTANT = 0, ENTITY = "PROFILE_BONE", FACTOR = 1)
/
TMFUNCTION(SET=5,NPOINTS=401)
/ TIME | TEMPERTAURE | TIME | TEMPERTAURE | TIME | TEMPERTAURE | TIME |
TEMPERTAURE
```

0.000000000000,0.000000000000 0.397065126000,0.050000000000 0.794130252001,0.070710678119

(Intermediate time step location and values removed for paper conservation available upon request)

158.826050400160,1.000000000000

```
/
TMFUNCTION(SET=6,NPOINTS=401)
/ TIME | TEMPERTAURE | TIME | TEMPERTAURE | TIME | TEMPERTAURE | TIME |
TEMPERTAURE
```

0.000000000000,0.000000000000 0.397065126000,0.050000000000 0.794130252001,0.070710678119

(Intermediate time step location and values removed for paper conservation available upon request)

158.826050400160,1.000000000000

```
END
/
CREATE( FIPREP,DELETE )
PARAMETER( LIST )
CREATE( FISOLV )
/RUN(FISOLV,BACKGROUND,FISOLVMEM=80000000)
/RUN( FISOLV, FOREGROUND )
/
```

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Glossary

Devascularization:	The interruption of the blood supply to a part of the body by the blocking or destroying of blood vessels
Diaphysis:	The shaft of a long bone.
Emboli:	The plural of embolus; an embolus is a foreign body purposefully introduced to the blood stream to block or occlude blood flow.
Embolization:	The process by which a blood vessel or organ is obstructed by an embolus or other mass.
Endosteum:	The thin layer of cells lining the medullary cavity of a bone
FES:	Fat Embolism Syndrome
Iatrogenic:	Induced in a patient by a physician's activity, manner, or therapy. Used especially of an infection or other complication of treatment.
Intramedullary:	A medical term meaning the inside of a bone. Examples include intramedullary rods used to treat bone fractures in orthopedic surgery and intramedullary tumors occurring in some forms of cancer.
Medullary:	Containing or consisting of or resembling bone marrow
Necrosis:	Death of cells or tissues through injury or disease, especially in a localized area of the body
Periosteum:	The dense fibrous membrane covering the surface of bones except at the joints and serving as an attachment for muscles and tendons
Protoplasmic:	Of or pertaining to the first formation of living bodies