M Mediouni, D Schlatterer, Amal Khoury, T von Bergen, S Shetty, M Arora, A Dhond, N Vaughan, A Volosnikov, (2017) Optimal Parameters to Avoid Thermal Necrosis During Bone Drilling: A Finite Element Analysis, Wiley Journal of Orthopaedic Research (JOR), PubMED-ID: 28181707, https://pubmed.ncbi.nlm.nih.gov/28181707/, http://onlinelibrary.wiley.com/doi/10.1002/jor.23542/epdf.

Optimal Parameters to Avoid Thermal Necrosis During Bone Drilling: A Finite Element Analysis

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Abstract

Bone drilling is a common part of orthopaedic surgery. Especially during open reduction and internal fixation (ORIF) of fractures. Reduced fracture fragments are spanned and connected with a metal plate and the fragments are drilled for placement of screws through the plate and into the bone. Bone drilling is a common and repetitive process in fracture care. Successful ORIF depends upon numerous biologic and mechanical factors. In order for the fracture to mend the internal fixation construct (metal plate and metal screws in the bone fragments) must remain stable and intact for a period of weeks or more. Loosening and failure of the ORIF construct may result for several reasons including patient non-compliance and early weight bearing. On the surgical side of the construct failure equation, inadequate reduction and fixation are two technical possibilities leading to construct failure. Construct failure may result from the fixation screws which may loosen over time leading to loss of reduction. Several reasons have been proposed for this failure pathway including; direct bone damage after bone drilling

itself leading to bone degeneration, or: (1) causing small cracks. By propagating unseen fracture lines, both of these technical drilling errors may lead to construct loosening and failure of fixation. (2) More commonly the bone drilling process itself causes excessive heat, which has been reported to lead to (3) local bone osteonecrosis. This causes loss of bone around the placed screws, ending in loss of screw purchase in the bone and (4) subsequent loosening of the bone-implant construct. How often construct failure is the result of bone drilling and local necrosis is not known. Surgeons are well aware of the potential impact heat generation may have on fracture surgery outcome. A better understanding of the variables of drilling and heat generation could lead to better drilling techniques, less construct failures, better patient outcomes and less financial strain on the health care systems charged with managing revision surgeries. A better understanding begins with some basic concepts from physics in that the heat generated from drilling is the direct result of friction between two surfaces. These variables of bone drilling which influence the magnitude of heat generated include, but are not limited to, the sharpness of the drill bit, the drill design, size and diameter which determine the surface area in contact with bone, the revolution rate of the drill, the hardness or density of the material (bone) being drilled and the advance rate of the drill, among other factors, which all influence the amount of frictional heat produced. Intra-operatively surgeons have taken precautions themselves by either irrigating the bone region being drilled or by making frequent changes to new drill bits to mitigate the chances of local thermal bone necrosis. Knowing the ideal revolution rate and best drill advance rate would also assist surgeons in avoiding or reducing bone necrosis. Knowing temperature levels during the initial (near cortex penetration) mid (medullary or cancellous bone penetration) and end (far cortical cortex penetration) stages of drilling would help to minimize the risk of thermal necrosis and identify the key drilling stages for intervention to avoid local thermal necrosis. Unfortunately the literature is void of any specific drilling guidelines on how best to drill bone and minimize heat generation. In an attempt to better understand several drilling factors we developed an elastic-plastic three-dimensional (3D) bone model and applied finite element model analysis (FEM) to various simulated drilling procedures. The FEM results clearly indicate that the depth of drilling and the drill speed both have a significant effect on the temperature during drilling procedures. The conclusion is that reduction of the feeding speed leads to a reduction in temperature. Our data suggests that reducing the feeding speed could be a simple useful and effective way to reduce drilling temperature, which could help to reduce the risk of patients developing necrosis during bone drilling, and in turn reduce the risk of ORIF construct loosening and failure. A future study planned is to determine how often the bone should be irrigated and using what exact volume of water to cool the drill bit to starting temperature.

1. Introduction/ background

The concept of drilling originally came from the Greek (trephination) which means surgery of the bone for therapeutic purposes. Cranial surgery has a history dating back thousands of years especially to the Neolithic period **[1]**. The ancient Egyptian civilization practiced dentistry **[2, 3]**, in a tomb near the river Nile two teeth were found, ingeniously connected by a gold wire passed through holes in both teeth **[4]**. This was probably made with the intention of giving support to a mobile tooth through a well-established adjacent tooth, a form of primitive splintage. Modern orthopaedic bone drilling began in 1850 with fracture fixation using instruments **[5]**. Drilling has become commonplace and is most often used during fracture fixation which is increasingly common due to the rising rate of trauma among people aged less than fifty years and an increase in fragility fractures among an ever growing elderly population. Fracture repair has become one of the most common orthopaedic surgery procedures in the USA. Drilling involves the delicate control of instruments with the numerous structures adjacent to bone including the soft tissues, nerves, skin muscle and vascular structures and the cortical bone itself. The surgeon must be able to quickly cease any advancement of the drill for avoidance of any tissue injury bony or soft tissue alike. However, the complexity of drilling is related to the complexity of fracture. Complexity also depends on fracture location in the bone since cortical bone density varies greatly from end to end (diaphyseal, metaphyseal, epiphyseal and articular), also the presence or absence of a pathological fracture (osteoporosis, primary and secondary tumor, hyperparathyroidism), or any other condition altering the bone's density and mechanical properties.

Drilling damages bone by causing small cracks which accumulate in the mineral matrix that cause osteocytes dysfunction **[6]**. The frictional heat of drilling may cause thermal necrosis of the bone. Faced with this clinical problem, many researchers have studied the temperatures associated with drilling to better understand the multitude of factors causing heat generation, with the intention to use this information to improve the drilling process by preventing and minimizing the risk of necrosis. Obtaining true temperature measurements while drilling is a difficult task and differs from bone to bone due to inherent density differences. There are many studies in the literature for drilling analysis most with contradictions and a general lack of consistency. Anderson et al. **[7]** published the first thermal change studies to forge teeth. Mathews et al. **[8]** showed that there is no change of the temperature in vitro or vivo. Hillery et al. **[9]** reported that when bone temperature rises above 55 degrees Fahrenheit for minimal time period of 30 seconds this results in permanent bone damage. The question arises: What are the drilling parameters that most influence the temperature change? According to literature, we can classify the parameters of drilling into two categories (see table 1). There are various results published for the optimum speed of drill, some researchers have shown that the increase in speed leads to the increase in the temperature **[10, 11]**. According to Augustin et al. **[12],** the temperatures above critical were recorded using 4.5 mm drill with higher drill speeds are 188, 462, 1,140 and 1,820 rpm. Others focused on low speed drilling (up to 3000 rpm) **[13]**. Only Matthews and Hirsch did not find any change in the temperature with speeds ranging 345–2900 rpm **[14]**. Using Numerical simulation, Davidson found that the maximum temperature increases with drill speed in range of 100,000–200,000 rpm **[15]**. These are revolution rates which are much higher than tools used for drilling today. Modern drilling tools have RPMs upwards of 4,000 at a full trigger pull force.

Drill feed rate is another parameter affecting the change in temperature. Drilling feed is calculated as a product of the drilling speed and torque which is indirectly the result of surface area and the force applied to advance the drill (see figure 4 and 5) **[16]**. The power is often used to compare different factors regarding drilling. The energy produced is directly related to the amount of heat generated. In order to reduce heat generation, it becomes necessary to find the optimum speed and forces to minimize the friction between metal drill bit and bone. Hsu et al. **[17]** developed a new system for automatic bone drilling in which power is equal to zero to prevent the problem of excessive protrusion of drill bit. Cordioli et al. **[18]** found that the depth of drilling affects the temperature increase of the drilled bone. Drilling time depends on the thickness of the cortical bone which represents the hardest part of bone structure. The frictional resistance offered by the compact cortical bone causes an increase in temperature more so than cancellous or osteoporotic bone **[19, 20]**.

Training simulators are increasingly popular for training novice surgeons in orthopaedic procedures **[21]**. Modelling of the drilling temperatures would be useful for improving the accuracy of these simulators.

Non-influenced parameters	**Influenced parameters**
Drill design	Speed of drill
Drill point angle	Feed rate
Drill diameter	Cutting forces
Drill material	Drilling depth

Tab1. Drilling parameters

2. Materials and Methods

In this study, we examined the acetabular fracture complexity by estimating the parameters such as cortical thickness and hardness or density. Acetabular fractures are not particularly known for construct failures per se. However they are complex injuries owing to the inherent complex geometry of the region. The operative treatment of these fractures is technically challenging. **[22, 23, 24]**. The pelvis is an Organ with the most complex three dimensional anatomy, it is heavily crowded with organs and structures, including neurovascular, gastrointestinal and genitourinary structures as well. This makes the navigation task with the drill, during surgery, a complex task. Any plunging of the drill to an adjacent organ outside the bone, during surgery, might end up with an latrogenic and in some cases a catastrophic complication. We must bear in mind that due to the anatomy of the pelvis some of the screws needed for fracture fixation are very long, reaching 130 mm in some cases, this fact makes the drilling task more complicated, and more likely to generate harmful heat the drilling time and heat generated during the drilling might be excessive. Due to this drilling complexity, it is not rare at our institution to see a broken drill bit in a post operative x-ray of a pelvis or an acetabulum after fracture fixation. For those reasons we thought that acetabular surgery would be a good representative bone to simulate in our study.

Hardware failure due to screw loosening after local thermal necrosis is theory and remains an unreported entity. It stands to reason that drilling does result in screw loosening construct failures non unions and possibly even infections, then if fixation could be improved by improved drilling techniques (IDTs) then these IDTs principles could be applied to other bones of similar thicknesses and densities and then be studied clinically to ascertain if fixation failures diminished with IDTS in the upper extremity where plate loosening is often seen in the humerus for example. This study begins with representative bone from the acetabular region (Fig1) and future work is planned to examine other bony regions.

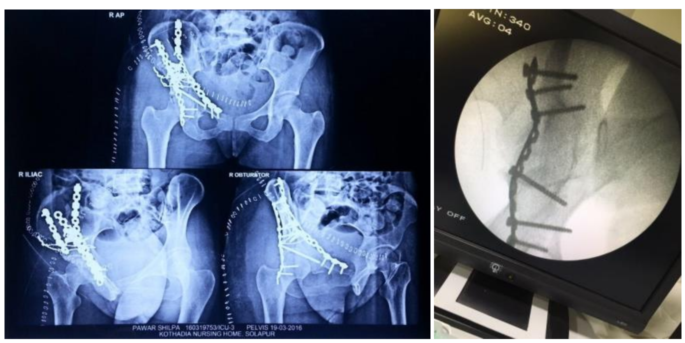


Fig1. Images of a both column acetabular fracture including pre- and post-fixation (Courtesy: Dept. of Orthopaedics, Padmashree Dr. D.Y. Patil Hospital, Mumbai, India)

With the advance of 3D images **[25, 26, 27]** and computer aided design **[28]**, 3D drilling simulation has become possible. Finite element analysis (FEA) is a major key in employed in developing new surgical techniques. Modelling the bone drilling process using FEA is very useful for the validation of experimental and analytical results **[29]**. Gok et al. **[30]** developed a new drilling protocol to prevent osteonecrosis secondary to bone drilling.

In this work, an elastic-plastic three-dimensional finite element model was used for drilling simulation. Figure 2 shows the geometrical configuration of the model. The features of drill and bone are described in figure 3. The model contains 3D objects which were modelled using the geometry features of software DEFORM-3D:

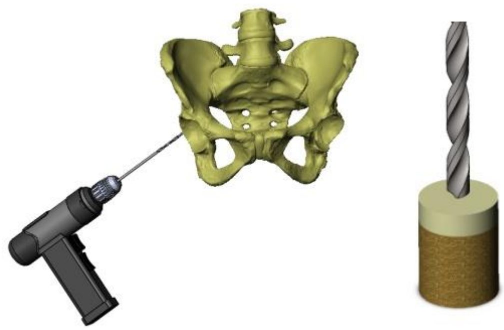


Fig2. Drilling of acetabular bone

Fig3. Features of drill

a) The drill is considered as non deformable object. The diameter, speed, rotation, point angle, helix angle are taken 2.8 *mm*, 0.1*mm/s* and 600 *rev/min,* 135°, 28° respectively.

b) A layer of cortical bone is an elastic-plastic model with diameter (10 *mm*) and thickness (6*mm*).

c) A layer of trabecular bone, this object is modelled as plastic taking into account of the density of material with diameter (10 *mm*) and thickness (10 *mm*).

The properties of the material play a significant role in the accuracy for solving the problem. This is necessary in order to avoid unreliable results. Material drill defined as absolutely rigid (no distortion). The material of cortical layer bone is given as an elastic-plastic isotropic. The material of trabecular layer is given as porous isotropic.

	Drill bit	**Cortical bone**	**Trabecular bone**

Thermal conductivity (W/mk)	36	0.56	0.05
Density (kg/ m^2)	7860	1640	640
Heat	5.0	2.86	2.0
Young's modulus (MPa)	206754	16700	1000
Poisson's ratio	0.3	0.3	0.2
Stress (MPa)	520	105	19

Tab2. Mechanical properties

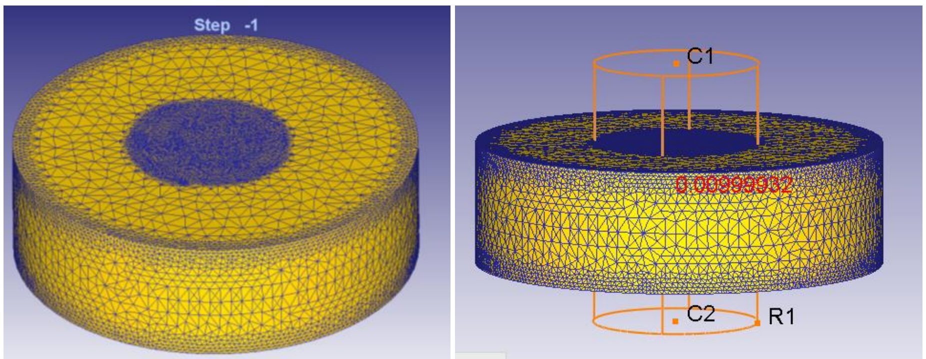


Fig 4. Interaction phenomenon of drill bit to bone

Figure 4 illustrates the contact geometry between drill bit and bone reflected in the simulated model. The region of interest is located in vicinity of hole. In this model, the two objects are constructed using cylindrical shape (tetrahedron mesh).

The software FEA module is responsible for the modeling drilling process. The strain is expressed as the ratio of total deformation to the initial dimension of the material body in which the forces are being applied. Thus, we have:

$\varepsilon = \dfrac{l - l_0}{l_0}$, with l is final length of the fiber and l_0 is the initial length of fiber.

The three-dimensional deformation is represented by the following equation:

$$\begin{cases} \delta x = \delta x_0 + \dfrac{\delta u}{\delta x}\delta x_0 + \dfrac{\delta u}{\delta y}\delta y_0 + \dfrac{\delta u}{\delta z}\delta z_0 \\ \delta_y = \delta y_0 + \dfrac{\delta v}{\delta x}\delta x_0 + \dfrac{\delta v}{\delta y}\delta y_0 + \dfrac{\delta v}{\delta z}\delta z_0 \\ \delta_z = \delta z_0 + \dfrac{\delta w}{\delta x}\delta x_0 + \dfrac{\delta w}{\delta y}\delta y_0 + \dfrac{\delta w}{\delta z}\delta z_0 \end{cases}$$ Where u, v, w are the displacement vectors and x, y and z are spatial coordinates.

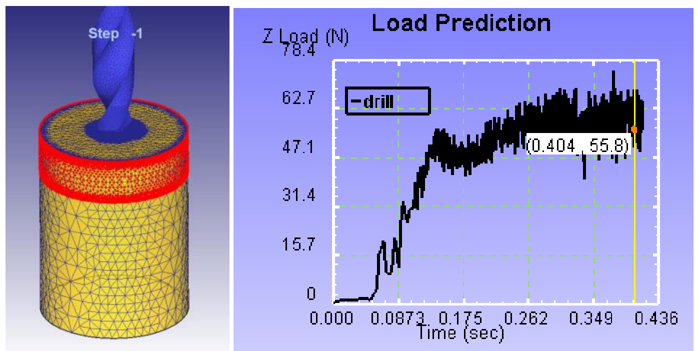


Fig 5. Strain and load

3. Mathematical Representations

The tensor characterizes the compression (expansion) and the change in shape in each point of the body under deformation. That is written in the matrix form:

$$\varepsilon_{ij} = \begin{pmatrix} \varepsilon_x & \dfrac{\gamma_{yx}}{2} & \dfrac{\gamma_{zy}}{2} \\ \dfrac{\gamma_{xy}}{2} & \varepsilon_y & \dfrac{\gamma_{zy}}{2} \\ \dfrac{\gamma_{xy}}{2} & \dfrac{\gamma_{yz}}{2} & \varepsilon_z \end{pmatrix}$$, where γ_{ij} are the components of deformation

The FEA is conducted by modelling the heat which is based on this boundary condition to be properly formulated **[31]**:

$T(x, y, z, t)_{t=0} = T_0$, where T_0 is an initial (ambient temperature) which is equal approximately to 20.0 °C.

The friction and plastic deformation generate heat and elevate the bone temperature. The mathematical formulations of thermo mechanical modeling are presented with this equation:

$pc\frac{\partial T}{\partial t} = k(\frac{\partial^2 T}{\partial x^2} + \frac{\partial^2 T}{\partial y^2} + \frac{\partial^2 T}{\partial z^2}) + G$, where p is the density, c is the specific heat, k is the heat conductivity, T is the temperature, t time, and G is heat generation rate

The heat generation rate G consists of the heating rate by the friction between tool and bone ($\dot{q}_f$) and heating rate from irreversible plastic deformation inside the bone ($\dot{q}_p$)

$$G = \dot{q}_f + \dot{q}_p$$

We know that the frictional force F_f is proportional to normal force:

$$F_f = \mu F_n$$

Frictional behavior and contact interaction between the tool and work piece in friction drilling are complicated **[32],** in this study, a constant coefficient of friction using Coulomb's friction law which equal to 0.25.

At the local contact point, the velocity of drill bit:

$V = 2\pi RN$, where R is the radius of drill bit and N is rotational speed

$$\dot{q}_f = 2\pi RN\mu F_n$$

And

$$\dot{q}_p = \eta\sigma\dot{\varepsilon}^{pl}$$

Where, η is the inelastic heat fraction and σ is the effective stress

4. Results and discussion

As shown in figure 6, residual stress and plastic strain exist on the surface of bone, the deformation change greatly during drilling in the x-direction. The plastic strain is bigger in the figure 6.10 compared to the figure 6.2. In this simulation, the residual stress exciting in the trabecular bone are pressure stress at around 19 MPa. Hence, the residual stress reduces with speed of drill. It is normal with rising of cutting speed, the cutting force goes down and plastic strain of bone becomes less. From the figure 6.5 the stress begins much larger (105 MPa) in the position nearby of cutter at cortical layer.

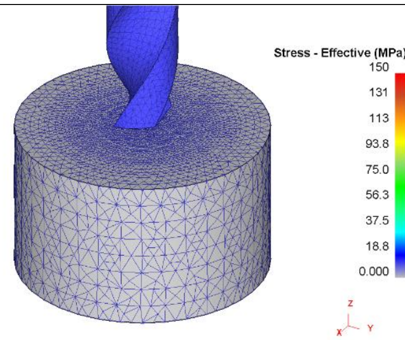


Fig 6.1

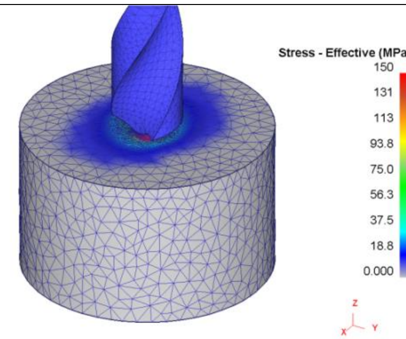


Fig 6.2

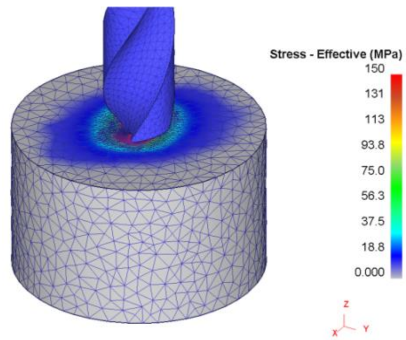


Fig 6.3

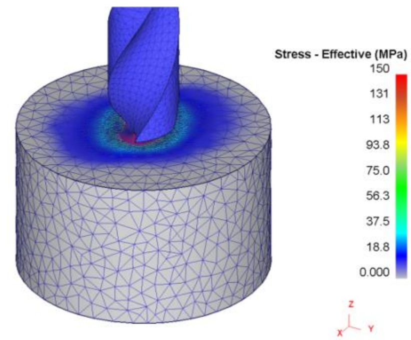


Fig 6.4

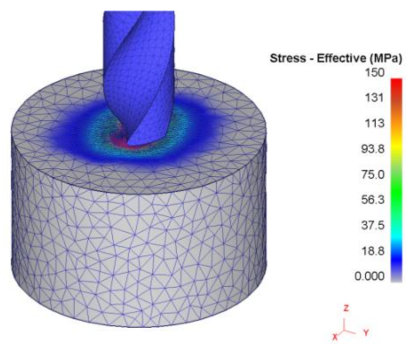
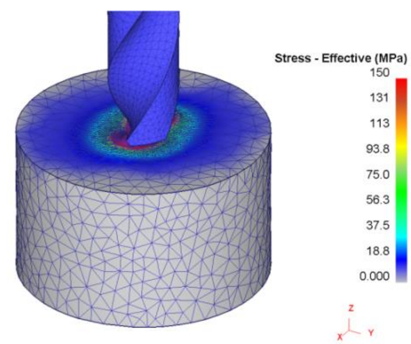


Fig 6.5



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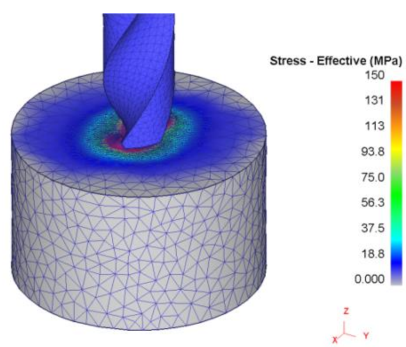


Fig 6.7

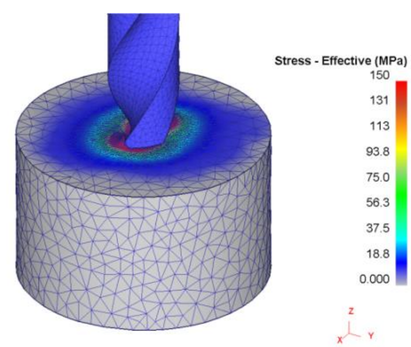
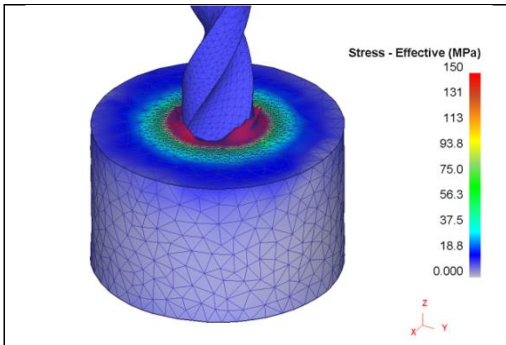


Fig 6.8



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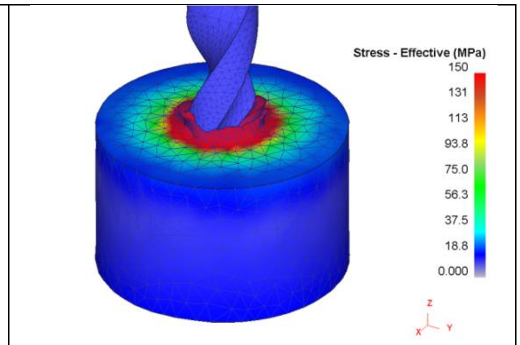


Fig 6.10

Fig 6. Stress during drilling

The results of the calculation of the variation of the temperature are summarized in the figure 7. The analysis of the results revealed a diversified distribution of temperature generated in acetabular bone. In this picture, with drill bit, the temperature begins to rise significantly (figure 7.1- figure 7.7), then goes down slowly (figure 7.9 and figure 7.10 specially (i.e. trabecular layer). Obviously, the temperature peaked in figure 7.7 which equal to 85° in cortical bone. Additionally, the increase in temperature is related to the increase in the speed of drill bit.

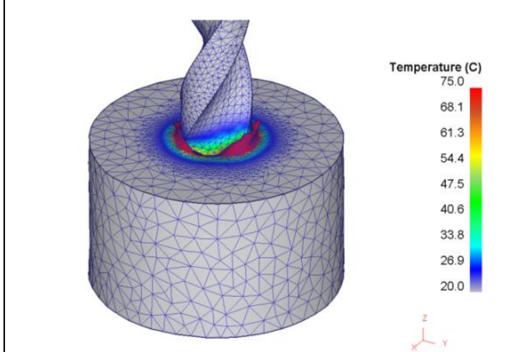


Fig 7.1

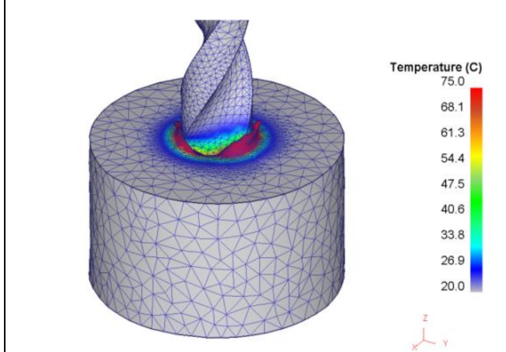


Fig 7.2

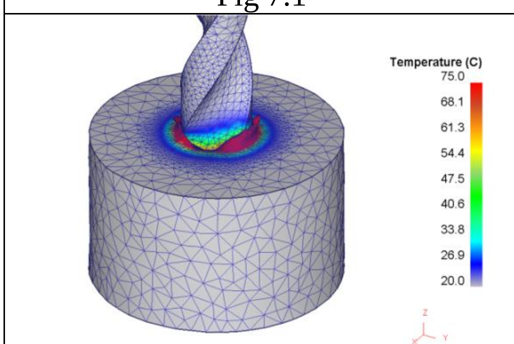


Fig 7.3

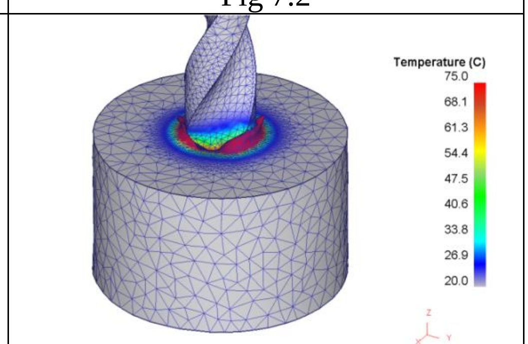


Fig 7.4

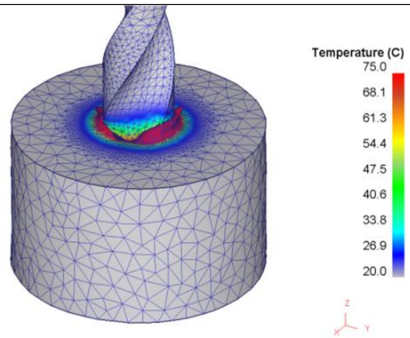
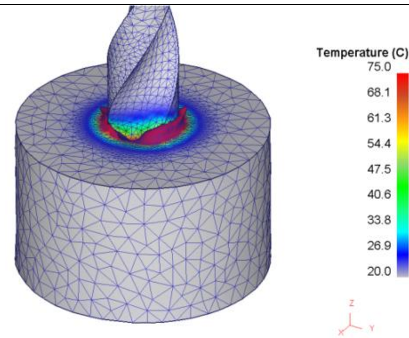


Fig 7.5



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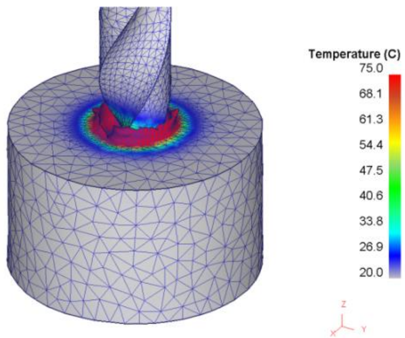

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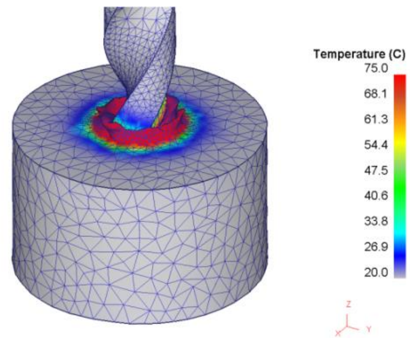

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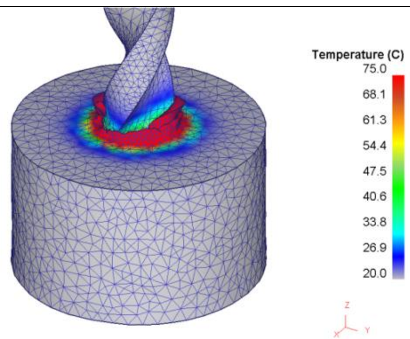

Fig 7.9

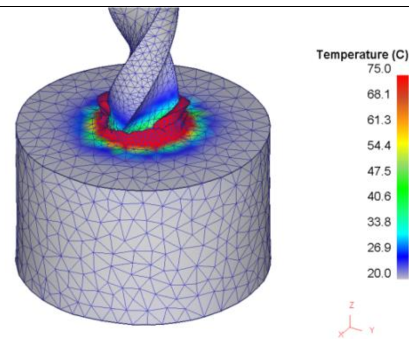

Fig 7.10

Fig 7. The temperature distribution in the acetabular during the process of drilling using a drill.

Our simulation enables the monitoring of bone temperature during a simulated bone drilling procedure. Our FEA studied a common drill diameter of 2.8mm and monitored the results bone stress in MPa and bone temperature. Our results show the stress and temperature generated when drilling cortical at a constant rotation of 600 rev/min. Drilling 3mm depth, feeding 1 mm/sec produced axial force acting on the drill bit from the bones of about 50-80 Newton (N) and a temperature of 75°C. When the feeding speed was reduced to 0.1 mm/sec, temperature reduced to 39°C, even though the axial force acting on the drill bit stayed constant unaffected and the rotation remained constant at 600 rev/min. Our simulated drilling procedure with finite element analysis indicates that the depth of drilling and the drill speed both have a significant effect on the temperature during drilling procedures. FEM models and simulators as developed in this research study would help to provide insights into the force and temperature dynamics involved with bone drilling and enable real-time monitoring of temperature during drilling training simulations for surgeons. Knowing the ideal drill speed, drill depth and rate of drill advance would be used in conjunction with irrigating or frequent drill bit changes by the surgeon to minimize bone necrosis and associated screw loosening. Reduced ORIF construct failures would in theory lead to better patient outcomes, less revision surgeries, less anesthetic complications and in theory less overall health care expenditures. This can be useful for pre-operative planning to identify the chances of potential overheating leading to risks. Further research could be done to predict how many drill bits would be required for a certain fracture, depending upon how many screws may be used which can be calculated pre-operatively. FEM provides information that would not otherwise be possible to measure, since it is very challenging and currently impossible to directly measure bone temperature in-vivo on the actual patient due to ethical constraints.

5. Conclusions and future works

The authors acknowledge that this simulation study was performed while showing controlling temperature in each layer of the bone. In our simulated model, the factors causing most bone necrosis (forces applied by surgeons and speed of drill) are more readily held constant, but would likely vary in the operating theater which could alter the rate at which bone temperatures reach critical levels.

Drilling of bone for placement of an implant is becoming more common in orthopaedic surgery. Avoidance of bone necrosis from the drilling process is thought to lessen implant loosening and lessen procedure failures. Our FEM simulation provides useful information such as that reduction of drill feeding speed during orthopaedic surgery can lead to a reduction in temperature. This suggests that reducing the drill feeding speed could be a useful and effective way to reduce temperature, which could help to reduce the risk of patients developing necrosis during bone drilling. The FEM simulation also provides a method of checking when overheating is likely to occur in-vivo enabling the surgeon to subsequently adjust the drill feeding speed. This can help to reduce patient complications, lower cost of patient treatment and improve recovery time after surgery by lessening the need for revision surgery.

This review analyzes the most influential factors on strain and temperature during drilling, information previously unreported in the literature. There are some factors which are still unknown, or insufficiently examined. From the above review the following aspects may be useful in future work:

- Allow a better comparison between experimental and simulated drilling.
- Provide an overview of the temperature during drilling for all human bones, taking into account the bone density variables such as: age, sex, location, diaphyseal, metaphyseal, upper or lower extremity and weight bearing or non-weight bearing bone.
- Drilling tools may be developed which can provide visual feed back to the surgeon regarding the pressure applied to bone and the RPMs of the drill bit so that the surgeon may adjust accordingly to avoid excess heat generation.
- Determine how often the bone should be irrigated and using what exact volume of water is required to cool the drill bit to starting temperature.

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