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

Short title : **3D Drilling simulation**

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Authors Contributions

- *Mohamed Mediouni* make substantial contributions to conception and design.
- *Daniel R. Schlatterer* and *Tobias von Bergen* participate in drafting the article or revising it critically for important intellectual content.
- *Amal Khoury* explains the complexity of acetabular fracture.
- *Manit Arora* participates in the interpretation of data.
- *Amit Dhond* and *Sunil H. Shetty* provide medical images and information about a drill bit.
- *Neil Vaughan* gives final approval of the version to be submitted.
- *Alexander Volosnikov* verify the simulation of Finite Element Analysis.

Abstract

The drilling bone may potentially cause excessive frictional heat, which can lead to local bone necrosis. This heat generation and local necrosis has been suggested to contribute to the resorption of bone around the placed screws, ending in loss of screw purchase in the bone and inadvertent loosening and/or the bone-implant construct. In vivo studies on this subject have inherent obstacles not the least of which is controlling the variables and real time bone temperature data acquisition. Theoretical models can be generated using computer software and the inclusion of known constants for the mechanical properties of metal and bone. These known Data points for the variables (drill bit and bone) enables finite element analysis of various bone drilling scenarios. An elastic-plastic three-dimensional (3D) acetabular bone mode was developed and finite element model analysis (FEM) was applied 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 reduction of the feeding speed leads to a reduction in bone temperature. Our data suggests that reducing the feeding speed regardless of RPMs and pressure applied could be a simple useful and effective way to reduce drilling temperatures. This study is the first step in helping any surgeon who drills bone and places screws to better understand the ideal pressure to apply and drill speed to employ and advance rate to avoid osteonecrosis.

Keywords: Drilling, osteonecrosis, Finite Element Analysis (FEA)

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 **[26]**. The ancient Egyptian civilization practiced dentistry **[22, 7]**, in a tomb near the river Nile two teeth were found, ingeniously connected by a gold wire passed through holes in both teeth **[8]**. 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 **[4]**. Drilling has become common-place 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. The complexity of drilling 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 **[20]**. The frictional heat of drilling may cause thermal necrosis of the bone. Faced with this clinical problem, many researchers **[23, 19, 24]** 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 **[17]**. There are many studies in the literature for drilling analysis most with contradictions and a general lack of consistency that will mentioned below.

Anderson et al. **[1]** published the first thermal change studies to forge teeth. Mathews et al. **[15]** showed that there is no change of the temperature in vitro or vivo. Hillery et al. **[10]** 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 which are: non-influenced parameters (drill design, drill point angle, drill diameter, and drill material) and influenced parameters (speed of drill, feed rate, cutting forces, and drilling depth). 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 **[25, 21]**. According to Augustin et al. **[2],** the temperatures above critical were recorded using 4.5 mm drill with higher drill speeds are 188, 462, 1140 and 1820 rpm. Others focused on low speed drilling (up to 3000 rpm) **[25]**. 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 **[7]**. 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 1) **[27]**. 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. **[28]** 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. **[3]** 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 **[11, 6]**.

Aim of this study was to overcome the drawback of others works and provide a simulation based on 3D Finite Element Analysis. In this article a mathematical model will be explained in order to provide exact values of temperatures during drilling.

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 **[13, 16, 9]**. The pelvis is an bone 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 and future work is planned to examine other bony regions.

In this work, an elastic-plastic three-dimensional finite element model was used for drilling simulation. Figure 1 shows the geometrical configuration of the model. The model contains 3D objects which were modelled using the geometry features of software DEFORM-3D of Scientific Forming Technologies Corporation (SFTC):

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 trabucular 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 **[17]**. 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 *(means a material having identical value of a property in all direction)*. The material of trabecular layer is given as porous isotropic. 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 *(measure of deformation representing the displacement between particles in the body)* 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:

$\varepsilon = \frac{l - l_0}{l_0}$, 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:

$$\begin{cases} \delta x = \delta x_0 + \frac{\delta u}{\delta x}\delta x_0 + \frac{\delta u}{\delta y}\delta y_0 + \frac{\delta u}{\delta z}\delta z_0 \\ \delta y = \delta y_0 + \frac{\delta v}{\delta x}\delta x_0 + \frac{\delta v}{\delta y}\delta y_0 + \frac{\delta v}{\delta z}\delta z_0 \\ \delta z = \delta z_0 + \frac{\delta w}{\delta x}\delta x_0 + \frac{\delta w}{\delta y}\delta y_0 + \frac{\delta w}{\delta z}\delta z_0 \end{cases}$$

Where u, v, w are the displacement vectors and x, y and z are spatial coordinates. 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:

$\varepsilon_{ij} = \begin{pmatrix} \varepsilon_x & \frac{\gamma_{yx}}{2} & \frac{\gamma_{zy}}{2} \\ \frac{\gamma_{xy}}{2} & \varepsilon_y & \frac{\gamma_{zy}}{2} \\ \frac{\gamma_{xy}}{2} & \frac{\gamma_{yz}}{2} & \varepsilon_z \end{pmatrix}$, 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 **[12]**:

$T(x, y, z, t)_{t=0} = T_0$, 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 is presented with this equation:

$pc\frac{\delta T}{\delta t} = k\left(\frac{\delta^2 T}{\delta x^2} + \frac{\delta^2 T}{\delta y^2} + \frac{\delta^2 T}{\delta z^2}\right) + G$, where p is the density (mass per unit volume), c is the specific heat, k is the heat conductivity, T is the temperature, t is 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 **[18],** 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

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

Where, ϑ is the inelastic heat fraction and σ is the effective stress (a force that keeps a collection of particles rigid).

The penetration of the tool allows the erasure of a part of an object which composed of several elements. This is due to the detection of elements in the contact region on a high accuracy given par the user. A detailed view of the distorted mesh shown at the figure 2.

4. Results and discussion

As shown in figure 3, 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 3.10 compared to the figure 3.2 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 3.5 the stress begins much larger (105 MPa) in the position nearby of cutter at cortical layer.

The results of the calculation of the variation of the temperature are summarized in the figure 4. 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 4.1- figure 4.5). 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.

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 feed rate was reduced to 0.1 mm/sec and the speed rotation at 600 rev/min, temperature augmented to 85°C. When we are reducing the speed at 300 rev/min, the temperature reduced to 39 °C.

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 change 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, and 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.

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 the 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.

Conflict of interest statement

No

Ethical Review Committee Statement

Not available

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	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

Table1 . Mechanical properties

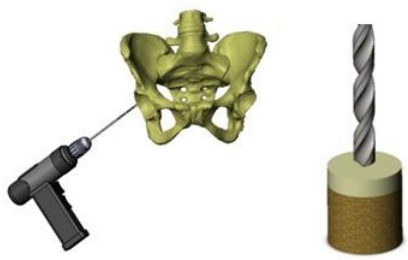


Figure 1. Drilling of acetabular bone

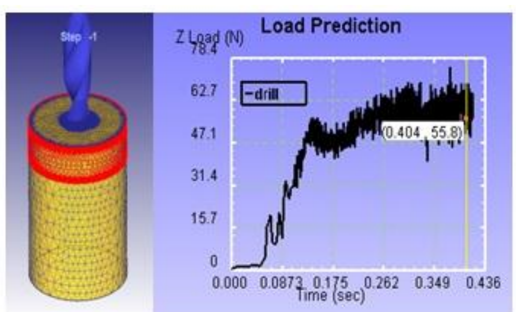


Figure 2. Strain and load

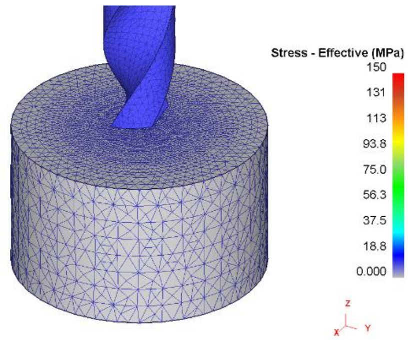
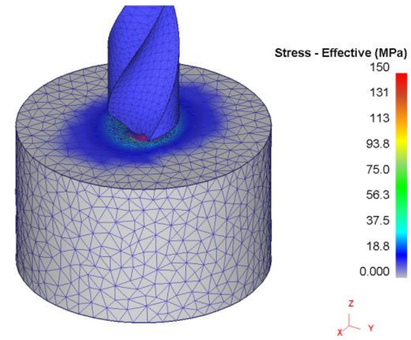


Fig 3.1



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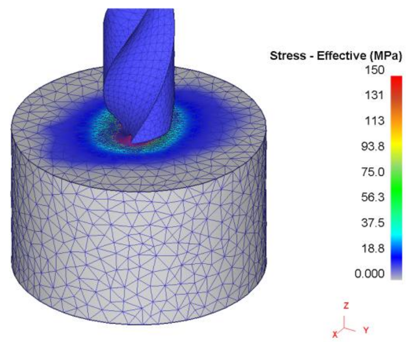

Fig 3.3

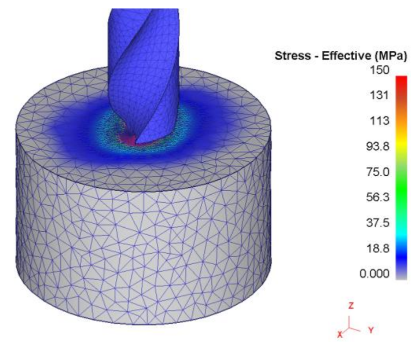

Fig 3.4

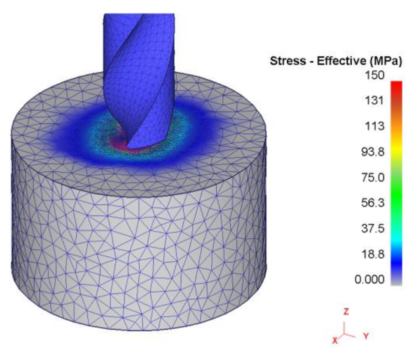

Fig 3.5

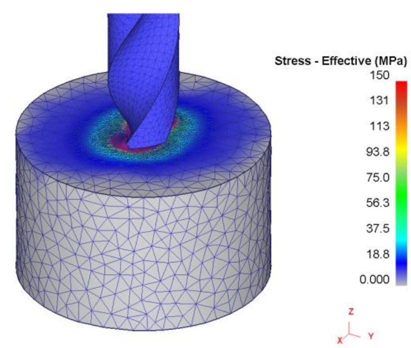

Fig 3.6

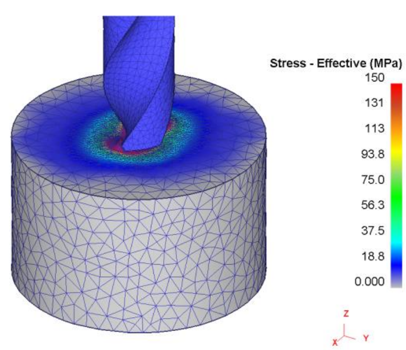

Fig 3.7

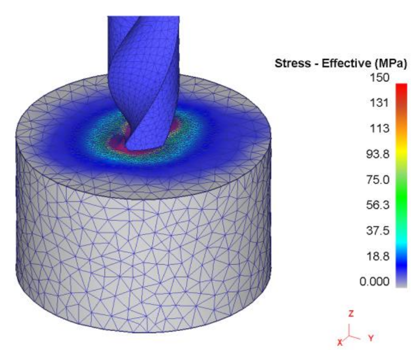

Fig 3.8

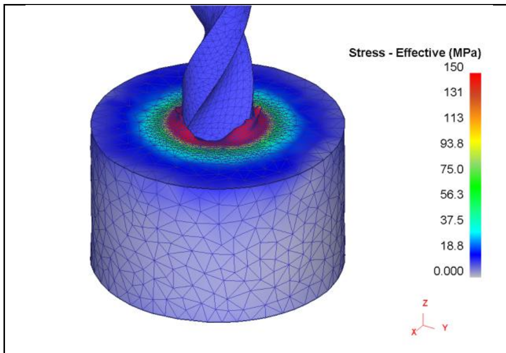

Fig 3.9

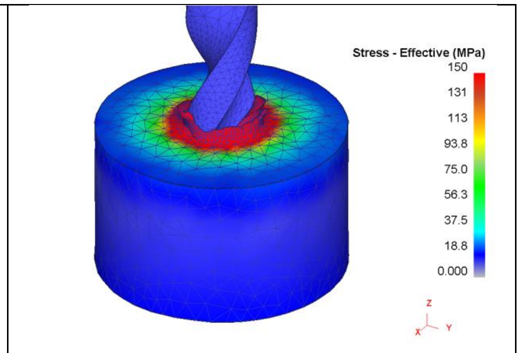

Fig 3.10

Figure 3. Stress during drilling, in (3.1- 3.2) stress is low with blue color and his value between [0-20] MPA. In figure (3.3-3.8), the stress begins to spread on all surface of bone which it is correct from a physical point of view. On the other side, the stress increase (figure 3.9-3.10) in the highlighted region of hole created during drilling. This is logic because the layer of cortical bone is more dense compared to trabecular bone. The red color indicates that the stress reach the interval [100-150] MPA.

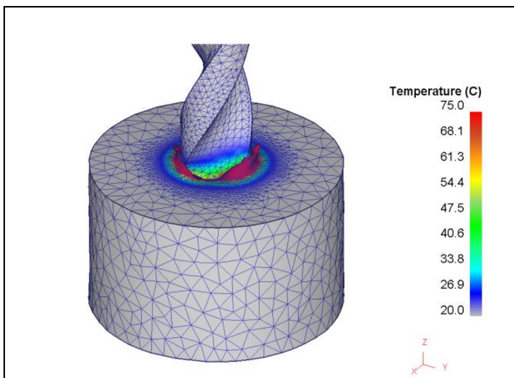

Fig 4.1

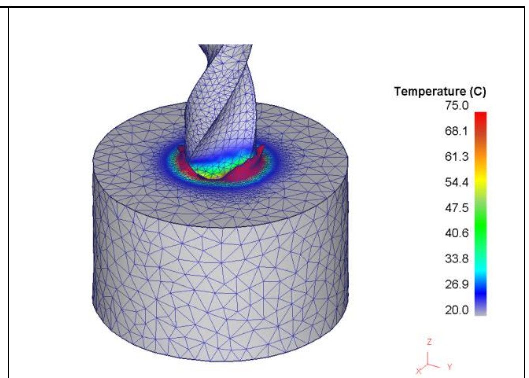

Fig 4.2

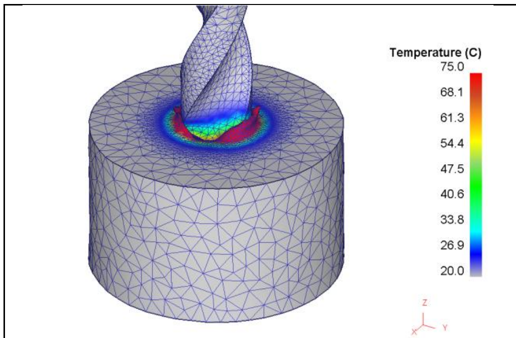

Fig 4.3

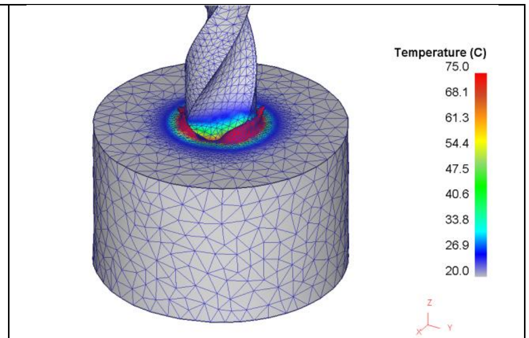

Fig 4.4

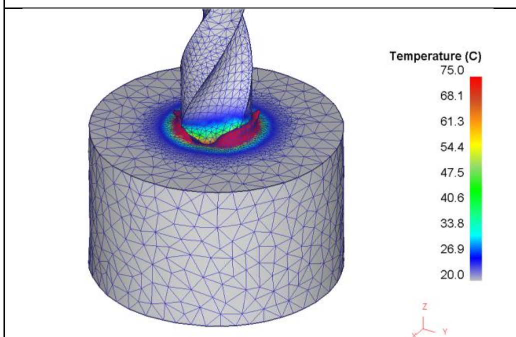

Fig 4.5

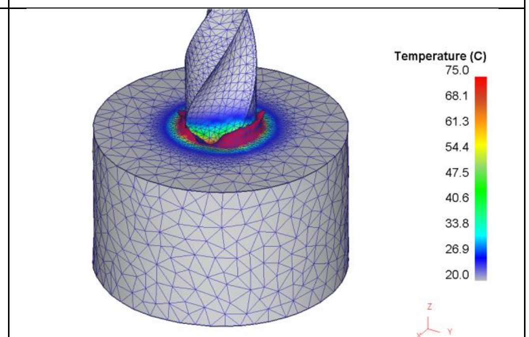

Fig 4.6

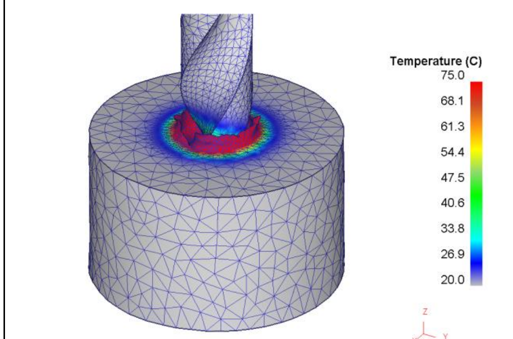

Fig 4.7

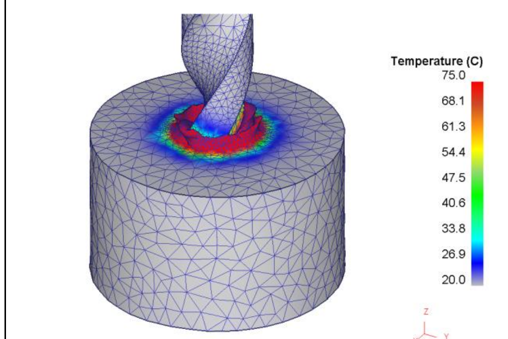

Fig 4.8

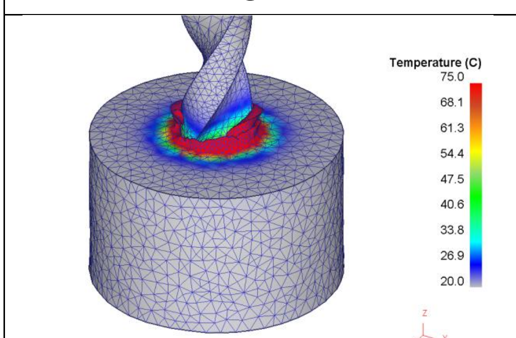

Fig 4.9

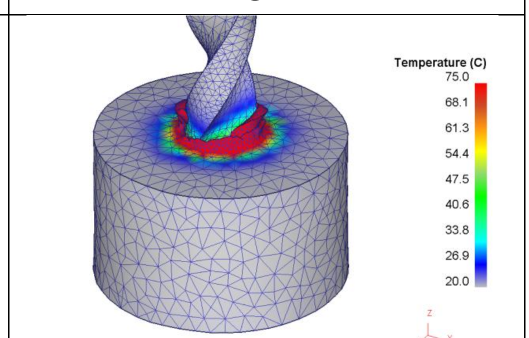

Fig 4.10

Figure 4. The temperature distribution in the acetabular during the process of drilling using a drill.