The thermal properties of bone and the effects of surgical intervention

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KEYWORDS
Bone; Drilling; Thermal effects

Summary
Much is written about the mechanical properties of bone; we are aware of how bone responds to different modes and degrees of stress and can use this knowledge to influence the mechanical environment around bones and joints for therapeutic effect. The thermal properties of bone are less well documented, yet all orthopaedic interventions produce significant thermal effects. This paper explains basic material thermodynamics and how this relates to bone and impacts on an orthopaedic surgeon’s practice.

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Introduction
Bone is a complex biological tissue, with organic and mineral phases. The interaction of the different phases of bone account for its unique, complex mechanical properties. Bone is equally complex from a thermological perspective. This behaviour is difficult to study, sensitive to testing conditions, specimen preparation and is anisotropic.1

Heat is thermal energy transferred between a system and its surroundings. The transfer can take place by three different mechanisms:

- Conduction—Thermal energy is transferred through the substance of the system.
- Convection—Thermal energy is transferred by relative motion of components of the system.
- Radiation—Thermal energy is transferred directly between separated parts of the system by electromagnetic radiation.

In physics, materials can be defined thermologically by certain parameters:

Specific heat: The energy required to raise the temperature of a system by 1 °C. More specifically this is defined per unit volume or unit mass. It is effectively a measure of how easily a material "heats up".

Thermal conductivity: The thermal energy transfer in steady state per unit area and unit time between two infinite parallel planes, the distance and temperature gradient between which are 1 unit length and 1 °C, respectively. It represents the ability of a material to transport heat.

Various researchers have calculated these parameters for bone2 (Table 1).

The biological effects of excessive temperature on bone
The importance of heat generation during bone drilling has long been recognised. In his theory and practice of...
Table 1  The thermophysical properties of bone.

<table>
<thead>
<tr>
<th>Animal species</th>
<th>Specific heat (cal/g °C)</th>
<th>Thermal conductivity (cal/cm s °C)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Fresh bone</td>
<td>Dry bone</td>
</tr>
<tr>
<td>Man</td>
<td>0.30 ± 0.01</td>
<td>0.30 ± 0.01</td>
</tr>
<tr>
<td>Elephant</td>
<td>0.30</td>
<td>0.28 ± 0.02</td>
</tr>
<tr>
<td>Ox</td>
<td>0.27 ± 0.01</td>
<td>0.28 ± 0.01</td>
</tr>
<tr>
<td>Dog</td>
<td>0.30 ± 0.01</td>
<td>0.26 ± 0.01</td>
</tr>
</tbody>
</table>

At the histological level, Erikson and Albrektsson\textsuperscript{11} noted cortical necrosis in living rabbit bone heated to 47 °C for 1 min. Cortical necrosis and a delay in the healing of surgical defects was reported for dog femora that were heated to between 43.3 and 68.7 °C by ultrasound after surgery.\textsuperscript{12} Rouiller and Majno\textsuperscript{13} described necrosis of osteocytes in the long bones of rabbits when they were exposed to a temperature of 55 °C for 1 min.

Bonfield and Li\textsuperscript{14} reported that irreversible changes to the mechanical properties of bone occur when dog femora are heated to 50 °C in vivo. They attributed this to a reorientation of collagen molecules secondary to a weakening of bonds between collagen and hydroxyapatite.

The effect of heat trauma has been most extensively studied in relation to skin. Moritz and Henriques\textsuperscript{15} concluded that the threshold temperature for epidermal necrosis at an exposure time of 30 s was 52–55 °C. Dermal blood vessels were more sensitive to thermal trauma than epithelial cells, with a threshold temperature for necrosis of 41–45 °C. Leach et al.\textsuperscript{16} concluded that temperatures of 47 °C appear to kill all cell types. Between 42 and 47 °C death of all cells types can be produced, provided the time of heating is long enough.

From the above discussion, it can be seen that the conditions for thermal bone tissue necrosis may be variable. Therefore, there is no agreement in the literature as to the threshold for thermal necrosis in bone and the ideal conditions to study this. A review of the literature reveals that many authors\textsuperscript{17–19} have taken 50 °C as the point at which bone necrosis occurs. This conclusion is drawn from the work of Lundskog\textsuperscript{2} and an average of the temperatures reported in the literature as being capable of causing necrosis.

Heat generation during drilling

The drilling process

The heat generated during drilling of bone comes from, firstly, the drilling process. Shear of the surface layer of a material by a drill bit breaks intermolecular bonds releasing energy. Secondly, friction from the non-cutting surfaces of a twist drill, such as the flank, flutes and shaft are another source of heat. The heat generated is partially dissipated by the presence of blood and tissue fluid, and part of the heat being carried away by the chips formed. However, bone is a poor conductor of heat and the temperature rise can be significant.\textsuperscript{20}
The temperature developed within a section of bone tissue depends on specific heat capacity (how readily a material heats up) and on the thermal conductivity (how readily a material transfers heat from a source).

When a heat source is applied to a specimen, a thermal gradient exists from the hotter heat source to the cooler specimen. Heat is transferred down this gradient. The higher the gradient the greater the rate of transfer. Heat continues to transfer as long as the gradient is maintained. The ultimate temperature a specimen develops depends on the balance between heat gain and heat loss. When these two phenomena are equal, the specimen is in a state of equilibrium. In the equilibrium state the temperature reached by a specimen is a function of the exposure temperature and thermal conductivity. Temperature equilibrium is, however, not normally reached when biological tissues are exposed to heat in a clinical situation such as drilling, as the exposure times are usually too short. The temperature reached in the non-equilibrium state of short exposures is a function of the exposure temperature and time, the specific heat capacity being constant.  

Different time-temperature relationships can produce similar effects in a specimen. Lundskog was able to measure the density of isotherms (lines connecting points with the same temperature) around a point heat source applied to a bone sample at various temperatures and exposures. He showed that the width of the 50\% isotherm was approximately the same for such different thermal exposures as 90\% for 10 s, 80\% for 20 s and 75\% for 30 s. As the exposure time shortens, so the distribution of isotherms becomes more compact. The reason for this is that heat requires a finite time to penetrate the tissue by conduction. The shorter the exposure time the more localised the heating process.

The drilling process generates heat within the drill and in the bone. The rate of heat generation depends on various parameters of the cutting process, and these in turn influence the temperature of the tool, bone and bone chips, depending on their relative thermal properties.

**Drill temperature**

The general formula correlating tool temperature $T_i$ is given by

$$T_i = C_0K_vv^2A^nW^{2n}h^{12n}$$

where $K_v$ is the specific cutting energy, $v$ the cutting velocity, $A$ the chip cross-sectional area, $W$ the thermal conductivity of work material, and $h$ the thermal capacity (density $\times$ specific heat) of work material.

$C_0$ and $n$ are constants and can be obtained experimentally. It is important to observe from the equation that the temperature $T_i$ is directly proportional to the specific cutting energy $K_v$, which in turn is dependent on the dynamic shear strength of the material, since cutting is a dynamic shear failure process. The specific cutting energy increases with increase in the dynamic shear strength for a brittle material. As bone is a brittle material this can be assumed to apply.

**Bone temperature**

The thermal capacity and conductivity of bone constitutes the other important factors affecting the temperature rise. These factors have an inverse relationship with the temperature. The lower the values, the higher the temperatures generated at the drill-bone interface. The thermal conductivity of the tool material has little effect on the temperature. When drilling metals the metal chips carry away nearly 85\% or more of the heat generated. Unlike drilling metals, the temperature of the bone will rise more during the process due to its poor thermal capacity and conductivity, chips carrying away a smaller percentage of the heat. It should be pointed out that compared to the situation with metals, the total heat generated under identical heating conditions will be much less during bone drilling due to its lower $K$ value. This is an important consideration for bone drill design.  

**Factors effecting the temperatures generated when drilling bone**

**The effect of drill force**

The literature is not clear regarding the influence of drill force on bone temperature. Some authors have found that the temperature rise or specific energy (energy required to drill a specific mass of material) decreases with force or feed rate (rate of drill advancement per unit time or per revolution of drill). Mathews and Hirsch, for example, applied loads of up to 118 N while drilling at various speeds in human cadaveric cortical bone and found that low temperatures are associated with high forces. They suggested that a high force causes the drill to cut more rapidly. Hence the drilling time is shorter and the temperature rise is smaller. Wiggins and Wiggins and Malkin found that specific energy is lower at higher feed rates; temperatures were not measured, but they would likely have been lower because less energy was consumed. This was confirmed in a later study by Krause et al. who drilled in bovine femoral cortical bone and showed that the temperature decreases as the feed rate increases.

Other studies have found that temperature increases with force. Eichler and Berg, for example drilled in cortical sections of bovine femurs and recorded temperatures that increased with force. Hensche and Peyton investigated drilling in teeth and found that the temperature rises with force. However, in these investigations, the forces used were less than 30 N, much lower than the maximum force of 118 N applied by Mathews and Hirsch.  

In contrast to these results, Krause found that the specific cutting energy initially increases with feed rate and then decreases. His study, however, pertains to orthogonal bone cutting (the tool moves in a straight path parallel to the surface of the specimen) and not to drilling. Krause hypothesised that the decrease in specific cutting energy at high velocities is the result of changes in the material’s fracture properties. Abouziga and James studied the temperature rise in bovine femurs using a surgical drill operating at 49,000 rpm with forces ranging from 1.5–9 N. They demonstrated that
Table 2  Temperatures recorded in bone during drilling.

<table>
<thead>
<tr>
<th>Study</th>
<th>Max temp (°C)</th>
<th>Distance from drill periphery</th>
<th>Bone type</th>
<th>Type of study</th>
<th>Cooling</th>
<th>Free running drill speed rpm</th>
<th>Force or feed rate</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thompson</td>
<td>65.5</td>
<td>2.5</td>
<td>Dog mandible</td>
<td>In vivo</td>
<td>No</td>
<td>125–2000</td>
<td>Not indicated</td>
</tr>
<tr>
<td>Pallan</td>
<td>65</td>
<td>2</td>
<td>Dog mandible</td>
<td>In vivo</td>
<td>No</td>
<td>125–2000</td>
<td>Not indicated</td>
</tr>
<tr>
<td>Rafel</td>
<td>23.5</td>
<td>3</td>
<td>Human mandible</td>
<td>In vitro</td>
<td>No</td>
<td>35,000</td>
<td>Not indicated</td>
</tr>
<tr>
<td>Mathews and Hirsch</td>
<td>140</td>
<td>0.5</td>
<td>Human femur</td>
<td>In vitro</td>
<td>No</td>
<td>345,885</td>
<td></td>
</tr>
<tr>
<td>Eichler and Berg</td>
<td>&lt;50</td>
<td>0.5</td>
<td>Human femur</td>
<td>In vitro</td>
<td>Yes</td>
<td>2900</td>
<td>20, 59N</td>
</tr>
<tr>
<td>Jacobs and Ray</td>
<td>95</td>
<td>0.5</td>
<td>Human femur</td>
<td>In vitro</td>
<td>No</td>
<td>700</td>
<td>10, 20 and 30N</td>
</tr>
<tr>
<td>Tetsch</td>
<td>38</td>
<td>2</td>
<td>Rat radius</td>
<td>In vivo</td>
<td>Yes</td>
<td>2500</td>
<td>Not indicated</td>
</tr>
<tr>
<td>Tetsch</td>
<td>300</td>
<td>1</td>
<td>Cat mandible</td>
<td>In vivo</td>
<td>No</td>
<td>20,000</td>
<td>Not indicated</td>
</tr>
<tr>
<td>Lavelle and Wedgewood</td>
<td>89</td>
<td>0.5</td>
<td>Human femur</td>
<td>In vitro</td>
<td>No</td>
<td>350</td>
<td>59N</td>
</tr>
<tr>
<td>Lavelle and Wedgewood</td>
<td>74</td>
<td>0.5</td>
<td>Human femur</td>
<td>In vitro</td>
<td>Ext irrigation</td>
<td>350</td>
<td>59N</td>
</tr>
<tr>
<td>Lavelle and Wedgewood</td>
<td>50</td>
<td>0.5</td>
<td>Human femur</td>
<td>In vitro</td>
<td>Int irrigation</td>
<td>350</td>
<td>59N</td>
</tr>
<tr>
<td>Pal and Saha</td>
<td>&gt;60</td>
<td>1</td>
<td>Bovine long bones</td>
<td>In vitro</td>
<td>No</td>
<td>65–2800</td>
<td>0.128 mm/rev</td>
</tr>
<tr>
<td>Krause et al.</td>
<td>55</td>
<td>Near</td>
<td>Bovine femur</td>
<td>In vitro</td>
<td>No</td>
<td>20,000</td>
<td>1.8–6.36 mm/s</td>
</tr>
<tr>
<td>Krause et al.</td>
<td>130</td>
<td>Near</td>
<td>Bovine femur</td>
<td>In vitro</td>
<td>No</td>
<td>100,000</td>
<td>1.8–6.36 mm/s</td>
</tr>
<tr>
<td>Eriksson et al.</td>
<td>41</td>
<td>0.5</td>
<td>Rabbit femur</td>
<td>In vivo</td>
<td>Yes</td>
<td>20,000</td>
<td>Not indicated</td>
</tr>
<tr>
<td>Eriksson et al.</td>
<td>57</td>
<td>0.5</td>
<td>Dog femur</td>
<td>In vivo</td>
<td>Yes</td>
<td>20,000</td>
<td>Not indicated</td>
</tr>
<tr>
<td>Eriksson et al.</td>
<td>96</td>
<td>0.5</td>
<td>Human femur</td>
<td>In vivo</td>
<td>Yes</td>
<td>20,000</td>
<td>Not indicated</td>
</tr>
<tr>
<td>Mathews et al.</td>
<td>185</td>
<td>0.5</td>
<td>Human femur</td>
<td>In vitro</td>
<td>No</td>
<td>60–700</td>
<td>60–120 N</td>
</tr>
<tr>
<td>Eriksson and Adel</td>
<td>33.8</td>
<td>0.5</td>
<td>Human mandible</td>
<td>In vivo</td>
<td>Yes</td>
<td>1500–2000</td>
<td>Low and intermittent</td>
</tr>
</tbody>
</table>
the temperature rises with force to a certain point and then falls with greater force. The rise and fall of temperature with force is the result of competing factors. The total heat generated is the product of the rate of heat generation and the duration of drilling. The rate of heat increases with load, while the duration of drilling decreases. According to his data, when the force is high the duration is the dominant factor in temperature determination, while when the force is low the rate of heat generation is more important in determining peak temperatures.

Conflicting results about the dependence of temperature rise on force may be the result, in part, of the wide variation in forces applied in different studies. A similar variation has been found in clinical studies. Hobkirk and Rusiniak measured the force exerted by 20 dental surgeons during drilling in bovine mandibular bone in vitro; they recorded maximum forces between 6 and 24 N. Much higher forces were recorded by Mathews et al. Dental burs cut in a different manner to orthopaedic drills, however, so the two processes may not be comparable.

The effect of drill speed

The amount of heat generated by a cutting bur is dependent upon the frictional forces and shearing forces at the cutting edges. Therefore the amount of heat generated is related to the number of revolutions in the cutting material and the number of cutting edges on the bur.

Mathews et al. studied the temperature of a variety of skeletal pins inserted under different rotational speeds and feed rates. They found that an increase in rotational speed from 300 to 700 rpm increased the total cutting bone-bur contact. This supposedly increased the amount of friction and heat generated; however, this increase in temperature was only observed 0.5 mm from the drill tip. Krause et al. studied various burs at different speeds and found a decrease in temperature with increased operating speed for certain burs only. He concluded that increases in feed rate had greater effects on temperature response than rotational speed.

Research in the dental literature has shown an increase in rotational speed reduced heat generation. However, dental burs operate at speeds of 3600–7500 rpm compared to orthopaedic drills at 60–800 rpm. The forces used are also different, dental 6–24 N and orthopaedic 60–120 N, so comparing the two situations is not straightforward. Many orthopaedic researchers have failed to show a relationship between drill speed and temperature elevation.

A possible factor in the varied relationship between temperature and rotational speed is that the rotational speed of an electric drill depends on the applied force. Abouzgia and James measured the operating speeds of various drills and found them to be at times as low as 50% of the operating speed depending on the force applied. Therefore apparent rotational speeds may not be the actual speeds.

The effect of drill condition

The condition of the drill is important. Mathews and Hirsch compared the temperatures generated by new drill bits to drills which had drilled more than 200 holes and showed signs of blunting of the cutting edges. The worn drills showed greater maximum temperature elevations and longer durations of temperature elevation.

Blockage of the flutes of a drill also reduces its drilling efficiency. Natali et al. compared standard orthopaedic drills with blocked flutes with normal drills. The peak temperatures and the duration of drilling was significantly greater for blocked drills. Flutes are essential to take material away from the cutting zone. This removes heated debri and reduces friction. The flutes of a twist drill often tend to clog when the depth of the hole being drilled becomes appreciable compared to it diameter. Once clogged, friction increases excessively and over heating or even charring of the organic matrix of bone may result. Wiggins and Malkin using a general purpose twist drill have shown that the cutting torque and the specific cutting energy increases with increasing hole depth while drilling in the axial direction. The torque varies linearly and the energy exponentially. The torque may increase further with clogging.

The effect of coolants

The use of coolants can minimise temperature elevations during bone drilling. The mean temperature of a high powered burr can be kept at or below 20 °C using cold Ringer’s irrigation. The temperatures achieved without irrigation for the same situation can be greater than 80 °C. In surgical practice both cortices of a bone are drilled to insert a screw. Irrigation can effectively cool the near cortex but the far cortex is difficult to cool. Mathews and Hirsch examined methods of cooling. Manual irrigation by an assistant is effective in cooling if a drill guide is not being used. Peak temperatures were reduced from 65 to 45 °C. When a drill guide was used, manual irrigation was only effective when the coolant was injected through the guide at a rate of 500 mL/min or greater, only then was it possible to keep temperature elevations below 50 °C. They recommended routine irrigation. When a drill guide was being used, they recommended that the drill guide be removed or back up the shaft of the drill to allow irrigation. If this was not possible, the drill guide should be modified to allow irrigation through the guide.

The effect of bone on drilling temperatures

Bone is thermally anisotropic. In the study of Abouzgia et al., the temperatures recorded in the longitudinal direction were consistently higher than those recorded in the transverse direction. Zelenov, investigating the thermophysical properties of human cadaveric femoral cortical bone, demonstrated a similar result. Lundskog’s work using infrared thermography to investigate temperature distribution in heterogenous specimens and in specimens perforated with drill holes found that the shape of isotherms was essentially circular and thus his bone specimens did not demonstrate thermal anisotropy. Lundskog’s work was performed on dry unspecified sections of elephant bone. The thermal conductivity of wet bone is approximately four times greater than dry bone and anisotropic according to Abouzgia.

Cortical bone is denser than cancellous bone and drilling the former results in the generation of higher temperatures.
Cortical bloodflow is very low normally. Eriksson measured peak temperatures during drilling dog femur 56°C; rabbit femur 40°C and human femur 89°C, he attributed the differences to cortical thickness. The mean cortical thickness of human femur being 6.5 mm, dog 3.5 mm and rabbit 1.5 mm, respectively. Cortical bloodflow in vivo may dissipate some heat produced by drilling during operative procedures. It is felt, however, that this cooling effect is unlikely to be significant. Cortical bloodflow is very low normally (2–3 ml/100 gm) and during drilling coagulation and occlusion of the small vessels probably occurs rapidly. Human red cells in vitro are totally haemolysed at 65°C for 1.5 min.

Mathews and Hirsch compared temperatures generated when drilling human femora in vivo and in vitro and found them to be equivalent. Prolonged ischaemia, as can occur when using a tourniquet, however, raises the threshold for thermal injury in bone. Interstitial fluid, particularly in cancellous bone, may also have a cooling effect. This can be mimicked by using wet bone specimens. Soft tissues surrounding bone are another insulating source or heatsink that can protect bone from excessive temperatures. This protective effect is lost in laboratory based studies. Predrilling the hole with a small diameter drill prior to enlarging the hole to the desired diameter is an effective way of reducing temperature rises. Peak temperatures can be reduced from 108°C (3.2 mm drill) to 46°C when the hole is predrilled with a 2.2 mm drill first. Smaller diameter drills require less energy to penetrate bone and thus generate less heat.

Conclusion

Bone is complex thermal tissue, the understanding of which is improving with research. All orthopaedic interventions produce significant thermal effects, the consequences of which can be far-reaching. Being more aware of the thermal effects of surgery allows one to control the possible detrimental effects, minimising the impact on our results.

References