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Review

# A Review of Surgical Bone Drilling and Drill Bit Heat Generation for Implantation

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**Abstract:** This study aims to summarize the current state of scientific knowledge on factors that contribute to heat generation during the bone drilling process and how these aspects can be better understood and avoided in the future through new research methodologies. Frictional pressures, mechanical trauma, and surgical methods can cause thermal damage and significant micro-fracturing, which can impede bone recovery. According to current trends in the technical growth of the dental and orthopedic industries' 4.0 revaluation, enhancing drill bit design is one of the most feasible and cost-effective alternatives. In recent years, research on drilling bones has become important to reduce bone tissue damage, such as osteonecrosis (ON), and other problems that can happen during surgery. Reviewing the influence of feed rate, drill design, drill fatigue, drill speed, and force applied during osteotomies, all of which contribute to heat generation, was a major focus of this article. This comprehensive review can aid medical surgeons and drill bit makers in comprehending the recent improvements through optimization strategies for reducing or limiting thermal damage in bone drilling procedures used in the dental and orthopedic industries.

**Keywords:** heat transfer; bone drilling; osteonecrosis; bone temperature; drill bit design



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## 1. Introduction

Bone drilling is a fundamental skill for oral, maxillofacial and orthopedic surgeons to treat musculoskeletal trauma due to accident, disease, or ageing [1]. Drilling holes in bone to implant screws for immobilizing broken sections is a common practice in orthopedic and dental surgery. Orthopedic drilling during surgical procedure generates an elevation in bone temperature and forces, leading to osteonecrosis which compromise fixational stability and strength of the implant joint [2]. Mechanical and thermal damages (thermal necrosis) caused by drilling equipment are the most critical factors as illustrated in Figure 1. The surgical outcome depends on the precision of the drilling technique [3,4]. Nowadays, the medical industry is giving first-rate results via the use of cutting-edge technology. These are used in conjunction with an appropriate information system to gather and analyze quality-related approaches. This pertinent information aids in the execution of a high-quality therapy procedure. Traditionally, several procedures have been used to ensure the quality of medical instruments and tools. These technologies now digitally regulate the whole manufacture of dentistry and orthopedic instruments, implants, and gadgets. These technologies are used to verify the quality of every dental and orthopedic product after they have been manufactured [2,3].

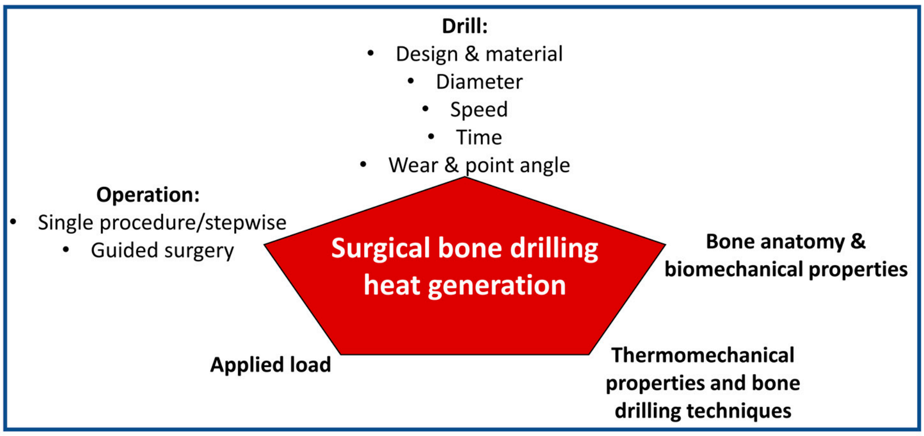


**Figure 1.** Heat generation factors in surgical bone drilling operations.

To ensure the success of bone drilling, both mechanical and thermal factors must be addressed effectively. Mechanically, the pressures applied during bone drilling can be utilized to characterize and assess the effectiveness of the procedure, consequently altering its outcomes. Bone is an anisotropic material with low heat conductivity [5]. These properties attribute to a rapid heat elevation to the surrounding bone tissue due to the heat trapped within the drill bit and bone tissue interface during drilling [3,6,7]. Drilling damage occurs because of a complex mix of bone cutting and extrusion at the drill tip. To ensure the safety of surrounding tissues, the cutting power, torque, and temperature must all be below the critical osteonecrosis level [8].

Numerous variables have been investigated to minimize heat dissipation during bone drilling, which include the drill design, drilling parameters, and coolant delivery [9], [10]. Numerous factors contribute to heat creation during drilling, although previous studies concentrated exclusively on one or a few of these relatively intricate characteristics. However, consensus on the ideal combination of drill design, drilling procedure, and coolant delivery is lacking. The optimal method for estimating the temperature of the bone during drilling is difficult to describe, as bone is a complex anisotropic biological substance composed of organic and inorganic components [6].

In conventional bone drilling, cutting conditions (i.e., speed, feed, and hole depth), drill geometric characteristics (i.e., helix angle, rake angle, clearance angle, tool material, drill diameter, and drill wear), bone-specific parameters (i.e., bone sex, bone density, and bone material), and irrigation (i.e., external, and internal) are regarded as critical parameters. Orthopedic drilling is also performed using non-traditional methods, such as water-assisted drilling, laser-assisted drilling, and ultrasonic-assisted drilling (UAD). Due to the intricate link between these factors and their effect, it remains a concern to more precisely approximate the complying set of actual parametric conditions [11].

In light of the aforementioned difficulties in existing bone drilling, one of the key objectives of a successful operation might be to alleviate thermomechanical damage during bone drilling. While numerous experimental and theoretical research projects have been conducted to investigate the parameters involved with bone drilling, only a few have been comprehensively evaluated. Additionally, due to the complexity of the interactions between these parameters, the problem remains to determine a desirable (and ultimately ideal) set of bone-drilling settings that minimize mechanical and thermal consequences. As a result, this article summarizes experimental and theoretical findings regarding the mechanical and thermal outcomes of bone drilling (i.e., drilling forces and heat generation) concerning bone-drilling parameters. This review aims to better understand bone drilling parameters and the interactions between mechanical and thermal responses to enhance surgical results and reduce accidental harm.

## 2. Bone Anatomy and Biomechanical Properties

Bone is a rigid tissue that protects various organs and helps support the body structure. Bones come in different forms and sizes, with a complicated internal and external structure. Osseous tissues (bone tissues) are lightweight yet strong and rigid, and they fulfil many purposes [12]. Throughout life, osteoblasts, and osteoclasts, which are two types of bone cells, build and repair bones. The names allude to the fact that the two varieties differ in density, known as cortical and cancellous bone, each having their unique appearance and properties [13].

Cortical bone is often known as a compact bone due to its significant density compared to that of the cancellous bone. The cortical bone makes up 80% of the total bone mass in an adult's skeleton. It gives the bone its smooth, white, and solid appearance. The normal structure of a bone is shown in Figure 2. Compact bone consists of densely packed osteons, also called Haversian systems. The osteon comprises concentric rings (lamellae) of a matrix surrounding a central canal termed the osteogenic (Haversian). The bone cells, called osteocytes, are in the spaces between the rings of the matrix. Small channels (canaliculi) sprout from the lacunae to the osteogenic (Haversian) canal to provide pathways through the hard matrix. The osteogenic canals comprise blood vessels, and these blood vessels converge on the bone surface. The dense packing of the Haversian systems in compact bone creates the appearance of a solid mass [14–16].

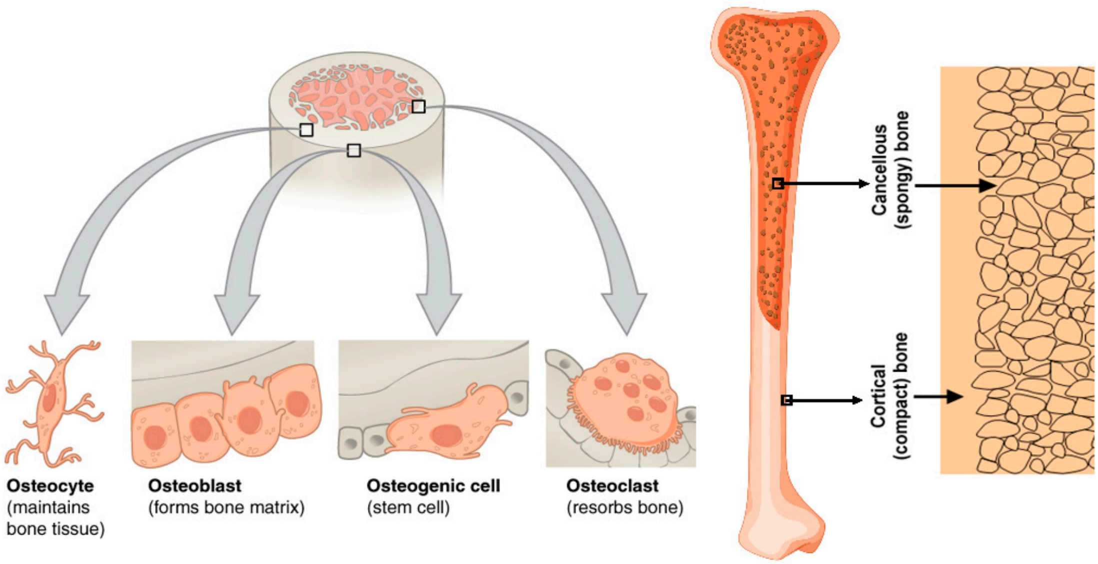


**Figure 2.** Human bone cells anatomy: bone has four cell types. Osteogenic cells become osteoblasts. Osteoblasts deposit matrix. Osteocytes form from osteoblasts trapped in calcified matrix. Different cell lineage osteoclasts resorb bone [17].

Even though bone cells only account for around 2% of bone mass, they are very crucial for proper functioning of bones. Bone tissue consists of four cells: osteoblasts, osteocytes, osteogenic cells, and osteoclasts. The osteoblasts are responsible for the mineralization of the bone matrix by synthesizing it. Osteocytes are mesenchymal cells that develop from osteoblasts that have moved into and lodged in a bone matrix that they have formed themselves. Osteogenic cells are bone stem cells that play an important role in bone healing and development. Bone resorption is caused by osteoclasts, which are responsible for the disintegration of bones. The osteoblasts then create new bone. Osteoclasts resorb, and osteoblasts form bone, which is continually rebuilt. An equilibrium between osteoblasts and osteoclasts maintains the bone tissue [15–18].

## 3. Thermomechanical Properties of Bone Tissue

It is difficult to assess the mechanical and thermal characteristics of bone tissue. The challenge arises from the large range of qualities that differ from person to person depending on age, gender, and disease [19,20]. The mechanical and thermal properties of bone are inconsistent since there are two forms of bone, which will be reviewed in the subsequent section.

### *3.1. Mechanical Properties*

Several researchers have discussed the mechanical characteristics of bone tissue over the years. A wide range of mechanical testing procedures for measuring bone characteristics have been described. Mechanical properties of bone tissue can be determined by different methods of testing, as shown in Figure 3.

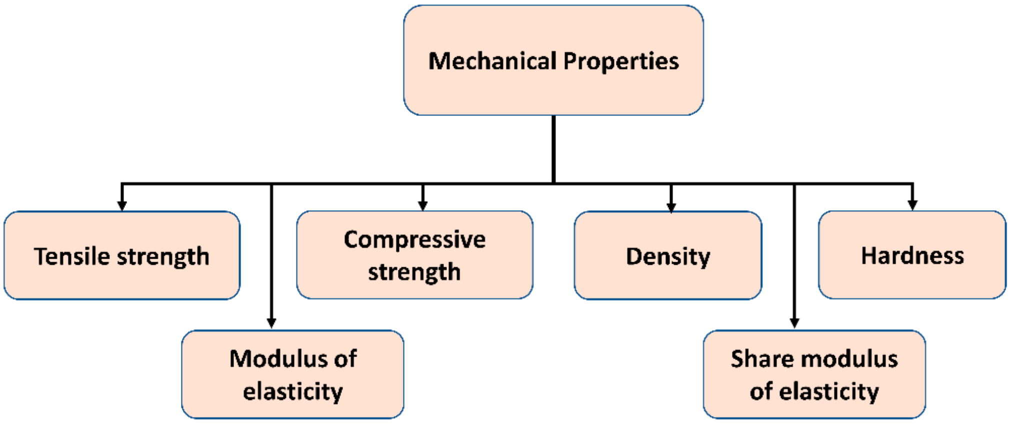


**Figure 3.** Mechanical properties of bone.

When a load is changed to stress and displacement to strain, the curve denoting the mechanical behavior of bone is referred as the stress–strain curve. Significant mechanical parameters, such as elastic modulus or Young modulus, yield stress, and ultimate stress can be identified in this curve (Figure 4). These properties highly depend on the loading mode (tensile, compression, bending, or shear) and influence how the bone tissue responds mechanically to drilling operations.

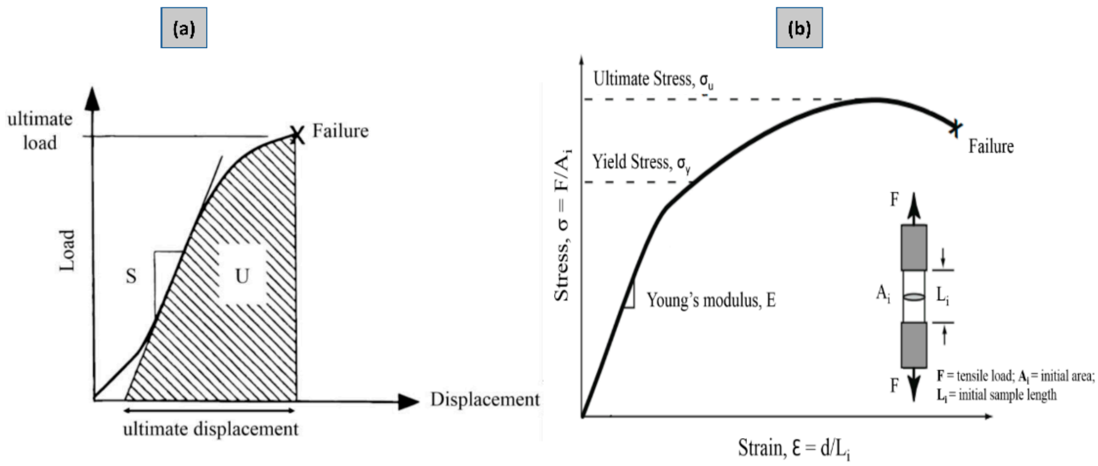


**Figure 4.** (**a**) Load-displacement curve; (**b**) tensile stress–strain curve of bone [21].

According to previous studies, the bone matrix is composed of two components: a mineral component composed of hydroxyapatite (HA), which accounts for 65–70% of the matrix, and an organic component composed of glycoproteins, proteoglycans, sialoproteins, and bone proteins, which accounts for the remaining 25–30% of the matrix [22,23]. Theories of composite materials are applied to bone with different phase compositions, and the probable ways of deformation under stress are discussed. Bone is characterized as a composite material made up of high elastic modulus mineral "fibers" that are embedded in a low elastic modulus organic matrix that is permeable to liquid through pores. As a result, from the standpoint of materials science, bone can be considered a composite material. Table 1 compares some mechanical properties of human and bovine bone reported in the literature.

**Table 1.** Mechanical properties comparison of bovine bone and human bone properties [10].

| Properties | Human Bone | Bovine Bone | Drill Bit |
|---|---|---|---|
| Density (kg m$^{-3}$) | $2.10 \times 10^3$ | $2.10 \times 10^3$ | $7.99 \times 10^3$ |
| Young's modulus (Pa) | $1.70 \times 10^{10}$ | $2.20 \times 10^{10}$ | $1.93 \times 10^{11}$ |
| Shear modulus (Pa) | $3.00 \times 10^6$ | $3.00 \times 10^6$ | $9.70 \times 10^8$ |
| Tensile strength (Pa) | $2.00 \times 10^8$ | $2.50 \times 10^8$ | $5.79 \times 10^8$ |
| Yielding strength (Pa) | $1.10 \times 10^8$ | - | $6.08 \times 10^8$ |
| Poisson's ratio | 0.40 | 0.33 | 0.30 |
| Specific heat (J kg$^{-1}$ K$^{-1}$) | $1.26 \times 10^3$ | $1.30 \times 10^3$ | $5.00 \times 10^2$ |
| Thermal conductivity (W m$^{-1}$ K$^{-1}$) | $3.80 \times 10^{-1}$ | $3.00 \times 10^{-1}$ | $1.70 \times 10$ |

### *3.2. Thermal Osteonecrosis Analysis*

Apart from the mechanical attributes, the thermal effect of bone drilling induces temperature rise. This effect is critical in terms of bone injury since it can result in necrosis, or the irreversible loss of bone cells, resulting in infection and decreased mechanical strength. Thermal damage occurs when a critical temperature is exceeded for an extended period. The established threshold for thermal osteonecrosis is 47 °C for 60 s [24–26]. To accurately predict temperature rise during bone drilling using experimental and computational models, it is necessary to understand bone tissue's geometry, heat input, and thermal characteristics. The primary parameters regulating thermal effects during bone drilling are specific heat (a measure of a material's ability to heat up), thermal conductivity (a material's ability to conduct heat), and thermal expansion coefficient.

The temperature and thermal conductivity are inversely related. Temperatures at the drill–bone interface rise as the values decrease. The temperature is mostly unaffected by the tool material's thermal conductivity. Nearly 85% or more of the heat created during metal drilling is dissipated in the form of metal chips. Since the bone has a lower thermal capacity and conductivity than metal, the temperature rises during the drilling process, resulting in a greater rise in the temperature of the bone. Due to bone's lower K value (conductivity), the total heat generated under the same heating circumstances is significantly lower than in the case of metals [27,28].

There are numerous techniques in determining a material's thermal conductivity [29–31]. They can be classified according to their steady-state or transitory natures [32], [33]. In steady-state configurations, a known continuous heat flow is considered to pass through an object (the measurement sample). These setups are often arranged in a parallel plate format to produce a temperature gradient between the sample and a known thermal conductivity reference sample. To determine the heat flux, a reference sample is required. Typically, these measurement systems are implemented using a heat flow sensor. The thermal conductivity can be calculated quite easily if no (lateral) heat loss and perfect heat transfer between the samples are assumed in Equation (1):

$$\dot{q}_{bone} = \dot{q}_{reference} = \frac{kA}{l} \cdot \Delta T \quad (1)$$

with "$\dot{q}$" being the heat rate (W), "$k$" the thermal conductivity (W/mk), "$A$" the cross-sectional area ($m^2$) and "$l$" the length (m) of the object (sample) (Incropera et al., 1996) [31]. The temperature difference between the two ends is denoted with "$\Delta T$" and the temperature drop is assumed to be linear within a homogenous sample. When the material and reference sample have the same dimensions, Equation (1) can be simplified and stated as the thermal conductivity of the material under test Equation (2) [31]:

$$K_{bone} = K_{reference} \cdot \frac{\Delta T_{reference}}{\Delta T_{bone}} \tag{2}$$

The transient approach is more complex and is based on the temporal behavior of the temperature change of a heated sensor embedded in the material.

The Cumulative Equivalent Minutes (CEM) methodology as denoted in Equation (3) considers a given temperature and duration of exposure and produces an equivalent number of minutes at the accepted critical temperature threshold [34–36]:

$$CEM_{47} = \int_0^T R^{(47-T(t))} \mathrm{d}t \tag{3}$$

Here, $\Delta t$ is the duration of exposure; $R$ is related to the temperature dependence of the rate of cell death for temperatures below the critical value of 47 °C, $R$ = 0.25, and temperatures at or above the critical value, $R$ = 0.5; $T(t)$ is the temperature of the bone at time $t$. Tissue damage is time dependent and temperature dependent and the necrosis threshold is measured in so-called cumulative equivalent minutes. This model takes a particular thermal dosage and compares it to the established critical parameters to determine the occurrence of necrosis.

## 4. Surgical Bone Drilling Practice

Bone drilling is a machining procedure in which a rotating cutting tool is fed into the bone parallel to its axis of rotation and creating a hole. The primary reason for drilling holes in bone tissue is to allow screws or other threaded devices for rigid fixation, which is accomplished by integrating bone (cortical or trabecular) with the screw threads. A typical block diagram of the bone drilling process is shown in Figure 5.

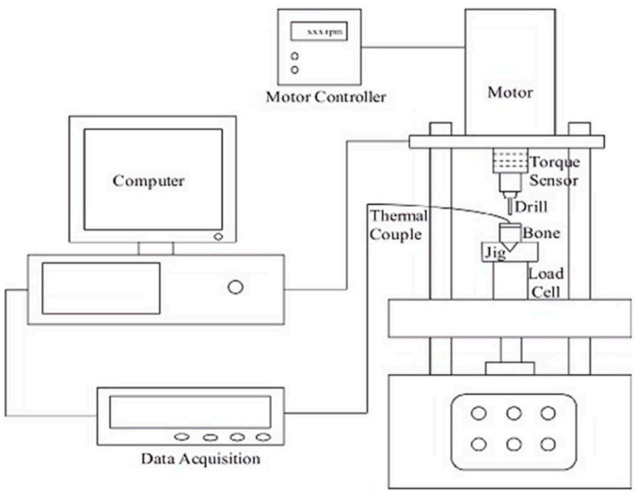


**Figure 5.** Schematic diagram of the bone drilling process [37].

Nowadays, bone drilling is an increasingly common surgical operation. As a result of the ageing population, the prevalence of bone diseases is expected to rise dramatically. Due to human ageing and associated disorders, it is vital to develop proper bone drilling methods to fully understand the impacts of drilling mechanics [36–39].

While the drilling technique is an essential and fundamental ability for surgeons, drilling poses a significant risk to the patient's body. To begin with, there is a risk that implanted devices will fail or that bone fractures will occur due to inadvertent fastening. Excessive drilling and high forces can lead to drill bit breakage within the bone tissue [40–43]. Motoyoshi et al. [44] recommended torque values for tightening an orthodontic mini-implant to avoid implant device failure. The second risk is inadvertent harm to the surrounding tissue because of an error in the drilling site's location or the hole's depth. When drill bits penetrate deeper than necessary, significant injury to the bone tissue, nerves, or vascular channels close to bone tissue can occur, which could result in medical emergencies, such as profuse bleeding, paralysis, or altered sensation. Finally, bone necrosis, often referred to as osteonecrosis, can be produced by either excessive force or high temperatures associated with drilling. Osteonecrosis is a severe risk to the bone structures since it not only delays bone cell regeneration but also facilitates bone fracture; therefore, the effects of drilling on temperature elevation have been thoroughly investigated. After Eriksson et al. [45] established in 1984 that bone temperature should not exceed 47 °C for 1 min to avoid osteonecrosis, numerous variables, such as machining settings, machining equipment, and cooling systems became the primary focus of bone drilling research.

Considering these issues, surgical education and mechanical testing of medical devices are critical, as surgical outcomes are highly dependent on the operator's command of medical devices. Moreover, there exists research potential into the impacts of surface coating or texture on drill bits and the enhancement of numerical models of temperature rise during bone drilling.

## 5. Conventional and Non-Conventional Bone Drilling

The conventional drilling operation is carried out by high-speed rotational impact of a cutting instrument, the drill bit, against the surface of a workpiece, guided by the action of a mechanical thrust. The cutting edge of the drill bit, which rotates at a fast rate of speed, typically hundreds to thousands of revolutions per minute, then shear deforms the workpiece material, removing chips [46–49]. In the medical industry, conventional bone drilling with twist drill bits and sophisticated drills has increased interest. The major objectives of this research on traditional orthopedic drilling are as follows:

1. Researchers undertake numerous tests to determine the effect of various parameters (feed rate, drill bit shape, spindle speed, and bone structure) and output reactions (such as temperature, thrust force, surface roughness, and delamination).
2. Conventional bone drilling raises the temperature of the bone, and a temperature greater than 47 °C is dangerous because it leads to thermal bone necrosis. Researchers have examined the influence of machining parameters on bone tissue thermal damage.
3. Delamination studies: The primary purpose of these investigations is to alleviate delamination caused by bone hole drilling.
4. Effects of tool materials, drill bit shape, and tool wear on the quality of holes and thrust forces induced during bone drilling.

Bone machining in conventional processes is challenging, especially when machining heterogeneous and anisotropic bone structures, due to its low thermal conductivity, heat sensitivity and abrasive nature [50,51]. In conventional methods, tool materials, geometry, and operating conditions must be properly optimized to lower the heat generation rates and avoid thermal or mechanical damage. In most conventional processes, the surface quality obtained is poor due to excessive tool wear and low cutting rates. The major objective in bone machining with non-conventional processes includes reducing drilling temperature or improving surface quality which is difficult through conventional processes [52]. Certain non-traditional drilling techniques, including ultrasonic aided drilling (UAD), water-jet drilling, vibrating drilling, automated drilling, and rotary ultrasonic bone drilling (RUBD), have demonstrated advantages over conventional drilling techniques. These are good in producing holes of superior quality with a smoother surface finish, making them suited for surgical applications [46–54].

## 6. Bone Drilling Heat Generation

Drilling into bone tissue involves two distinct processes: the first is cutting the bone tissue by the drill bit's cutting edges, and the second is friction between the drill bit material and the bone [33,55,56]. Both activities generate heat in the bone tissue. The created heat is partially absorbed by the formed chips, while the remainder is attributed to the increase in the bone's temperature. Due to the limited conductivity of bone tissue, the temperature increase in the bone cannot be dispersed quickly. During drilling, there are two primary sources of heat: plastic deformation and friction. During the cutting process, plastic deformation and friction occur between the bone tissue and non-cutting components of the drill bit [31,33,57–59]. Drilling inevitably generates heat. Referring to Figure 6:

1. The primary sources of heat are shear deformation of the work material (1), friction between the cutting chips from the work material and the rake face of the cutting tool (2), and friction between the cutting edge and the under surface of the work material that touches the relief face of the cutting tool (3).
2. Secondly, the indirect heat sources are simply driven by friction between cutting chips, particularly between bone chips and flutes, or between bone chips and the drilled wall of the work material when travelling the flute.

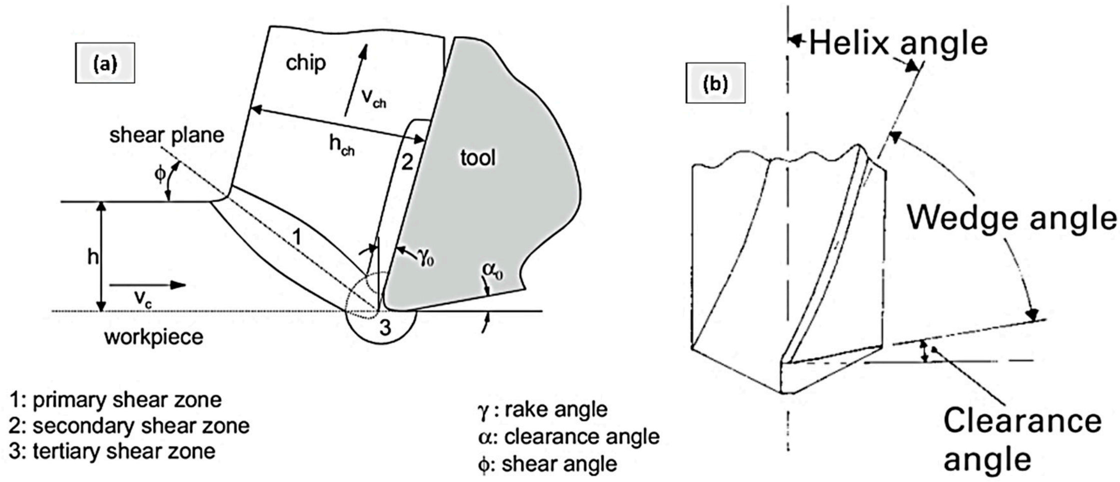


**Figure 6.** Schematic representation of (**a**) material removal and heat generation regions during chip forming process; (**b**) rake and relief angles [2].

In the case of bone drilling, roughly 60% of the heat energy created during the drilling process can be transferred to the bone chips, with the remainder being transferred to the surrounding tissues and the drill bit itself [60]. Numerical and mathematical models are now being developed to analyze the formation and dissipation of heat during bone drilling. Relief angles are incorporated into the cutting tools to alleviate thermal dissipation and mechanical wear caused by friction between cutting tools and the emerging surface of work materials. Chacon et al. [2] demonstrated that relief angle substantially affects the temperature elevation scale during bone drilling. The friction coefficient μ is used to account for the shear stress of the surface traction, $T = \mu p$, (where $p$ is the contact pressure). In this case, the frictional contact between the drill bit and cortical bone was modelled with a constant coefficient of friction of 0.7 [55].

To allow the modelling of heat generation during drilling of the cortical and trabecular bone tissue (as illustrated in Figure 7), the geometry of the cutting edges and the contact areas between the drill bit and bone tissue must be carefully studied to deduce the deformation zone in the bone tissue and the friction zones at the contact points between the drill bit geometry and the surrounding bone tissue [61–65].

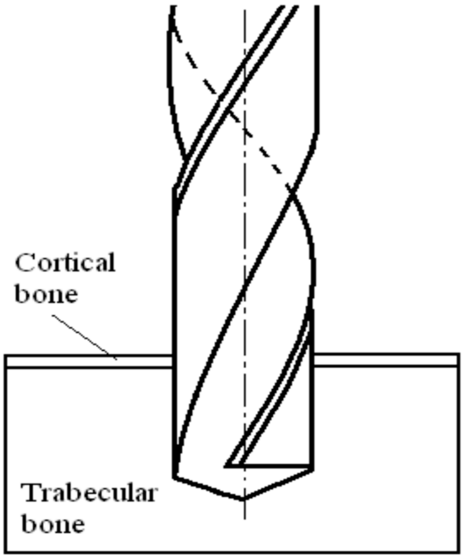


**Figure 7.** Schematic showing drilling into cortical and trabecular bone [21].

## 7. Characterization of Bone Drilling

Drilling behavior is determined by various factors, which includes the cutting tools used, the machining conditions, and the work material's mechanical, thermal, and chemical qualities. Cutting forces (thrust force and torque) and resulting temperatures can be used to evaluate the drilling performance [66–68]. Drilling time and feed rate are other critical parameters to be considered while drilling with constant thrust force. Additionally, drilling behavior can be defined by the cutting tool's life, which is measured by cutting edge wear, the quality of drilled holes as measured by surface roughness and dimension accuracy, and the cutting chips formed during drilling [69–71]. Since these characteristics cannot be determined directly from the mechanical properties of materials, drilling tests must be conducted to characterize the drilling behavior.

### *7.1. Cutting Temperature*

Temperature elevation during drilling is mostly determined by two methods: thermal images acquired with an infrared camera, or thermocouples embedded in bone. Both systems have their advantages and disadvantages. When an infrared camera is used, thermal images of the work piece's surface can be taken while drilling, making the temperature rise inside the drilled hole more visible than when thermocouples are used. However, cortical bone typically has a thickness of less than 5 mm in the radial direction. Thermocouples are not always acceptable due to space constraints. It is critical to distinguish the measurement objective and choose the appropriate method [72–74].

### *7.2. Cutting Tool Wear*

Repeated contact between the cutting edge and the emerging surface of work components results in wear and dulling of the cutting edges, necessitating the application of a greater thrust force to advance the drill bit [40,43,75]. Cutting tool wear can result in defective cutting, typically associated with an increase in temperature and the onset of vibration because of an increase in the surface roughness of the cutting edges. Observing cutting edges with an optical microscope or a scanning electron microscope (SEM) is one imaging technique for analyzing cutting edge wear [43,54,76,77]. According to the literature, both abrasive wear and plastic deformation can modify the geometry of the chisel and cutting edges, as well as the rake face of the drill bit, as shown in Figure 8.

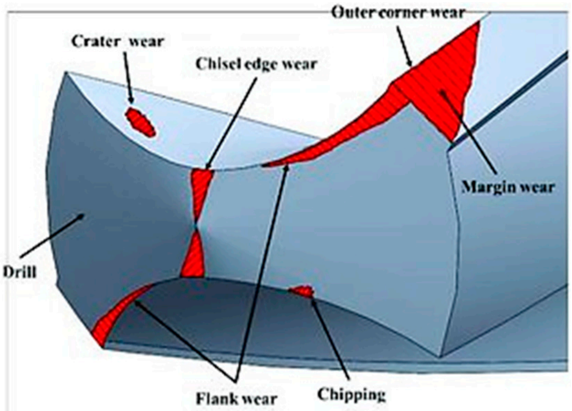


**Figure 8.** Drill bit wear [78].

## 7.3. Exposure Time

Heat is transferred from a region with a high temperature to a low temperature region. The heat transmission rate between two regions depends on the temperature gradient between them. Heat is transferred from high to low temperatures until equilibrium is reached, and the temperatures in both locations are equal. Throughout the drilling process, heat is transferred from the interface between the drill bit and the bone tissue, which is the source of heat creation, to the adjacent locations [2,33,76,78,79]. Because the drilling period is so brief, equilibrium will not occur. This indicates that the amount of heat created at the drilling location is proportional to the duration of the drilling process. Different studies reported a range of temperatures at which thermal necrosis begins, including 47 °C, 50 °C, and 55 °C. These discrepancies are due to the fact that the drilling time durations in the experiments varied [54,80–84].

## 7.4. Initial Temperature of the Drill Bit

The amount of heat generated during bone drilling is dependent on the initial temperature of the drill bit. The drill bit's initial temperature may remain elevated following the high-temperature sterilizing operation [85,86]. Yuan-Kun Tu [22] conducted simulations by utilizing four initial temperatures, $T_o$, at 30, 35, 40 and 67 °C, respectively. The rest of the parameters were constant during drilling activities. Figure 9 illustrates the temperature fluctuations of bone as a function of drilling duration for four different initial drill bit temperatures.

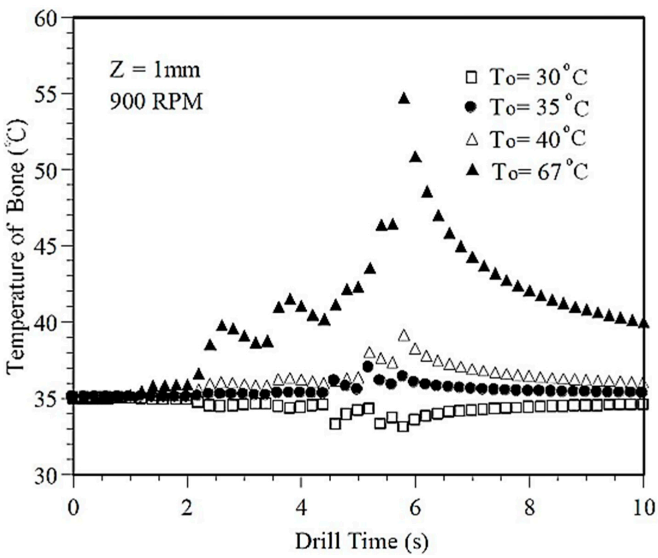


**Figure 9.** Variation in temperature with the drill time for various initial temperatures [21].

The simulation findings indicate that the beginning temperature of the drill bit significantly affects the temperature of the bone surrounding the drilling site, as illustrated in Figure 8. It is suggested that dentists cool the drill bit before undertaking surgical drilling if the drill bit is sterilized at a high temperature before use.

### 7.5. The Effect of Feed Rate

Temperature increase is inversely proportional to the feed rate. Increasing the thrust force speeds up the drilling process, lowering the maximum temperature that can be reached. Figure 10 illustrated that, for both pilot drills and twist drills, the maximum temperatures raised in bone decreased as the feeding rate under a given drill speed. The studies conducted by Shin and Yoon demonstrate that increasing the feed rate decreases the maximum temperature [87]. Other investigations have discovered that as force is applied, temperature increases. However, the forces employed in these experiments were relatively light (thrust force was less than 30 N). Davidson and James concluded that these contradictory results might be explained by the fact that the maximum temperature increases with feed rate up to a certain point, after which the temperature declines slightly [55].

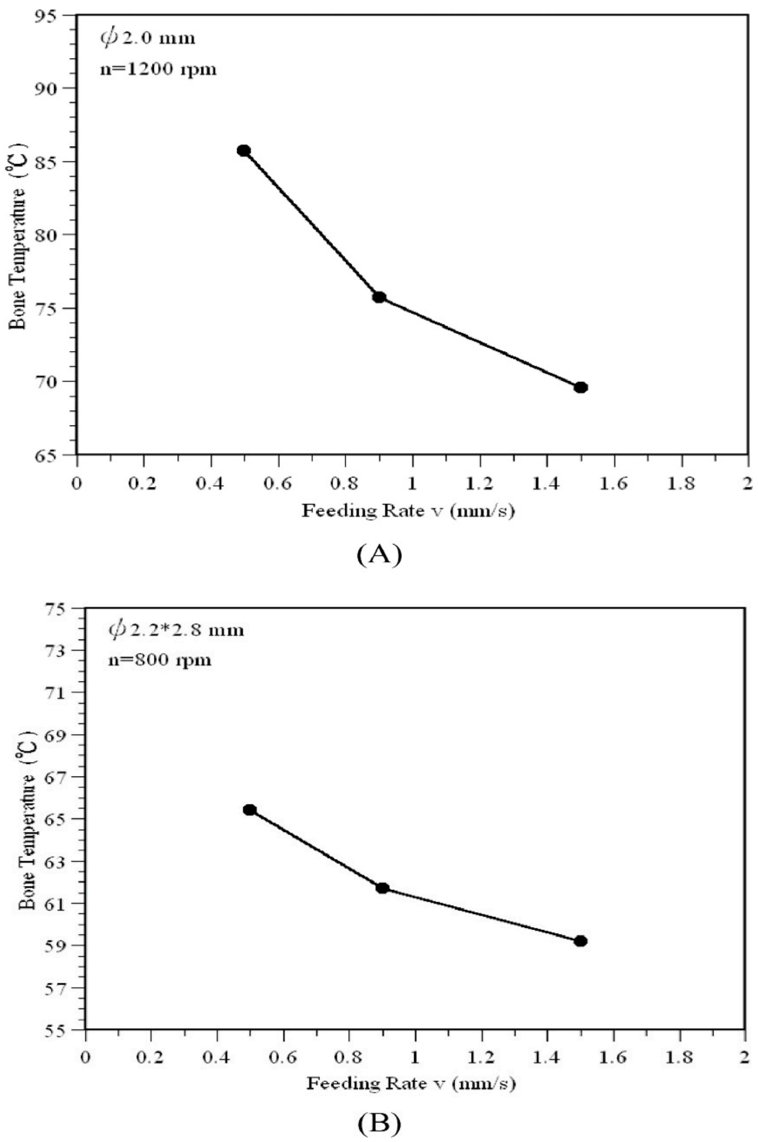


**Figure 10.** Maximum temperature decreases as the feed rate using two types of drills: (**A**) pilot drill, (**B**) twist drill [37].

When researching the impacts of feed rate, two elements should be considered: the high thrust force, which may cause increased deformation and hence a rise in the generated heat, and the short drilling length, which causes less damage to the bone tissue.

### *7.6. Drilling Speed*

Drilling speed significantly affects temperature rise and heat transfer during the drill operation. As cutting speed increases, the generated heat increases and became focused in the cutting region (shear deformation zone). This suggests that the effect of a bone's thermal behavior is particularly significant during high-speed drilling. Multiple experimental studies have demonstrated that increased spindle speed leads to decreased thrust force and torque during bone drilling [41,88,89]. Researchers also developed a theoretical model based on the assumption that thrust force and torque decrease with spindle speed. Increased spindle speed thermally increases the amount of friction energy created by friction forces acting on the drill bit's rake face. Friction energy is approximately linearly linked to spindle speed. Because a significant percentage of the cutting energy is converted to heat, increasing temperatures are expected at higher speeds. Numerous experimental tests and mathematical models have established that spindle speed rises with temperature [90–93].

### *7.7. The Effect of Coolants*

Coolant is used to keep the drilling site at a safe temperature. There are two forms of irrigation: manual irrigation, in which the dentist or an assistant manually adds coolants, and automatic irrigation, in which coolants are added by the drill guide [54,94–97]. Manual irrigation significantly reduces the temperature. Irrigation via the drill guide is successful only when the coolant is injected at a relatively high pace.

## **8. Surgical Drill Bit Geometry**

The geometry of a surgical drill has a significant effect on the drilling process. Drilling force, torque, temperature, and hole quality are just four geometry-dependent characteristics. All relevant details relating to the geometry of surgical drill bits will be disclosed in the following sections. Although various authors have investigated this subject, there is no widespread agreement on the optimal geometry. A drill bit is composed of three components: the drill point, the body, and the shank; the anatomy of a typical drill bit is seen in Figure 11. The drill point component includes the angle of the point, the chisel edge, the web thickness, the rake angle, and the clearance angle. In comparison, the drill body is made up of a helix angle and a flute [40,88,98,99].

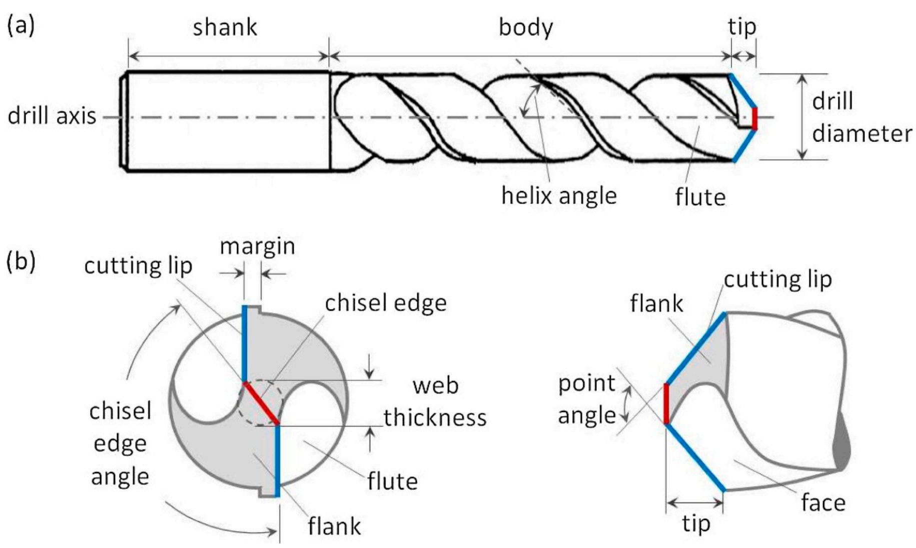


**Figure 11.** Surgical drill bit nomenclature: (**a**) twist drill bit; (**b**) axial view of drill bit tip [16].

### *8.1. Drill Diameter and Predrilling*

Surgical bur or drill instruments typically have a diameter of 0.4–3.8 mm [100,101]. There is evidence suggesting that the diameter of the drill bit has a significant impact on the forces and temperature rise during surgical drilling. By increasing the diameter of the drill bit, the contact area between the drill bit and the bone is increased, resulting in a greater amount of bone material being removed per revolution [102–105]. Numerous experimental studies concur that increasing the diameter of the drill bit increases the thrust force, torque, and temperature generated during bone drilling.

To minimize thermal damage to bone, researchers used a two-step drill bit. The concept of step drilling is connected to a bore hole predrilling. A two-step drill's body has two distinct diameters. At the beginning of the drill point, the smaller diameter predrills the hole for the larger diameter (Figure 12) [41,43,64,106]. Augustin et al. found no evidence that their step-drills result in lower bone temperatures than a conventional drill bit of the same diameter [107]. Udiljak et al., on the other hand, showed a difference of 17 °C in favor of the two-step drill at low cutting rates (6.53 m/min) [108]. Predrilling is the other option to reduce the bone temperature. Predrilling means drilling in multiple steps to reach the final diameter of the drill hole. Since the drill diameter gradually increases, the friction between the drill bit and bone is decreased.

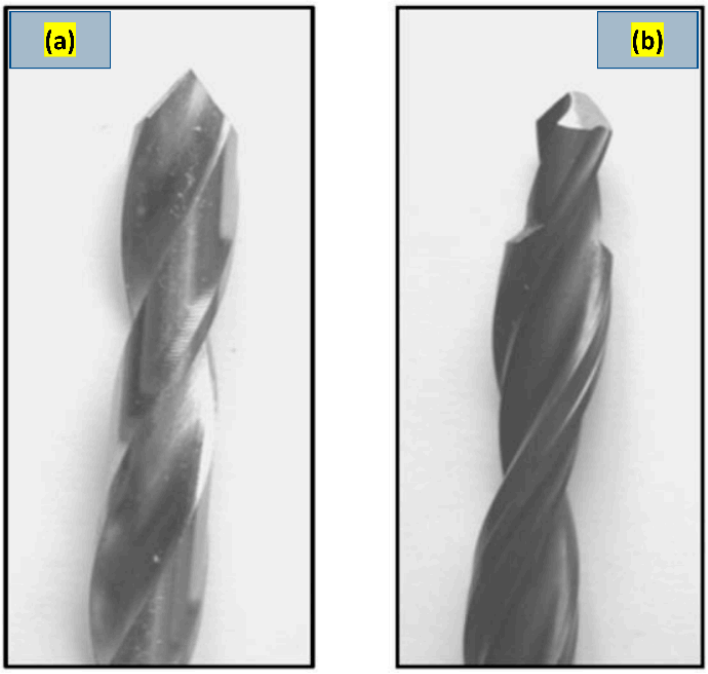


**Figure 12.** Standard surgical (**a**) twist drill; (**b**) two-step drill.

To summarize, it is well established that the drill's diameter significantly affects the drilling temperature. This should be viewed through the lens of thermal necrosis. Predrilling may be an effective technique for reducing the thermal burden during drilling, particularly for big diameters. However, it should be remembered that predrilling costs more time, extending the duration of the operation. This is a fact that should be considered before making additional recommendations.

### *8.2. Point Angle*

The point angle is formed by the two cutting edges of the drill bit. It is primarily used to center the drill (Figure 13). Researchers continue to argue about the effect of tip angle on bone drilling performance. Numerous researchers asserted that a point angle between 70 and 120 degrees had no damaging effect on heat generation. However, the range of analyzed point angles is limited, and the findings are unconvincing [10,109–111].

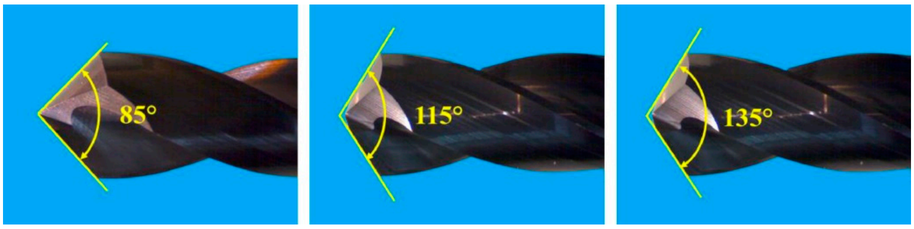


**Figure 13.** Drill bit point angles [109].

Udiljak et al. [112] observed an increase in force with higher point angles but the bone temperature remained unchanged. However, it is well accepted that force controls the creation of heat during bone drilling; increasing force raises the bone temperature. Others discovered that point angle considerably affected bone injury, but their data are rather inconsistent. For instance, the first group demonstrated that increasing the tip angle of the bone increases its warmth and force. In comparison, others discovered a decrease in heat generation and torque with increasing point angle. According to these data, a conflicting effect of torque and thrust force contributes significantly to bone injury. When the point angle is raised, the thrust force increases, but the torque (twisting force) decreases. To summarize, various researchers have examined the significance of the point angle. The optimal point angle for bone drilling, on the other hand, has yet to be found [113,114].

### *8.3. Drill Bit Helix Angle*

The helix angle is formed by the drill bit's longitudinal axis and a tangent to the land's leading edge. Surgical twist drill bits are frequently slow spiral, which results in a short helix angle [2]. This modest helix angle was optimal for bone drilling. Helix angles are classified as slow (12–22°), regular (28–32°), and fast (34–38°). A short helix is used to drill soft material, while a standard helix is used for general drilling. While a slow helix is favored when drilling brittle materials (cast iron, bronze, and brass), it is not preferred when drilling deep holes [41,111,115]. The primary component affecting heat generation concerning drilling dimensions is the diameter of the drill bit, as it has a greater cutting surface. Since the helix angle has a negligible effect, it can be ignored. Temperatures decreased as the helix angle increased, although the effect was negligible compared to other elements, such as drilling speed and feed rate [40].

### *8.4. Clearance Angle*

The clearance or relief angle is the angle ground at the flank which allow bone debris to depart the cutting edge without rubbing against the drill flank or hole wall (Figure 6). Additionally, this angle facilitates the cutting process by allowing the cutting edge to penetrate the bone. The clearance angle varies according to the drill diameter, and the clearance angle for a general-purpose drill bit is between 8–15°. Due to the tremendous cutting force required to cut the bone, a low clearance angle may cause the drill to jam or shatter. On the other hand, an excessive clearance angle increases drill temperature due to insufficient number of cutting lips to evacuate heat from the drilling hole [38,95,116–118].

In bone drilling, a greater clearance angle reduces the temperature of the bone, the thrust force, and the torque. According to Farnworth and Burton, a clearance angle of 15° is sufficient for drilling the pig femur. Similarly, Saha et al. incorporated a clearance angle of 12–15° into the new drill and observed an improvement in drilling performance [119]. Fuchsberger advocated a larger clearance angle of 18 to 24° to minimize heat damage. The ideal clearance angle for bone drilling, on the other hand, remains unknown [104].

## 9. Temperature Measuring Method

Generally, two methods are employed to determine the temperature of bone during drilling: thermocouples, or an infrared thermographic camera. Few research articles have reported on the application of both techniques. A thermocouple is a temperature sensor comprised of two distinct conductors (often metal alloys). Due to the temperature differential between the ends of both wires, there is a relative difference in the voltage between the conductors and the usable range of thermocouples. The distinction between them is due to the metals' composition, which dictates the temperature range and sensitivity. Low-sensitivity thermocouples are (b, r, and s types), whereas high-sensitivity thermocouples are (e, j, k, and t types). Thermocouples are frequently constructed with a wide temperature range in mind, such as high sensitivity thermocouples. Figure 14 illustrates an experimental setup in which a thermocouple is introduced into a bone sample to the depth of the drill side wall. The inbuilt thermocouple is coupled to a data recorder and is used to monitor the temperature during bone drilling [10,41,43,120].

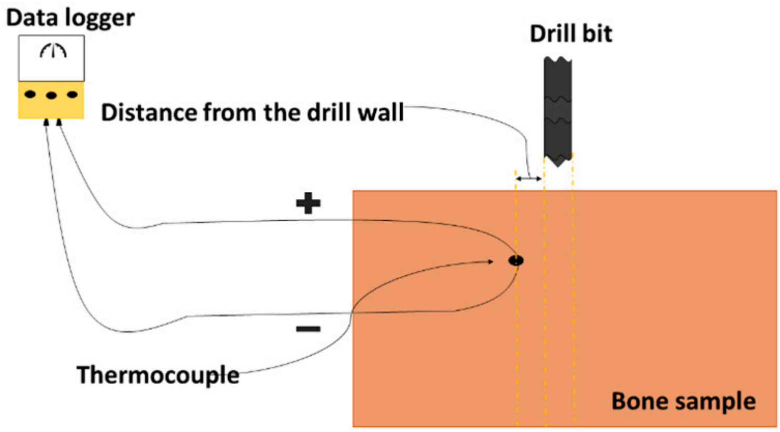


**Figure 14.** Temperature measurement using a thermocouple [26].

The use of thermocouples in medicine was originally documented by Horch and Keiditsch [103] in dentistry. Thermocouples are also installed into the drill bit, with the drill bar bonded around the drill barrel, allowing temperature measurement by rotating the instrument [120]. Other options include the insertion of several thermocouples at multiple surface or bone tissue depth locations. Typically, thermocouples are placed at increasing radial distances of 0.5 mm, 1 mm, and 3 mm from the drilling site, which is suitable for analyzing the effects of temperature on the drilling process [84].

Figure 15 illustrates a thermal infrared (IR) camera at the drilling site recording temperature rise. The IR thermal camera communicates with a computer, in which raw temperature distribution data can be saved in real-time. In a bone drilling experiment, the IR thermal camera would make monitoring temperature elevation more convenient. The thermal IR camera should be situated near the drilling site to capture the heat distribution more accurately. To better understand the temperature profile, IR thermography can be utilized in conjunction with the thermocouples.

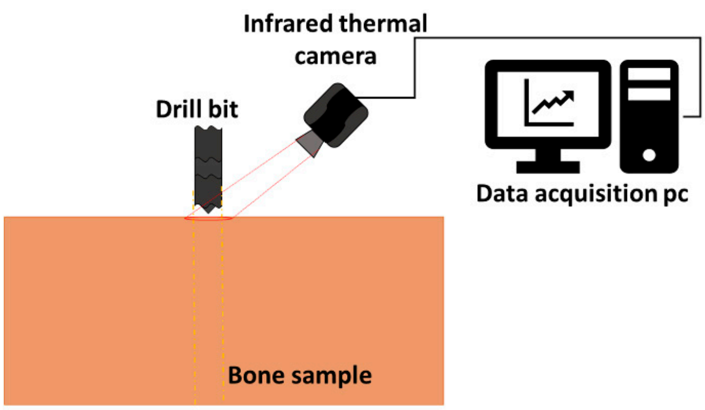


**Figure 15.** Temperature measurement using a thermal infrared camera [26].

## 10. Challenges in Bone Drilling

Orthopedic surgery involves surgeries conducted to treat disorders in the human body's bones, joints, and ligaments. Artificial hips, knees, spinal implants, plates, and screws to cure shattered bones and joint fusions and implants to treat arthritic disorders are examples of such operations. Physical issues, such as a broken bone, may necessitate surgery and, as a result, drilling of the bone is inevitable. The surgeon may need to drill, cut, or shape bones to anchor the implant to the bone (Figure 16), allowing healing to occur and the patient to resume normal activities.

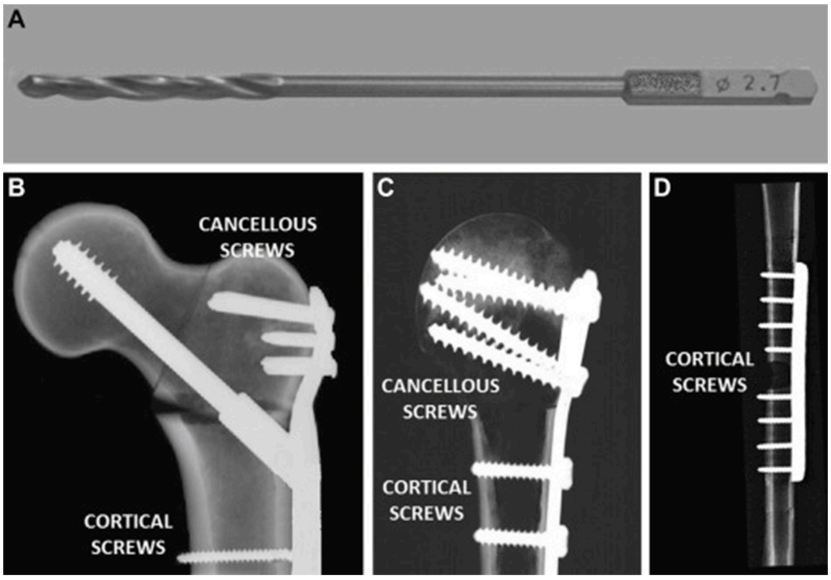


**Figure 16.** Surgical drilling for fracture repair. (**A**) Surgical drill bit, (**B**) proximal femur, (**C**) proximal humerus, (**D**) humeral diaphysis [121].

Orthopedic surgeons confront a few obstacles while drilling bone. Simply put, the trend of drill wear is not obvious to a surgeon, forcing them to decide on drill replenishment only on feel. Orthopedic drills (Figure 16) are frequently repeated without tool wear data. As a result, drills that have become dull or are discarded when they still have useful life are used. According to the Association for Surgical Technologists, healthcare facilities encourage the reuse of single-use equipment to deliver efficient, cost-effective medical/surgical care.

Heat transmission from drilling into the bone is increased due to dull and inefficient bone drills, poor geometries, and inadequate chip evacuation. This heat output may be greater than the bone's ability to recover, resulting in bone death. While the precise temperature at which thermal osteonecrosis begins has yet to be discovered, 50 °C is widely considered the key figure, as bone regeneration is virtually completely impeded from this point on. Furthermore, poor-performing bone drills frustrate surgeons and lengthen procedures unnecessarily. Unfortunately, consequences resulting from localized bone death caused by heat during bone drilling have received little attention. There has been researched on screw fixation, in which screws come out of the bone after surgery. One possible reason for screw failure is that the screws are attached to bone that has been heat damaged due to drilling temperatures above the bone's ability to regenerate.

## 11. Summary: Future Directions/Guidelines

The success of bone drilling surgery is highly dependent on limiting thermal and mechanical damage. Thus, critical bone-drilling parameters, such as drilling conditions, drill bit geometry, bone structure, surface roughness, hole quality and many recent experimental and theoretical investigations examined the influence of drilling techniques on mechanical and thermal responses during bone drilling. While bone drilling's fundamental features are comparable to material shearing in metal and polymer machining, the inhomogeneity and anisotropy of the bone structure can greatly hinder the quality of bone drilling. Additionally, due to bone's low thermal diffusivity, the heat created during bone drilling can cause severe tissue damage, eventually resulting in thermal necrosis.

We have generally synthesized the literature on bone drilling to demonstrate the progress toward alleviating bone damage through modification of present surgical drilling processes. It is envisaged that this comprehensive analysis of mechanical and thermal impacts of bone drilling parameters would lead to additional contributions toward identifying favorable conditions that are optimal for certain surgical reasons and thus improve the likelihood of successful surgery. To maximize its efficacy and reduce bone injury during bone drilling, a few factors should be considered.

In typical drilling procedures, the cutting speed has the greatest effect on the cutting temperature. Increased cutting speeds increase temperature and decrease applied forces and torque to the bone. Drills should have a small chisel point angled toward the rake; this minimizes the area in which the material are forcefully pushed through. Horizontal chisel tips create hideous chips when they are pushed together out of the flutes. Drills with curved rakes provide improved material evacuation up the flute. Flutes should be longer than the hole to ensure that the entire circumference of the drill shaft does not brush against the hole. Full circumference contact generates heat when drilling foam and prevents the drill from going deeper than the flute length.

A clean surface polish on the drill flutes and tip enables the material to be evacuated more easily. Grind marks on the flutes of today's drills are likely to prevent bone chip evacuation, increasing heating and wear. Alternatives to stainless steel would result in drills that are far more resistant to wear. Cobalt chromium of medical-grade is frequently used in orthopedic implants and would be generally accepted if its performance exceeded that of stainless steel. Oral and maxillofacial, surgical burs in dentistry have been made from stainless steels coated with diamond ends, and dental burs have been fabricated by various types of metals, including stainless steel, tungsten carbide coated and uncoated with diamond ends. Alternatively, research has demonstrated that carbide can be used in specific orthopedic drills but has not been extensively adopted. It is unknown whether this is due to cost or drill fragility due to greater hardness over toughness.

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