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Review

Research progress in bone cutting technology for dental implant sites preparation: A review

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Abstract: With the increasing improvement of living standards and the popularization of dental implant restoration, dental implantation has become the preferred treatment for patients with missing teeth. The implant sites preparation is one of the most important procedures in dental implant surgery. The thermal and mechanical damage caused to the bone tissue during this process can directly affect the formation of osseointegration. To mitigate these adverse effects, many scholars have used methods such as optimizing cutting parameters and improving the structure of surgical tools to better control heat generation and cutting forces. At the same time, many new processing technologies such as milling, ultrasonic machining, and laser machining have also been explored for shaping implant site and have made some progress. This review aims to discuss the advantages and limitations of these techniques used in osteotomy, summarizes the current research status in 97 literatures of related fields.

Keywords: dental implant; implant site preparation; bone tissue cutting processes; thermal damage; mechanical damage

1. Introduction

Tooth loss impacts several key functions, including mastication, pronunciation, appearance and digestion, thereby causing inconvenience to patients and diminishing their quality of life. Since the theory of "osseointegration" was proposed in the 1950s, dental implant restoration has gradually become an important means of treating tooth loss [1], and dental implant restoration technology has also been continuously developed. The implant sites preparation is one of the most important procedures in dental implant surgery, as it directly affects the osseointegration of the implant and bone tissue during subsequent treatment, determining the success of the implant. At present, the commercialized implants are mainly with rotary cylindrical or conical shape, so the application of implant technology is based on the shaping of circular implant hole in the mandible or maxilla. The preparation process fundamentally involves the cutting and processing of bone tissue. Different from the milling and cutting performed in semi-closed spaces during orthopedic surgery [2], implant site preparation occurs in a fully enclosed space, with significant differences in the tool structure, cutting method, heat generation and heat transfer method, as well as composition and structure of bone tissue. Currently, cutting with drill bits is mostly used in clinical practice.

The quantity and bioactivity of the bone inside the osteotomy site are crucial for the formation of osseointegration and long-term stability of the implant. Many methods such as the guided bone regeneration (GBR) and socket preservation (SP) were used to provide a stable support for the implants [3]. However, during the site preparation and implant placement, the bone tissue inside the drilled cavity would be subjected to mechanical and thermal damage. Experimental studies have shown that although the rotation speed of the drill bit does not affect the initial stability of the implant, it can change the bone integration situation, as different rotation speeds cause different degrees of mechanical damage to bone cells [4]. When the implant site is formed, the thermal damage to the bone surface inside the osteotomy has a greater impact on osseointegration. Studies have shown that bone cells experience irreversible necrosis when exposed temperature exceeding 47 ℃ for more than 60 s, and osteonecrosis will lead to failure of osseointegration, failing implant restoration. In current clinical protocols for implant site preparation, the temperature inside the bone is mainly controlled by using a sequential set of drills increasing in diameter size and irrigation with normal saline, as well as cooling the drill by repeated lifting and copious amounts of irrigation. However, precise control is challenging to achieve, and there are cases of necrosis leading to inflammation of the bone tissue around the implant in clinical practice [5].

To mitigate the thermal and mechanical injury caused to bone tissue during the osteotomy for the implant site preparation, many scholars have conducted extensive research on the optimization of drilling process parameters and the geometry of drill bits, and proposed many feasible solutions, such as controlling the rotational speed and feed rate. In addition, some scholars have explored novel methods for implant site preparation, such as ultrasonic-assisted processing, laser ablation, and milling. These techniques show great potential in minimizing thermal and mechanical damage, but their application in dental implantology is still immature and mostly remains in the exploration stage.

This review summarizes the osteotomy methods for implant site preparation based on the anatomical structure characteristics and mechanical properties of the jaw bone, and analyzes the research priorities of thermal and mechanical injury control during osteotomy, providing a useful reference for clinical implant surgery research.

2. Bone tissue cutting theory for implant site preparation

Implant site preparation differs significantly from industrial machining due to the heterogeneous nature of bone tissue and variability of cutting forces during the process. Unlike the uniform material and stable cutting forces typical of mechanical processing, implant site preparation involves the removal of bone tissue, which is complicated by complex physiological structure of the jawbone. The anisotropic composition and mechanical properties of the bone result in intricate deformation during the cutting process is, unstable cutting forces, and challenges in maintaining precise temperature control. To ensure the geometric accuracy of implant site and to reduce mechanical and thermal damage to the bone tissue, it is necessary to conduct in-depth analysis from the structural characteristics and mechanical properties of the bone.

2.1. Anatomical structure of maxilla and mandible

The maxilla and mandible are the two largest bones in the maxillofacial region in terms of volume and area, with muscles and ligaments attached to them, as shown in **Figure 1a**. Teeth grow in the alveolar fossa forming an arc-shaped arrangement of upper and lower dental arches. Bone tissue is composed of cortical bone with a high density in the superficial layer and cancellous bone with porous structures in the inner part. Cortical bone consists of bone cells and interstitial bone plates, which have high mechanical strength and play a major supporting role in bearing mechanical behaviors, and cancellous bone is composed of a large number of interwoven bone trabeculae, which form a network-like structure that can effectively distribute stress and absorb impact [6], as shown in **Figure 1b**.

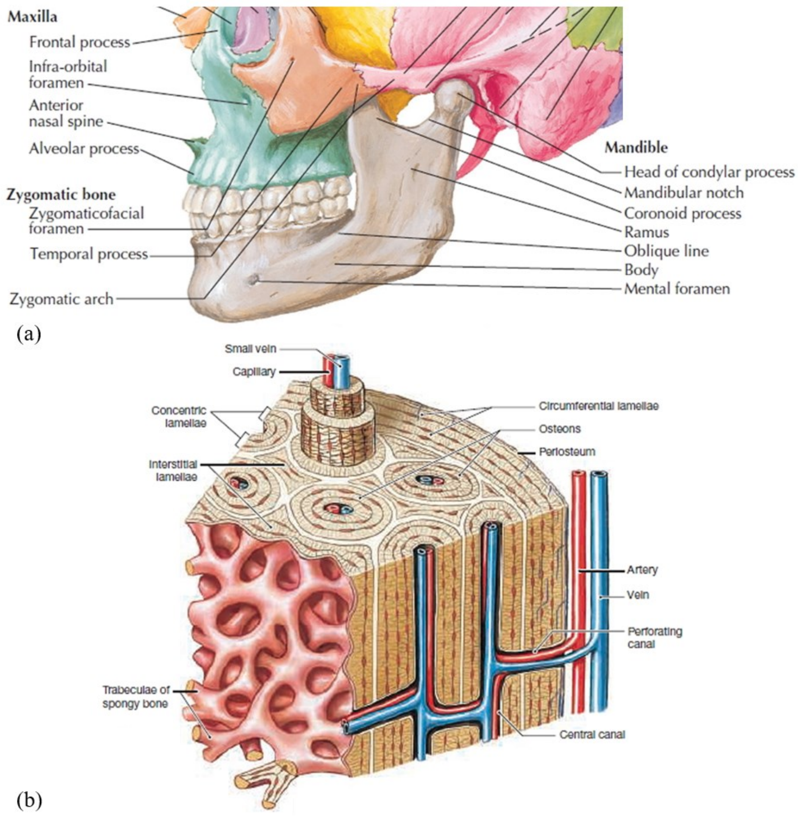

Figure 1. Anatomical structure of maxilla and mandible. **(a)** Maxilla and Mandible; **(b)** The structure of cortical and cancellous bone [7].

Due to the complex structure and composition of bone tissue, its mechanical properties are significantly different from those of uniform metal materials commonly used in mechanical processing, exhibiting complex anisotropic characteristics. At scales ranging from 1 to 10mm, the strength and tensile/compressive modulus of cortical bone are greater along the longitudinal direction (aligned with the bone axis) compared to the radial and circumferential directions. The brittleness of cortical bone under transverse loading is greater than that under longitudinal loading [8], and the fracture toughness under shear loading is significantly greater than that under tensile loading [9], which reveals the arrangement direction of bone cells has a significant impact on the strength of cortical bone [10]. Similar to porous materials, the mechanical properties of cancellous bone (5mm–10mm in size) are significantly influenced by the porosity, trabecular structure, and density, so the cancellous bone owns greater compressive strength, lower tensile strength, and lowest shear strength [11].

2.2. Influence of heterogeneous material properties of jawbone tissue on cutting performance

The anisotropic characteristics of bone tissue are also reflected in the cutting characteristics, which poses a great challenge for the study of cutting mechanisms and process control. As early as the 1970s, Jacobs pointed out that the differences between bone material cutting and metal cutting, emphasizing the directionality of bone cell arrangement necessitates consideration of parallel, cross, and perpendicular cutting directions, and an orthogonal cutting model for bone materials is constructed to optimize cutting parameters [12]. At the same time, other scholars also conducted related research, including analyzing the fracture patterns of bone materials by observing the chip morphology, and constructing models to describe the formation mechanism of chips [13]. After decades of development, the research on the cutting mechanism of bone tissue has gradually deepened. Liao et al. found that during orthogonal bone cutting, as the cutting thickness increases, the cutting mode transitions among shear cutting, shear crack cutting, and fracture cutting, and a cutting model was constructed based on fracture mechanics [14]. In addition, they concluded through surface morphology analysis that the material damage mode is related to the cutting method and direction. Feldmann et al. [15] found that during the cutting process, as the cutting depth increases, there is a transition from plastic cutting to two modes of fracture cutting, and a method for calculating the fracture toughness of quasi-brittle materials using orthogonal experiments was proposed. Research by Luo et al. [16] showed that the cutting angle does not significantly affect the cutting behavior at small cutting depths (<50 microns), but has a greater impact when the cutting depth is greater than 130 microns, and so they constructed a fracture toughness calculation model for large cutting depths.

Bone density is also one of the important factors affecting the cutting process. With in vitro experiments, many scholars have proven that bone density can significantly affect cutting force [17,18]. Chen et al. [19] constructed a three-dimensional dynamic finite element model to obtain the temperature distribution during the osteotomy to explore the changes in the thermal damage area with varying cutting parameters under different bone qualities, their results showed that the temperature of the bone wall increases with the increase in cortical bone thickness and bone density (**Figure 2**), with the highest temperature point located at the junction of cortical and cancellous bone. Sezek et al. [20] found that under certain conditions, a 12% increase in bone density can lead to a 10% increase in cutting temperature.

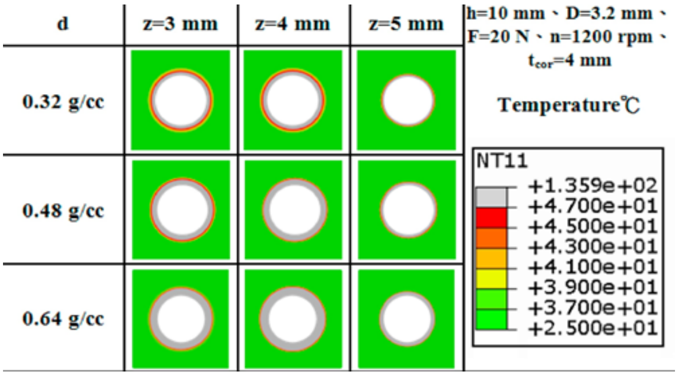

Figure 2. Contour lines of bone temperature distribution at depths of 3, 4, and 5 mm for different bone densities (d = 0.32, 0.48, and 0.64 g/cc) [19].

The nonuniform properties of bone tissue on material composition and structure, as well as special cutting characteristics, pose great challenges for the study of cutting mechanisms. After decades of exploration, many advances have been made in the study of the mechanical and cutting properties of bone, but there are some problems to be solved such as the transient changing cut force and the vibration causing hard control to the accuracy and stability of the operation. During the process of forming dental implant site, the narrow and relatively enclosed operation space in oral cavity makes the surgical environment particularly complex. On the other hand, the bone conditions are personalized, and even for one patient, the bone properties are also various at the different implant sites, which makes the analysis and control of the cutting process particularly difficult. Therefore, further research on the cutting theory of bone materials is still required.

3. Implant site preparation with drilling

Till now, the osteotomy with drill bur is still the main method for dental implant site preparation in clinical dental implant surgery. Due to the closed implant site preparation environment and the poor thermal conductivity of bone tissue, heat is generated within the bone, which can cause thermal necrosis of bone tissue, and leads to fibrous tissue intervening in the implant-bone interface, therefore affecting the bone integration process and reduces implant survival rate [21]. Clinically, in order to control the temperature of the bone tissue on the implant bed surface during bone cutting, a sequential of drills with a diameter increasing from small to large is used for osteotomy step by step to reduce the thickness of cutting and control the generation of heat. During the osteotomy process, the drill is repeatedly lifted out and cooled by irrigation (**Figure 3**).

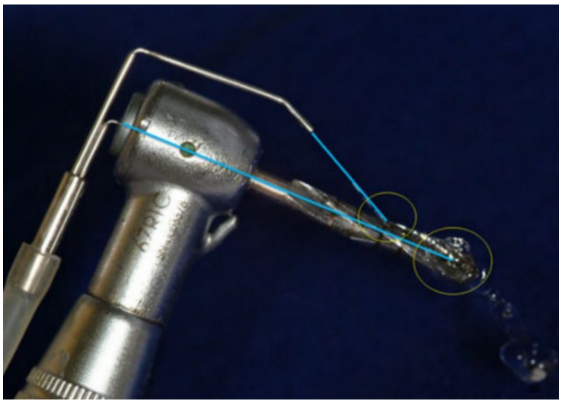

Figure 3. Implant site preparation with cooling water irrigation [22].

3.1. The influence of drilling parameters on the bone temperature

Birkenfeld et al. [23] exposed the alveolar bones to 40 ℃, 50 ℃, 60 ℃, and 100 ℃ for 1 min and then scanned them under scanning electron microscopy (SEM), they found that there were no significant changes in the alveolar bone at 40 ℃ and 50 ℃, but at 60 ℃ and 100 ℃, the bone surface became rough and micropores appeared. Tahereh et al. [24] studied the effect of using different rotational speeds and feed rates to prepare implant sites on the stability of implants through animal experiments, after measuring the implant stability quotient (ISQ) at multiple stages, they found that the rotational speed and feed rate had no significant effect on implant initial stability, increasing the speed and decreasing the feed rate both helped to improve the long-term stability of the implant.

Scholars have conducted many researches on drilling for bone materials, aimed to solve the problems of drill structure damage, thermal damage and microcracks of bone materials, analyzing the impact of drill material, structure, and cutting process parameters on the trend of temperature changes and cutting forces, and proposing optimization solutions. Li et al. [25] designed two types of drill bits using superhard materials, and found that at a speed of 500r/min, the temperature of the bone tissue was 33.5 ℃ with a brazed stepped drill compared to 42.9 ℃ with a brazed twisted drill, proving that both the shape and material of the drill bit affect the temperature rise. Lee et al. [26] pointed out that a vertex angle of 70°–90° or 120°–140°and a back angle of 12°–15°on the bone drill can reduce the friction generated by cutting, and also suggested using a helix angle of 24°–36°to reduce bone debris blockage and using coolant to remove debris and cooling down. Akhbar [27] evaluated the impact of the height and width of the drill bit edge on the thermomechanical damage (maximum bone temperature, osteonecrosis diameter, osteonecrosis depth, maximum thrust and torque) of bone tissue using finite element method, and obtained the optimal range of drill bit edge height and width for controlling thermomechanical damage. Shu et al. [28] proposed a three-stage drill structure (**Figure 4**) that can achieve a change in cutting mode, significantly reducing cutting force, temperature, and bone damage compared to conventional drill structures. Liu et al. [29] designed a crescent-shaped drill bit with optimized tool rake angle, which has shown experimentally to lower cutting force and control the temperature within a threshold without irrigation. Chen et al. [30] designed a planting drill suitable for low-speed cutting, which generates less heat during the cutting process and does not require cooling water irrigation. Matthews et al. [31] found that the temperature of bone tissue increases with the increase in the number of times the drill is used, their experiment found that when the number of times the drill is used exceeds 40 times, the drill is severely worn and continued cutting will lead to a significant increase in the temperature of bone tissue. Therefore, drill wear is also an important factor affecting temperature rise. The experimental results of Karaca et al. [32] showed that the drill coating also had an impact on the temperature, and the temperature of the TiBN coated drill was significantly higher than that of the uncoated drill.

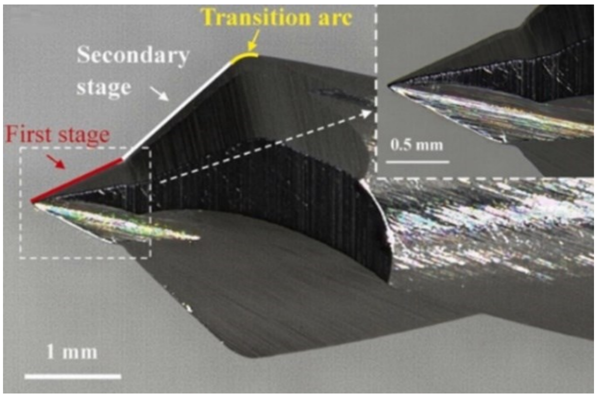

Figure 4. Micro-textured medical drill bit [28].

The process parameters of drilling, such as feed rate, rotational speed, and the use of cooling water, also have a significant impact on both cutting heat and cutting force. Reingewirtz et al. [33] studied the effects of various parameters on bone heat production and time consuming during the osteotomy in vitro bovine femurs, as shown in **Table 1**. Salomó-Coll et al. [34] used artificial bone blocks made of polyurethane material and specially designed thermocouple temperature measurement devices to evaluate the different effects of drill diameter, cutting speed, bone density, and cooling water on heat generation during the preparation of implant site in artificial bone blocks, and they found that differences in bone density were one of the main factors causing changes in body temperature. When bone density was different, changes in drill diameter and drill speed had different effects on heat generation. The use of cooling water can significantly reduce the heat generated by the high-speed rotation of the drill bit during cutting, so using cooling water at higher speeds and not using it at lower speeds can avoid excessive heat generation [34]. Song et al. [35] explored the effects of oblique angle, rotational speed, and feed rate of the Kirschner wire on drilling force and temperature through simulation and experiment, and they found that an increase in oblique angle and feed rate would lead to an increase in cutting force. An increase in rotational speed helps reduce the force; In addition, the increase of bevel angle, rotational speed and feed rate will lead to the increase of maximum temperature [35].

Table 1. The influence of different parameters on bone heating and drilling duration [33].

Variable	**Duration**	**Temperature**	**Rotational speed**
Motor type	=	=	
The power of the contra-angle decreases ↑	↑↑	↓	Low speed 400–800rpm 400–800rpm
Quality ↑	↓↓	↑	
Pre-drilling depth↑	↓	=	
Speed ↑	↓↓↓	↑↑, ↓ then steady	High speed 24,000–40,000rpm 24,000–40,000rpm
Coolant	=	↓	

3.2. Temperature measurement and simulation on drilling of implant site

In order to achieve further temperature control, it is necessary to obtain real-time temperature data around the implant site. However, it is currently difficult to achieve real-time measurement of the temperature inside the borehole [36]. Many researches have been conducted to measure the temperature data inside the bone. Finite element simulation and in vitro experimental measurement are commonly used methods currently. In vitro experiments typically use two main measurement methods. One is to embed thermocouples in the bone tissue around the drill hole, but due to space constraints, the number of embedded thermocouples cannot be too large. This method can only generate temperature data for some locations around the drill hole, and the nonlinear relationship between thermocouple signals and temperature may affect the accuracy of the results. In addition, thermocouples measure the temperature at a specific point and cannot provide thermal profiles or distributions. The other method is to measure the temperature around the borehole with a thermal imager [37]. Since thermal imaging devices can only measure the temperature of the surface of cutting position, which is different from the inside temperature.

The shaping process of the implant site is the cutting process of the bone tissue. The generation and conduction of cutting heat are the fundamental causes of temperature changes, therefore, the heat conduction equation of metal cutting process is the basis for theoretical research on the temperature field of bone tissue [38]. Experimental research has found that when using a static guide plate to guide the shaping of the implant site, due to the obstruction of cooling water by the guide plate, it is a good approach to pull the drill needle multiple times to fully cool and reduce the temperature of the bone tissue in order to improve the cooling effect [5]. Feldmann et al. [39] proposed an analytical temperature prediction model that uses torque signals during drilling to simulate the heat generation of the drill bit. Additionally, extensive experimental research was conducted by using a customized CNC device which has a force sensor and thermal imager to measure axial torque and force as well as temperature rise to validate the model. (as shown in **Figure 5**) [39].

In order to simulate the complete temperature distribution around the drill site, a large number of scholars have conducted finite element analysis to investigate the temperature field distribution during the implant site preparation. Sezek et al. [20] used finite element simulation to model bone as a hollow cylinder and set it as an isotropic material, analyzing the temperature changes during cortical bone preparation under different parameters such as rotational speed, feed rate, drill diameter, drill force, bone density, and bone structure (as shown in **Figure 6**). The simulation results were validated by combining with in vitro control experiments using fresh calf cortical bone. Cui et al. [40] established a theoretical model of drilling process based on heat conduction theory, including full-hole drilling and half-hole drilling, and revised and verified the model through in vitro experiments, and an improved theoretical model for full-hole preparation was established (**Figure 7**), and the accuracy of the model was verified through experimental data of full-hole drilling.

However, due to the complexity of the actual implant site preparation and the differences between the simulation model and the actual process, the temperature distribution around the drill hole obtained by the simulation model also deviates from the actual drilling process. The experimental measurement of the temperature inside the bone tissue during osteotomy could be only conducted in vitro model through pre-embedded thermocouples, and the on-situ temperature measurement only could be done on bone surface during surgical operation. So, the temperature inside the bone tissue during clinical operation could not be reached till now, which is a significant challenge for controlling bone temperature during dental implant osteotomy.

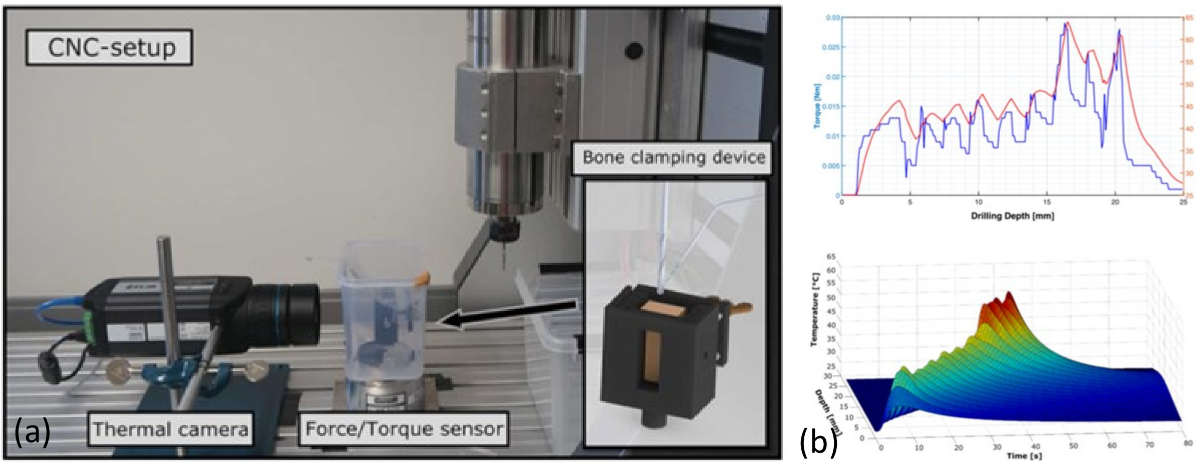

Figure 5. Measuring device and results [39]. **(a)** The three-axis CNC device used in the experiment; **(b)** Measuring torque and maximum temperature rise with drilling depth.

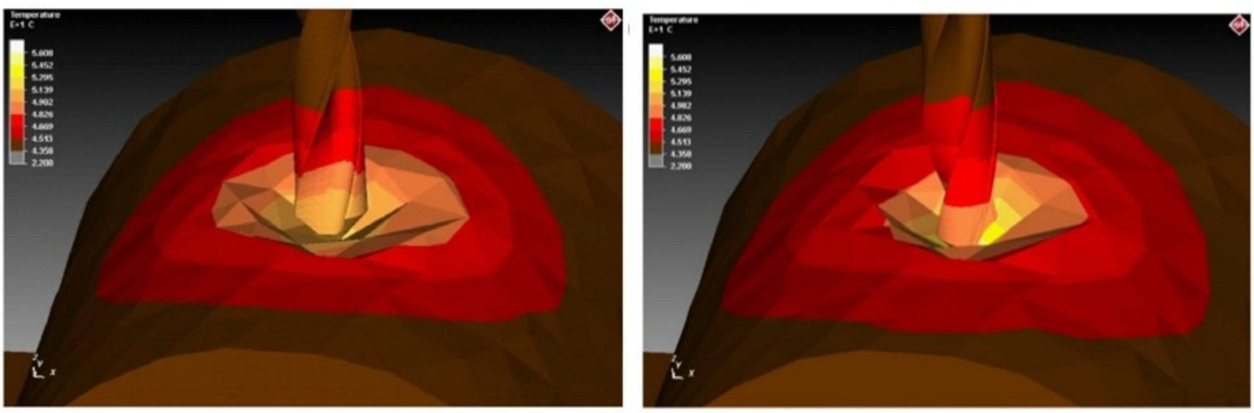

Figure 6. Finite element simulation of temperature field during drilling at various depths [20].

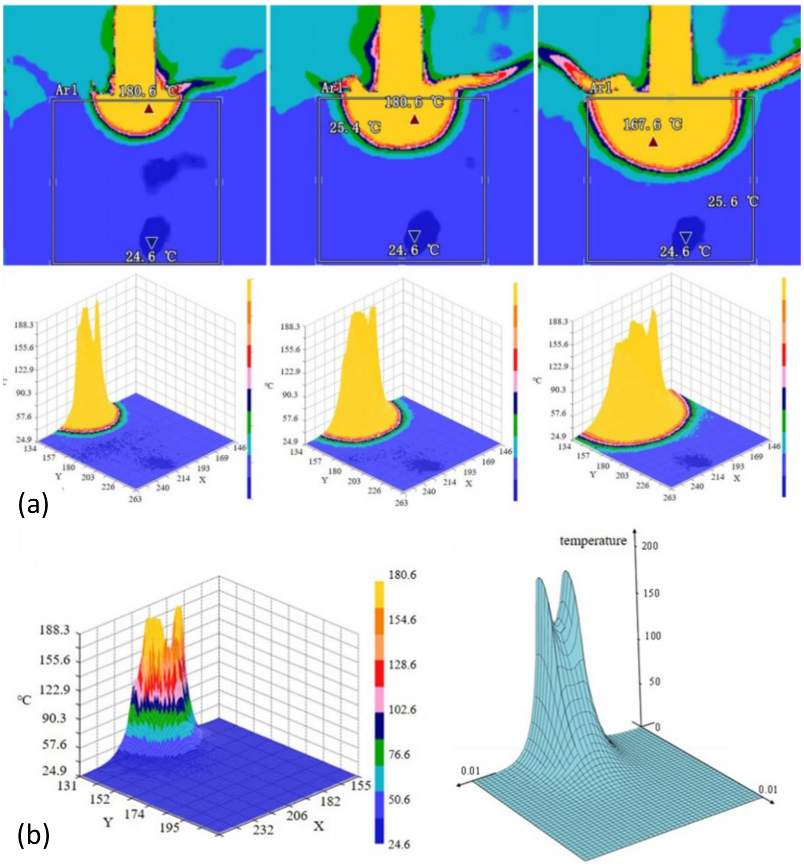

Figure 7. Temperature Field around the implant site [40]. **(a)** Temperature distribution of images at different times; **(b)** Temperature distribution of bone section during half-hole drilling.

As a commonly used technique for implant site preparation method in clinical surgery, drilling has been deeply studied for many years. In order to achieve temperature and cutting force control during the process, scholars have tried a large number of methods, such as changing cutting speed, using cooling water [34], improving cooling pattern [41], optimizing the structure of the drill bit and its surface treatment method [25], ultrasonic-assisted technology [37], etc. Many of these methods are applicable to dental implants or can promote the further improvement of dental implant site formation technology.

4. Implant site preparation by milling

In recent years, the spiral milling technology has been gradually applied in the aerospace field for machining composite materials, titanium alloys and other difficult-to-process materials. It uses special tools to achieve circular hole machining through eccentric milling. Many scholars have conducted comparative studies on spiral milling and drilling technology, and believed that spiral milling technology has higher machining accuracy, lower cutting temperature, and less material damage [42–44].

Considering that the biological composite material characteristics of bone tissue are similar to some difficult-to-machining materials in aviation, and that there are high requirements for the accuracy and quality of implant site preparation, large hole depth-diameter ratio, difficulty in discharging bone debris, high temperature control requirements for bone tissue in implant site, and complex step-by-step drilling processes, combining robot trajectory planning and control technology with industrial spiral milling technology of hole is expected to achieve a robot-based implant site spiral milling preparation process. As a traditional precision machining method, milling has been increasingly widely used in bone tissue cutting for orthopedic surgery in recent years, but it has been less applied in the hole formation in bone tissue.

4.1. The influence of bone tissue milling parameters on cutting force and cutting temperature

The cutting force and cutting temperature during the milling process of bone tissue are important factors in the surgical process. Over the years, scholars have conducted in-depth discussions on this and believe that optimizing milling parameters such as rotational speed, feed rate, and cutting depth is an effective means of controlling milling force and temperature. Al-Abdollah et al. [45] used artificial neural networks to construct a relationship model among milling force, temperature, feed rate, and spindle rotational speed for milling cancellous bone, and found that the average milling temperature increases with increasing feed rate and decreases with increasing spindle rotational speed. In addition, they also built a network to optimize the feed rate and spindle rotational speed during robotic milling, balance the milling force and average temperature during the process, and reduce bone tissue damage [46]. Chen et al. [47] proposed a cortical bone milling force analysis model based on the orthogonal cutting distribution method, which considers the influence of bone tissue anisotropy and improves the accuracy of cutting force prediction during milling in different processing directions. In addition, they also explored the temperature diffusion mechanism of high-speed bone milling, and believed that the main source of heat in high-speed bone milling was the friction between bone debris and the teeth of the mill cutter, and that spray cooling could significantly reduce the bone temperature during milling [48]. Tahmasbi et al. [49] used response surface analysis to study the influence of cutting depth, feed rate, mill rotation speed, milling direction, and tool diameter on milling force, and pointed out that factors positively correlated with milling force include feed rate, cutting depth, and tool diameter. Increasing the tool rotation speed will reduce the cutting force, while milling perpendicular to the direction of the bone unit will result in lower milling force than parallel milling. Liao et al. [50] constructed a cutting force model that considers the anisotropy of bone materials, and established a temperature field model based on heat flow, as well as using the depth of cell damage that may be caused by thermal effects to assess the degree of thermal damage to bone tissue.

In addition to improving milling parameters, optimizing the structure of bone milling tools is also an important method for minimizing bone damage. Sugita et al. [51] proposed a concept of bone tissue processing based on micro-cutting characteristics and improved the design of bone milling tools, which are equipped with serrations and straight edges, as shown in **Figure 8a**. At higher feed rates, the cutting force is reduced by 40%, which can improve processing efficiency and reduce cutting temperature. Liao et al. [52] designed a new type of composite milling cutter (**Figure 8b**), which has micro-cutting edges arranged on the back of the main cutting edge. The cutting mode can be changed by adjusting the feed rate. At the same time, the tool is made of solid diamond composite material, which has good thermal conductivity. Experiments showed that this design can effectively reduce bone tissue fracture damage and cutting temperature when compared to conventional milling cutters. Hu et al. [53] found that the use of micro-texture milling cutter (**Figure 8c**) results in lower milling force and smaller and narrower micro-cracks on the surface of bone tissue, thereby reducing bone damage. At the same time, micro-texture cutting tools have better hydrophilicity compared to conventional milling cutters.

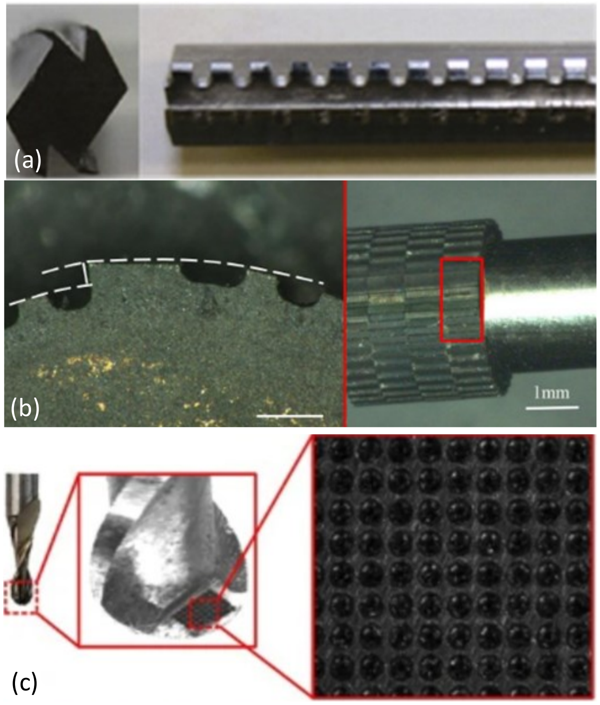

Figure 8. Novel bone milling tools [51–53]. **(a)** Milling cutter with serrations and straight edges; **(b)** Milling cutter with micro-cutting edges; **(c)** micro-texture milling cutter.

4.2. Robot-assited milling for bone tissue

Due to the high stability and accuracy, robots can better meet the requirements of milling parameter control. As early as 2003, Federspil et al. [54] attempted to apply robotic milling to ear neurosurgery. Since then, with the continuous deepening of research, robotic bone tissue milling has gained a lot of research and clinical applications. Chen et al. [47] considered the low stiffness of the robot in their milling force model, and further derived a robot milling force prediction model to meet the application scenarios of the robot, and verified the accuracy through experiments. Some scholars have conducted research on the control of robotic bone milling force, analyzing the robotic milling surgery process and the milling mechanism of bone materials, constructing a milling force prediction method, and achieving cutting force control by controlling the robotic feed rate [55,56]. Xia et al. [57] established a milling temperature prediction model for cancellous bone of vertebral lamina using neural networks, which considers the effects of bone density and milling parameters on milling temperature, to help selecting appropriate milling parameters when milling spinal lamina.

At present, in clinical dental implant surgery, clinicians mainly operate handpiece for implant site preparation by free-hand, which is affected by many external factors, resulting in challenges in the quality of implant sites and thermal damage to bone tissue. However, the preparation of implant sites through bone milling is still in the early research stage and has not been widely used in clinical applications. With the development of medical robot technology, the advantages of minimally invasive, high precision, trajectory operation diversity, and real-time feedback of robots, combined with the advantages of spiral milling in processing accuracy, efficiency, temperature control, and applicability, will surely form a new type of implant site preparation technology, and with which, further optimizing the tool structure and process parameters will greatly promote the automation and precision of dental implant site formation in the clinical field.

5. Application of ultrasound in implant site preparation

Ultrasonic wave is a mechanical wave with extremely short wavelength, generally shorter than 2 cm in air, which must rely on a medium for propagation. Due to the interaction between ultrasonic wave and medium, the medium changes will result in mechanical effects and cavitation effects. Ultrasonic machining is very mature in the industrial field. For example, by using small amplitude ultrasonic vibration tools, materials that are difficult to machine can be gradually abraded through a specialized process, achieving ultrasonic polishing on the surface of the workpiece. In the medical field, ultrasonic machining is mainly used for cutting hard tissue, which can avoid damage to soft tissues such as blood vessels.

5.1. Clinical application of ultrasonic osteotome in implant site preparation

The ultrasonic osteotome or named piezosurgery is a medical device that uses high-intensity focused ultrasound technology for bone surgery, with work principle of converting electrical energy into mechanical energy using a transducer, and achieving bone tissue cut through high-frequency ultrasonic oscillation [58]. The ultrasonic osteotome (**Figure 9**) has high accuracy in bone cutting, with a horizontal swing amplitude of about 60μm–200μm and a vertical swing amplitude of 20μm–60μm during operation. At operating vibration frequency (24 kHz to 29 kHz), the piezosurgery will only cut mineralized tissues with high acoustic impedance (such as bone), while soft tissues (such as blood vessels and nerves) need to be cut at frequencies above 50 kHz. Therefore, the piezosurgery can avoid damage to blood vessels and nerves during bone cutting surgery [59]. At the same time, the water and air flow during the piezosurgery operation and the small amount of bone cutting help to control the temperature at the bone cutting site, reducing both thermal and mechanical damage to the bone [60].

Figure 9. Ultrasonic osteotome [59].

Some recent studies have shown that using piezosurgery for implant site preparation can provide initial stability similar to conventional methods and better long-term stability for implants [61,62]. Rebaud et al. [63] compared the bone damage caused by piezosurgery and drilling techniques using in vitro experiments, and SEM scanning showed that the piezosurgery group caused less bone damage and had better bone surface quality than conventional process. The results of Soheilifar et al. [64] showed that ultrasonic implant preparation can achieve higher ISQ values 3 months post-surgery, with a shorter inflammatory period, potentially creating better conditions for immediate implantation. Alberti et al. [65] used piezosurgery and drilling to complete the formation of implant site in 40 patients, and the postsurgical measurements of bone density at different time stages showed that the piezosurgery group had higher bone density around the implants, which was also more conducive to bone healing. Mozzati et al. [66] tried to apply piezosurgery to zygomatic implant, and found that this method can achieve better visibility of the surgical area, and the implant survival rate and postoperative complication rate are similar to conventional method, making it a feasible alternative to drilling for zygomatic implant.

5.2. Ultrasonic-assisted drilling for implant site preparation

Ultrasonic-assisted drilling is also a promising method for implant site preparation, which is a vibration machining technology removing material by superimposing high-frequency vibrations during the drilling process. It is mainly divided into three types according to the different vibration directions: axial vibration drilling, torsional vibration drilling, and axial torsional compound vibration drilling, as shown in **Figure 10** [67]. Compared to conventional methods, ultrasonic-assisted machining can effectively reduce the cutting force and temperature during the cutting process of bone materials, thereby reducing thermal and mechanical damage of the bone tissue [2]. Alam et al. [68] conducted experiments on bovine cortical bone and compared the axial force and torque of different drilling methods, and their results showed that the axial force and torque of ultrasonic-assisted drilling were significantly lower than those of conventional drilling, and it had better effects on tissue debris removal from drill. Gupta et al. [69] conducted in-vitro experiments on pig bones using both conventional and ultrasonic-assisted drilling methods, and the experimental results showed that the ultrasonic-assisted method produced fewer and smaller microcracks on the inner surface of the implant site; higher rotational speed and lower feed rate helped to further control the generation of microcracks, while ultrasonic-assisted surgery also provided better stability for the implant. Sing et al. [70] established a model for the cutting force and torque of ultrasonic-assisted bone drilling, and found that the cutting force and torque decrease with increasing rotational speed, and increase with increasing drill diameter and feed rate.

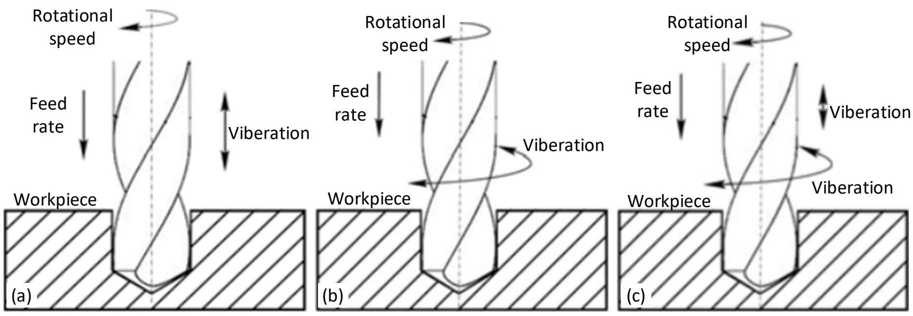

Figure 10. Schematic diagram of ultrasonic-assisted drilling [67]. **(a)** Axial vibration drilling; **(b)** Torsional vibration drilling; **(c)** Axial torsional compound vibration drilling.

Shakour et al. [71] conducted an in vitro experiment to study the process parameters of ultrasonic vibration-assisted drilling, showing that at a drill speed of 1000 r/min, both cutting force and temperature were lower than conventional drilling. The research results of Agarwal et al. [72] also reached similar conclusions, and found that ultrasonic-assisted osteotomy helps to control the size of bone debris, thereby promoting the discharge of bone debris. However, some scholars' research shows that under some conditions, ultrasonic-assisted drilling can result in higher temperatures [73,74], suggesting that further exploration of process parameters is necessary. In order to suppress the temperature rise during ultrasonic-assisted surgery, Gupta et al. [75] studied the effects of rotation speed, feed rate, drill diameter, and vibration amplitude on temperature changes, and established a statistical model for predicting the maximum temperature at the interface between the drill and the bone tissue, and they found that the temperature increases with the increase of rotation speed, feed rate, and drill diameter, and decreases with the increase of vibration amplitude. In addition, some scholars have used machine learning methods to construct a thermal injury prediction model, which was validated through in vitro experiments with an error of ± 1.7 ℃ [76].

Regarding the surface quality of the implant site, Zhang et al. [77] compared the surface quality of cortical bone drilled with ultrasonic vibration-assisted drilling and conventional drilling, and their results showed that ultrasonic vibration-assisted method achieved better surface quality, and the surface roughness of the drilled hole was closely related to the cutting parameters and vibration parameters, as shown in **Figure 11**. Yan et al. [78] found that using tricone drill ultrasonic vibration assisted bone cutting can effectively improve the poor centering ability of conventional bone drilling, with the lower force, temperature, and surface roughness.

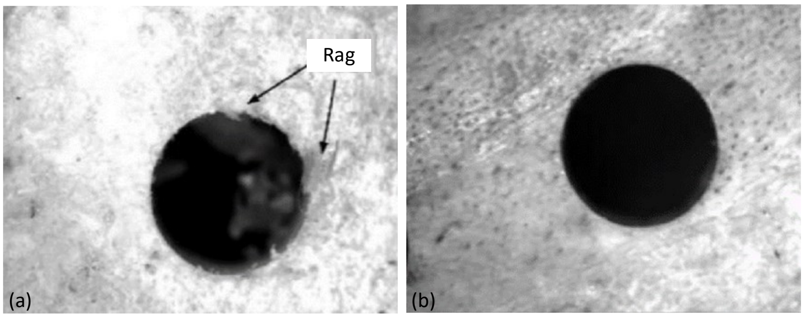

Figure 11. Comparison of surface quality of implant sockets with different drilling methods [77]. **(a)** Conventional drilling; **(b)** Ultrasonic-assisted drilling.

Overall, the application of ultrasonic osteotomes has been widely used in dental and maxillofacial surgery, such as maxillary sinus elevation [79], impacted wisdom teeth extraction [80], and autogenous bone grafting [81]. However, there are currently few studies on its application in implant site formation, mainly focusing on the comparison of clinical effects between piezosurgery and conventional drilling methods for implant site formation. The large-scale application of this technique still requires further discussion. Ultrasonic-assisted drilling technology can help reduce the cutting force during the formation of implant sites and reduce the microcracks. However, due to the lack of specialized surgical instruments for dental implanting, most of the related technologies are still in the theoretical research or in vitro experimental stage. The clinical application in the field of dental implanting needs theoretical breakthroughs and instrument development, but it is expected to become a new implant site preparation technology.

6. Laser ablation technology for implant site preparation

Laser processing technology, especially laser drilling, has been widely and maturely applied in industry. The advantages of laser drilling are that it is a non-contact process, does not generate cutting forces, and can control and reduce the heat affected zone through high energy concentration. In the medical field, using laser as the energy source to act on biological tissues and achieve the removal of biological tissues is called “tissue ablation” [82]. Laser action on biological tissues involves multiple forms of influence, as shown in **Figure 12**. Divided by the energy density and time of action, it can be divided into photochemical interaction, thermal interaction, photothermal ablation, and plasma-induced ablation [83]. Except for photochemical interactions, the other forms can achieve bone tissue ablation.

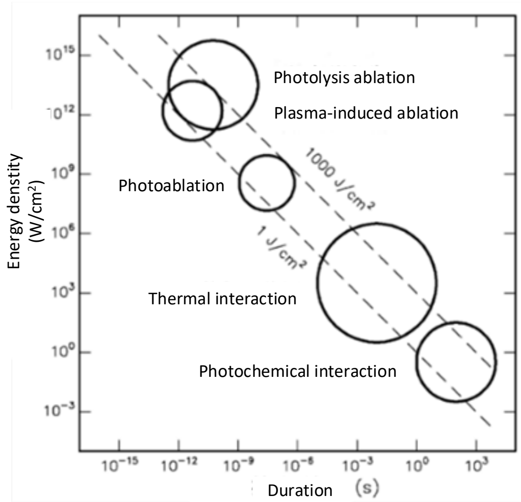

Figure 12. The acting forms of laser on biological tissue [83].

6.1. Interaction mechanism between laser and bone tissue

In recent years, laser technology for implant site preparation has gradually become a research hotspot, but there is still difficult to conduct a completely laser system of implant site preparation. El-montaser et al. [84] began using Er:YAG laser in 1999 to prepare holes with a diameter of about 0.7 mm in the rat skull, and studied the bone integration of titanium screws in the holes. In recent years, experiments on animal or human for laser shaping implant sites have also been conducted on a trial basis. Schwarz et al. [85] used a 2940 nm pulsed infrared Er:YAG laser device and a conventional drilling technique to form implant sites in the mandible of four beagle dogs, as shown in **Figure 13a**, and observed that neither the Er:YAG laser group nor the drilling group caused thermal side effects such as carbonization, melting, or cracking of the adjacent alveolar bone. Swider et al. [86] used Er:YAG laser (Lite Touch®) to try to shape the implant site at tooth position 35, as shown in **Figure 13b**, proved that Er:YAG laser has the potential to replace drilling protocols for shaping implant sites.

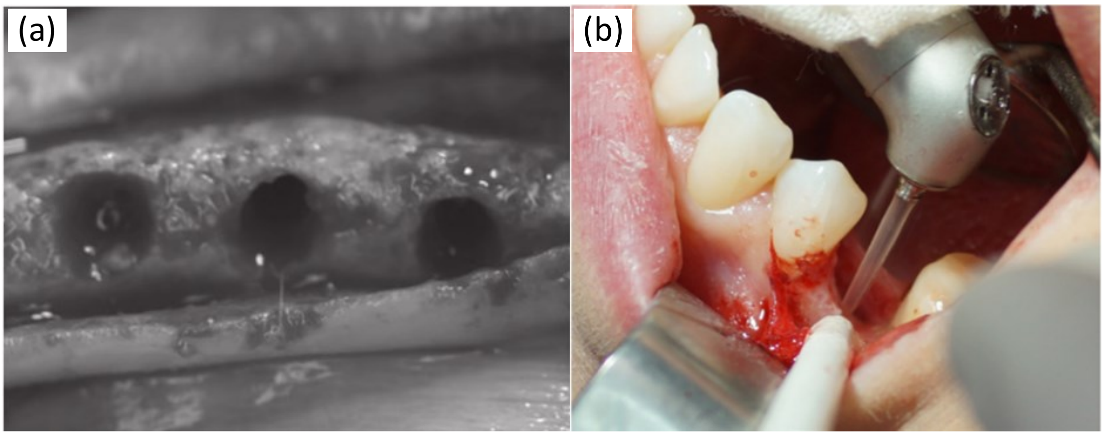

Figure 13. Laser forming of implant sites in animal and human jaw bones. **(a)** Laser-formed implant site on dog [85]; **(b)** Laser-formed implant site on human's mandible [86].

Existing research suggests that the process of laser ablation of bone tissue is dominated by thermal interactions, with mechanisms including vaporization ablation and micro-explosion ablation. Vaporization ablation refers to the process of laser energy heating the bone tissue to a vaporization temperature and then removing and eliminating the bone tissue [87]. Micro-explosion ablation refers to the process of removing tissue by micro-explosion due to the different boiling points of components within bone tissue, as shown in **Figure 14a**. During the heating process, substances with low-boiling vaporize first, causing local pressure to increase, which in turn causes micro-explosions in some tissues, as shown in **Figure 14b** [88,89].

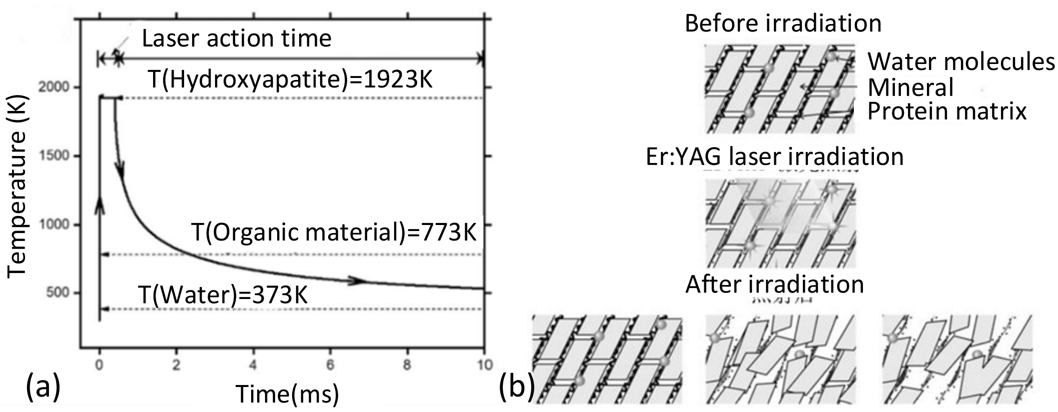

Figure 14. Laser actions of vaporization and micro-explosion. **(a)** The vaporization temperatures of water, organic matter, and hydroxyapatite [87]; **(b)** Schematic diagram of micro-explosion [88].

6.2. The influence of laser parameters on the ablation effect of bone tissue

The influence of laser parameters on ablation effect is the research focus on laser ablation of bone tissue, which determines the degree of bone damage. When shaping the implant site, choosing larger laser pulse width, energy density, and power density can enhance the ablation effect on the bone tissue. However, it is necessary to adjust the laser parameters to ensure that the heat within bone tissue remains below the thermal damage threshold, preventing damage and carbonization, which can affect the subsequent osseointegration of the implant and bone. On the other hand, when using different types of laser sources, the ablation effect varies due to the correlation between the laser wavelength and the absorption peak of the components within the bone tissue. The main components of bone tissue include hydroxyapatite, collagen, and water. The absorption peaks of hydroxyapatite are within 9μm–11μm, 6μm–7μm for collagen, and around 3μm for water [90]. In theory, laser with these wavelengths has good effect on ablation of bone tissue. In practical applications, the most studied laser types for bone tissue are Er:YAG (wavelength 2.94μm) and Er,Cr:YSGG (wavelength 2.78μm) lasers, which achieve precise cutting of bone tissue through explosive vaporization [91]. The short-pulse CO_2 laser with 9.6μm wavelength is also located near the absorption peak of bone tissue, when Eyrich et al. [92] used this laser to treat bone tissue, the temperature only increased by 1.88 ℃, which is lower than the 3.3 ℃ temperature rise caused by using Er:YAG laser. Due to cost, efficiency, and tissue damage, lasers with other wavelength bands are less commonly used for bone tissue ablation, such as free-electron lasers (wavelength 6.1μm) [93] and femtosecond lasers (wavelength 1.03μm) [94].

When using conventional drilling technique, the shape of the implant site is determined by the shape of the drill bit itself. During the laser ablation process, the extent of bone tissue removal is determined by the combination of the laser trajectory and laser parameters. Taking a single pulse laser as an example, the laser is a Gaussian beam, and the energy distribution in the plane follows a Gaussian distribution. Therefore, the beam diameter increases with the increase of the distance from the laser focusing plane, and the laser beam energy decreases with the increase of the distance from the focal point. Therefore, it is necessary to determine the focal position of the laser beam focus reasonably according to the energy range required for tissue ablation and the requirement for reducing thermal damage. As shown in **Figure 15**, if the focal point of the laser is too close or too far from the cutting target, the laser may only cause heating effects rather than effective ablation [95]. During the laser-assisted implant site forming process, the ablation area is formed by the superimposing of the ablation areas from individual pulses. Therefore, the superimposing pattern of the pulses also affects the ablation outcome. Pantawane et al. [88] used multiple partially overlapping pulses and laser trajectories, with a continuous laser trajectory center spacing of 0.3 mm, to perform three-dimensional cutting of bone tissue, achieving good tissue removal effects.

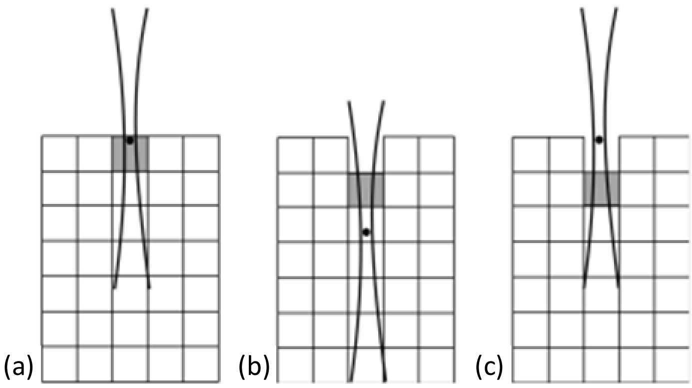

Figure 15. Schematic diagram of laser focus position [95]. **(a)** The focus is on the bone surface; **(b)** The focus is inside the bone; **(c)** Focus above the bone surface.

Laser parameters that more effectively ablate bone tissue represent easier access to or even exceeding the thermal damage threshold of bone tissue, thus balancing ablation efficiency and reducing thermal damage has gradually become an important problem needed to solve. During the laser processing of metal, cooling water is required to ensure processing efficiency, and similarly, during tissue ablation, cooling method also play a role in enhancing efficiency. When laser light is incident on a liquid medium such as water, the medium absorbs most of the energy and forms cavitation bubbles, which is called cavitation effect. During the formation and collapse of cavitation bubbles, shock waves and refractive flows are generated in the surrounding area, which can be used to influence the tissue ablation process by regulating these shock waves and refractive flows [96]. Kang et al. [97] used Er,Cr:YSGG laser (wavelength 2.79) to compare the ablation of bovine tibia under dry, wet (using water or perfluorocarbon), and spray conditions, and found that under the influence of environmental water coverage, laser ablation can achieve the same ablation volume as in a dry environment without carbonization occurring.

Overall, most of the research on laser-forming implant site focuses on isolated bone tissue, and further in vivo exploration is still needed for laser-forming implant sites on animal and human. Existing research mainly focuses on adjusting appropriate laser parameters and cooling conditions to achieve efficient bone tissue ablation while avoiding thermal damage. However, the mechanism research of laser ablation on bone tissue is still at initial stage, lacking a comprehensive understanding of the temperature field of laser action. With further research, the widespread clinical application of laser ablation forming implant sites is expected.

7. Conclusion

The implant site preparation is one of the most important operations in dental implant surgery. Currently, drilling technology is still mainly used for forming the implant site in clinical practice. To mitigate thermal and mechanical damage to bone tissue, scholars have conducted extensive research and continuously improved the structure of the drill bit and optimized the process parameters. With the continuous development of various processing technologies, many novel bone cutting techniques have emerged, providing new ideas for efficient and minimally invasive dental implant sites forming. Although these technologies are still mostly in the theoretical and experimental research stage, they show significant potential for clinical application in dental implanting.

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