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A Transient Heat Transfer Analysis of Thermal Necrosis-Aided Dental Implant Removal

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ABSTRACT

A prevalent and widely favoured solution for replacing lost teeth is the use of dental implants. The removal of dental implants, even when they are osseointegrated but unsuccessful, can be traumatic, resulting in the loss of healthy bone and adding complexity to the treatment procedure. Reducing the trauma associated with implant removal can be achieved by intentionally weakening the bone-implant attachment. To achieve this objective, a suggested approach involves utilising thermal necrosis to aid in the minimally invasive removal of implants. The objective of this study was to use finite element analysis to explore the optimal power output for intentionally inducing thermal necrosis in a dental implant. SolidWorks software was utilised to create a three-dimensional model of a dental implant assembly, which includes an abutment, screw, and implant body integrated into a segment of mandibular bone. The model was subsequently analysed using ANSYS software, applying device powers ranging from 5 to 40 W in 5 W increments on the top surface of the abutment. The results of the study showed that there was a considerable elevation in the temperatures of the bone and implant, even when employing the low power settings commonly used in electrosurgical procedures. Elevating the power level has led to a decrease in the time required for the bone and implant to reach 47°C, the initial temperature at which bone necrosis occurs. However, it is crucial to take into account the significant temperature rise in the implant body at higher power levels. The implementation of lower power settings could present a viable approach to achieving controlled osteonecrosis.

1. Introduction

Replacing missing teeth is often done using dental implants, which are widely embraced as a common and popular alternative [1, 2]. Positioned within the alveolar bone beneath, these implants provide support for various dental applications, including single-tooth replacement, anchored bridges, and full-arch rehabilitation [3]. The extraction of osseointegrated dental implants which have experienced failure, is typically associated with various complications.

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There are two primary categories of implant failure which are early failure and late failure. Insufficient osseointegration is commonly associated with early failure which occurs during the healing period [4]. The nature of this failure is primarily physiological, attributed to factors like infection, surgical trauma, and the subtle movements of the implant during the bone remodelling process. Late failure manifests after the occlusal loading of the implant and can stem from either biological factor, such as peri-implantitis and bacterial plaque subsidence, or mechanical causes like screw loosening, abutment/screw fracture, implant fracture, and implant misplacement [5]. Addressing late failures with a biological origin involves considering nonsurgical or surgical debridement, along with the use of local antimicrobial therapies, as recommended treatments [6]. In cases where implants exhibit pain during function, mobility, radiographic bone loss exceeding half the length of the implant body, and uncontrollable exudate, the recommended course of action is to remove the implant [7]. While implant failures due to mechanical factors are infrequent, the removal of implants becomes more challenging and traumatic due to their osseointegrated nature, intensifying the severity of the situation [5]. To accomplish this, commonly employed tools include trephines, drills, piezo-surgery, bone chisels, or fixture removal kits utilising reverse high torque [8]. Research on the extraction of failed implants using the reverse high torque technique is scarce, despite its demonstrated effectiveness in minimising bone damage during the removal of implants compromised in osseointegration.

Extraction of osseointegrated implants has the potential to cause damage to various underlying tissues, including neighbouring teeth, palatine bones, sinuses, and nerves. For the implant cavity, an improved bone augmentation is usually necessary, and a healing period of 9 to 12 months is required before the placement of another implant can be considered. The resulting trauma often leads to delayed treatment schedules, significantly diminishing the patient’s quality of life, and further exacerbating the medical expenses associated with the treatment [9].

Reducing the force needed for implant removal can mitigate the loss of healthy tissues and trauma resulting from the extraction of osseointegrated implants. One suggested method involves inducing a controlled thermal necrosis at the bone-implant interface, aiming to achieve a limited and controlled weakening of the attachment. A few studies have shown that, following initial unregulated random thermal therapy, implants can be swiftly extracted, leading to the development of jaw osteonecrosis. Conducting comprehensive investigations is essential to explore the effects of temperature elevation at the bone site and implant, and to pinpoint the optimal conditions for utilising thermal necrosis in achieving a non-traumatic implant removal. Reports indicate that peri-implant bone thermal damage, resulting in implant failure, has been attributed to the use of high-frequency surgical devices and dental lasers. Also, serious inflammation and necrosis of the jaw can result from uncontrolled heating. Temperature exceeding 47°C is commonly acknowledged to have a discernible impact on bone tissue [10, 11]. Osteonecrosis with a localised and nonprogressive thermal effect is observed at a temperature of 47°C. Upon reaching 56°C, the denaturation process affects alkaline phosphatase. Temperatures of 60°C and beyond could lead to the occurrence of extensive and advancing osteonecrosis.

Despite the promise shown in earlier studies, there has been a lack of controlled and systematic investigations to explore the complete potential and limitations of this method. Subjecting the bone-implant interface to controlled heating at 47°C can strategically weaken the attachment, inducing localised and controlled thermal osteonecrosis. This approach holds potential benefits in mitigating the trauma associated with implant removal. The objective of this study was to explore the ideal device power required for inducing intentional thermal necrosis on a dental implant through the application of finite element analysis (FEA). The heat power level ranging from 5 to 40 W with an increment of 5 W was imposed to the implant that placed in a mandibular bone segment, simulating contact with unipolar electrocautery tips across a spectrum of low to high output settings. The temperature rise in the models was plotted, pinpointing specific time intervals when the bone and implant body achieved a temperature of 47°C.

2. Materials and Methods

2.1 Model Development

In this study, a three-dimensional (3-D) model of the lower jawbone or mandibular bone segment was utilised. This model was derived from a series of computed tomography images, and the resulting scanned dataset was imported into Mimics 20.0 (Materialise, Leuven, Belgium). Subsequently, the bone tissue layers—cortical and cancellous—were distinguished based on Hounsfield units. The bone tissue models underwent further refinement and modifications using SolidWorks 2020 (SolidWorks Corp., Concord, Massachusetts, USA), a computer-aided design software. The final bone model, representing a toothless posterior mandible, consisted of an approximately 2-mm thick cortical layer enveloping a core of solid cancellous bone. The overall dimensions of the bone model were 20 mm in height, 30 mm in length, and 8–10 mm in width. This bone model was classified as type II according to the Lekholm and Zarb classification [12]. Figure 1(a) depicts the configuration of the mandible section model considered in the analysis. For the dental implant system modelling, a solid cylindrical, single-threaded implant body with a length and diameter of 10 mm and 3.75 mm, respectively, was constructed. These dimensions were based on the dual-fit implant (DFI) system manufactured by Alpha-Bio Tec (Petach Tikva, Israel). The implant body was modelled with an internal hexagonal connection and V-shaped thread. A straight abutment, 3.5 mm in height, was designed and affixed to the implant body. The connection between the abutment and the implant body was established using an abutment screw model with a width of 2.2 mm and a length of 8 mm.

In order to replicate the positioning of the implant body within the bone, a bone-level approach was adopted. This involved aligning the flat surface of the implant platform with the most superior surface of the mandibular bone model. Using the “Subtract” tool within the SolidWorks software, a hole with a diameter of 3.75 mm was created to facilitate and confirm the position of the implant. Figure 1(b) shows the placement of the implant assembly in the bone model.

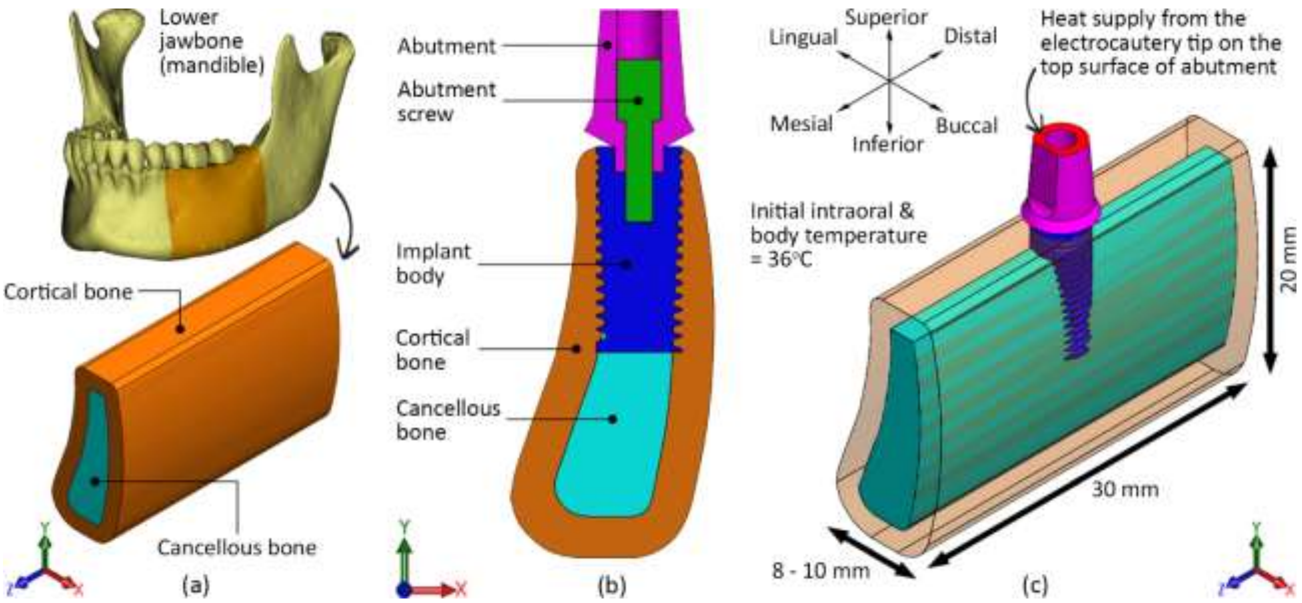

Fig. 1. The mandibular bone model consisting of the cortical and cancellous layers. (b) The placement of the implant assembly in the bone. (c) The application of heat at the top surface of the abutment

2.2 Computational Analysis Pre-processing

The implant body, abutment, and abutment screw were made of grade 5 titanium alloy (Ti-6Al-4V). All the material properties of geometry including the cortical and cancellous bones were assumed to be isotropic, homogenous, and linearly elastic. To perform the heat transfer analysis, two main thermal properties were taken into account which are thermal conductivity (k) and specific heat (C_p). Also, the density of each geometrical model was considered. Table 1 exhibits the physical and thermal properties of the implant parts and bone tissues used in the analysis. All these physical and thermal properties were defined prior to the meshing process of the solid models.

Table 1
The list of physical and thermal properties of all models utilised in the present study

Model	Density, ρ (g/cm^3)	Thermal Conductivity, k (W/m°C)	Specific Heat, C_p (kJ/kg°C)	References
Implant parts (Ti-6Al-4V)	4.38	6.52	0.57	Prabhu *et al.,* [13]
Cortical bone	1.30	0.59	0.44	Prabhu *et al.,* [13]
Cancellous bone	1.30	0.59	0.44	Prabhu *et al.,* [13]

In terms of contact modelling, the cortical and cancellous bones were assumed to be in perfect unity wherein the interfaces among both bone layers were modelled as bonded. This type of contact was also applied to the contact surfaces among implant components, and between the implant body and bone tissue [14]. Utilising the direct contact method for executing bonded contact modelling serves to eliminate the possibility of any relative motion at these interfaces.

All the bodies were set to be at a constant temperature of 36°C at the initial stage of simulation [13]. This particular temperature selection aligns with the standard temperature of the human body. Considering power levels of 5, 10, 15, 20, 25, 30, 35, and 40 W, we simulated the effects of temperature rise in the assembly during implant removal [14]. These power outputs were applied on the top surface of the abutment as shown in Figure 1(c). This simulation replicated how the implant assembly interacts with a unipolar electrosurgery tip in different power outputs. During the analysis, a time step of 0.2 seconds was used, and the measurements were conducted over a total duration of 4 seconds [13]. The analysis involved measuring the time in seconds to ascertain the duration required for the cortical-cancellous bone assembly and implant body, to reach a temperature of 47°C.

2.3 Mesh Sensitivity Test

To mitigate the influence of purely numerical factors on the analysis results, a mesh convergence test was carried out. Executing this test involved transforming all geometries into finite element models through the employment of solid tetrahedral elements within the ANSYS software (ANSYS Inc., Houston, TX, USA). These tetrahedral elements, featuring a four-node structure, afford three degrees of freedom. In the preparation of the models, six varying mesh densities were taken into consideration, spanning from around 1,690,000 elements (Tet F, the highest density) to 190,000 elements (Tet A, the lowest density).

The focal point of the mesh sensitivity test was the assessment of the maximum principal stress value observed in the bone across all models. A slight variance in stress levels between the refined and coarser models was evident from the results. Following a single refinement, the model showed convergence, with a mere 2.7% variation in the stress values. The converged model, with around 400,000 nodes and 260,000 elements, was formed. Figure 2 illustrates how the maximum principal stress value is distributed among all meshed models. The figure also presents a comparison of the meshing elements generated in the refined and coarser models.

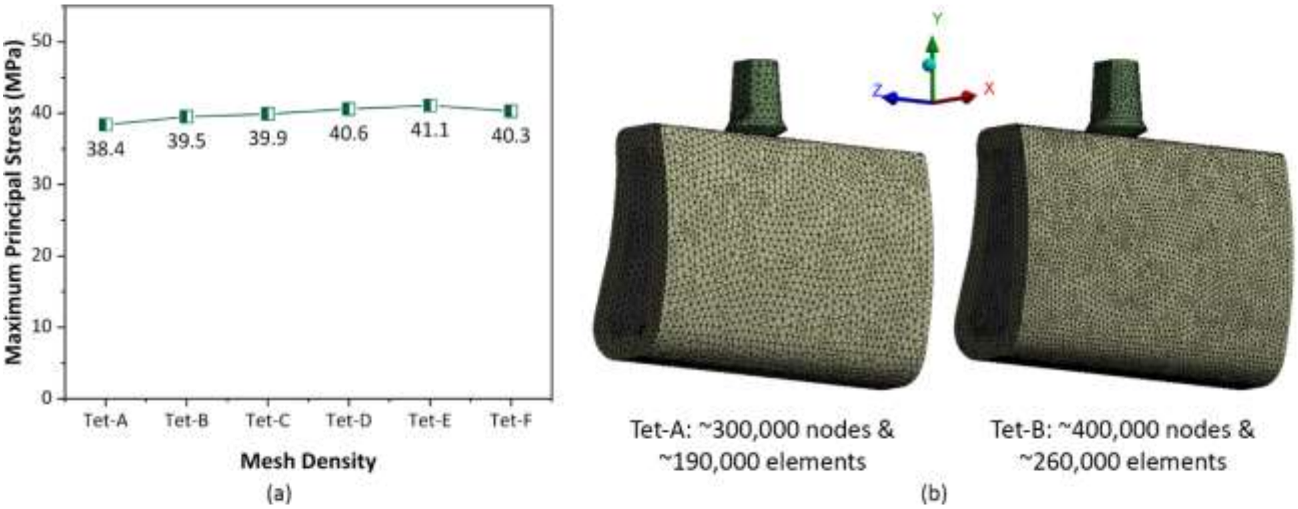

Fig. 2. (a) The maximum principal stress plot for all the models. (b) The configuration of tetrahedral meshes for the coarser (left) and more refined (right) models

3. Results

For the achievement of noteworthy results, our focus has been placed on evaluating the duration required for the cortical-cancellous bone union and implant body to reach a temperature of 47°C across all the power levels. Also, the colour contour plots of the result indices were instrumental in reinforcing the interpretation of the findings, vividly illustrating the distribution of temperature. Table 2 presents a concise overview of the duration taken for the bone and implant body to reach 47°C in each of the eight varying power outputs.

Table 2
The duration recorded by the bone and implant body to reach 47°C in different power outputs

Part	Time (s)							
	5 W	10 W	15 W	20 W	25 W	30 W	35 W	40 W
Cortical-cancellous bone	4	3.03	2.61	2.36	2.2	2.06	1.96	1.88
Implant body	3.04	2.28	1.95	1.76	1.62	1.52	1.44	1.38

3.1 Duration of Cortical-Cancellous Bone Union Reaching 47°C

The results indicate that the increase in heat supply (power output) from the electrosurgery tip generally reduces the time it takes for the surrounding bone to reach the temperature of bone necrosis (47°C). Figure 3(a) illustrates that the longest duration was recorded at 5 W power (4 s), while the shortest duration occurred at 40 W power (1.88 s). It is evident that the reduction in duration from the 5 W to 40 W power levels ranged from 28.5% to 9.5%. A more significant difference in duration decrease was observed when applying lower heat supply values (5 –20 W) compared to higher ones (25 – 40 W).

Regarding the colour contour plots of the bone temperature, it is worth noting that a higher temperature level was observed in the cervical bone region surrounding the implant neck. The dispersion of high-temperature values in that region became even wider as the power output increased, as illustrated in Figure 4. This is substantiated by the escalating value of the maximum temperature (47 – 124.1°C) generated in the models of 5 to 40 W. The middle and inferior regions of the bone structure were primarily influenced by moderate and minimal thermal effects. The dissipation of heat from high to low-temperature areas was uniform in the radial direction across all power outputs.

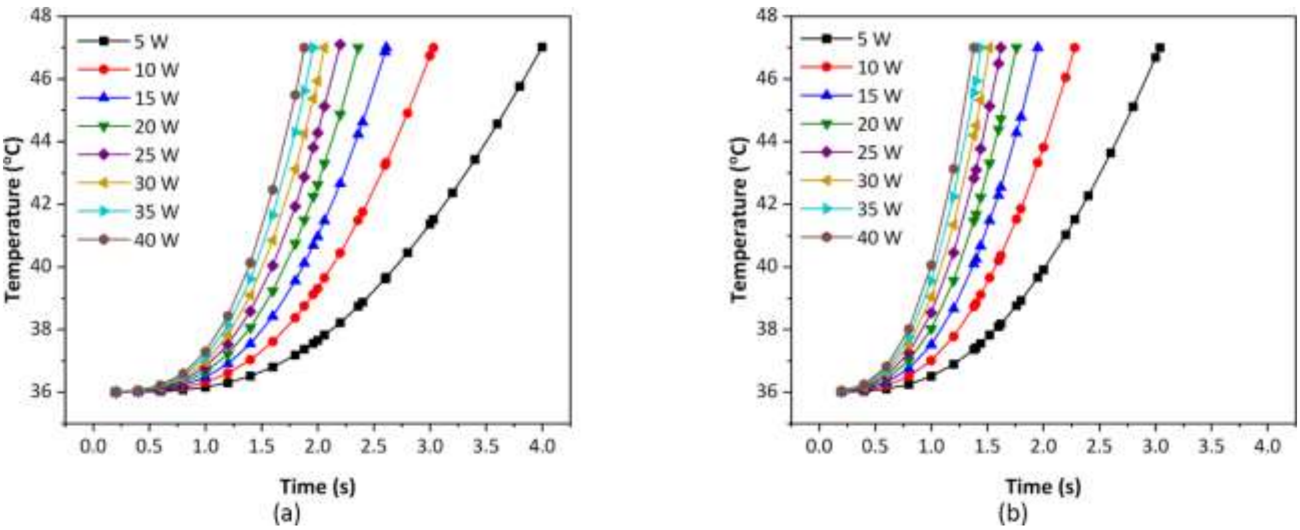

Fig. 3. The plot of duration for the (a) cortical-cancellous bone union and (b) implant body to reach 47°C in all power levels

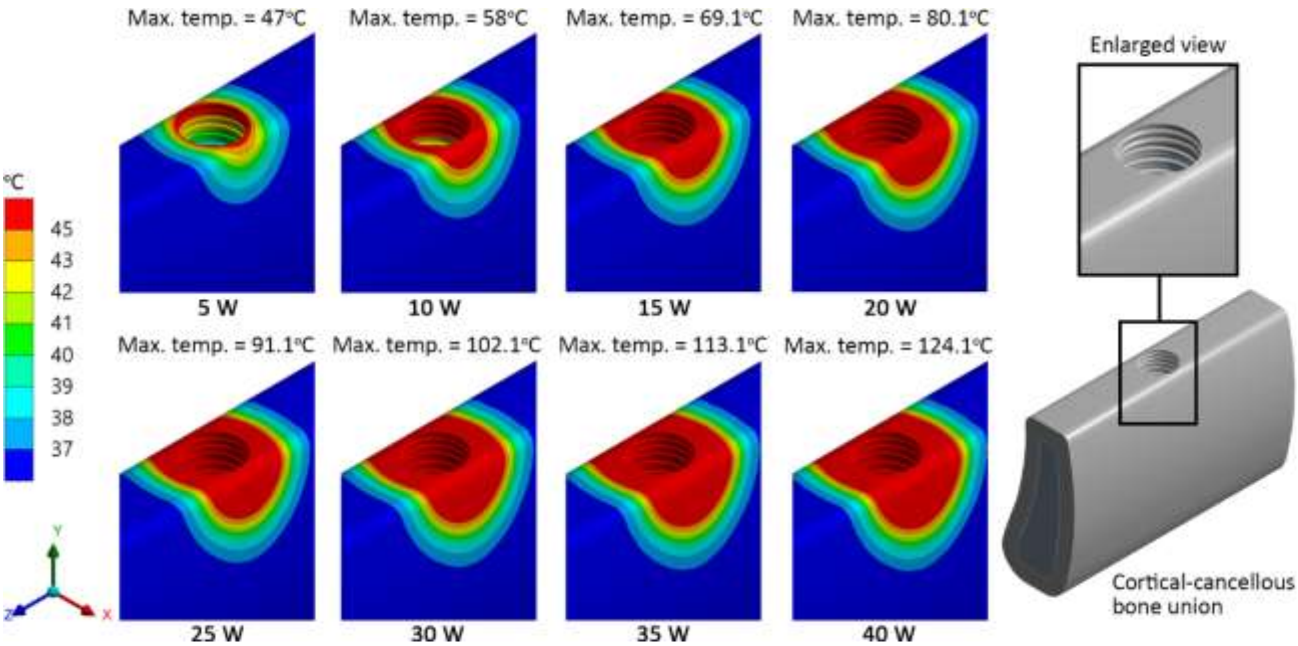

Fig. 4. The colour contour plot of temperature levels within the bone in 4 s across all power levels

3.2 Duration of Implant Body Reaching 47°C

For the implant body, the trend of temperature value increase that reaching 47°C across various power outputs appears to correspond closely with those recorded in the surrounding bone. The elevated power levels have notably reduced the time it takes for the implant body to reach the temperature of bone necrosis. The longest duration, 3.04 s, was observed in the 5 W model, while the shortest duration, 1.38 s, was recorded in the 40 W model. The difference in the duration reduction for the 5 – 40 W models ranged from approximately 30.7 to 11.7%. Higher power outputs showed less disparity in the duration decrease compared to lower power outputs. Additionally, in comparison to the bone, it is evident that the total duration for the implant body to reach 47°C was significantly shorter across all power levels, with percent differences ranging approximately from 22.2 to 33.3%. Therefore, it simply indicates that the implant body possesses a considerably higher temperature value than the bone when the bone reaches 47°C irrespective of the power outputs.

The temperature distribution plot over the implant body reveals that the coronal portion, especially at the edge of the abutment connection, sustains a high temperature magnitude. Figure 5 illustrates that the thermally loaded area increases in size as the power output escalates, extending up to the middle part of the implant body. This aligns well with the increasing maximum temperature values recorded, ranging from 55.7 to 193.3°C for power levels of 5 to 40 W. Regardless of power outputs, the temperature level was evenly decreased from the coronal to apical parts of the implant body. The apical part exhibits lower temperatures, indicated by blue colour plots.

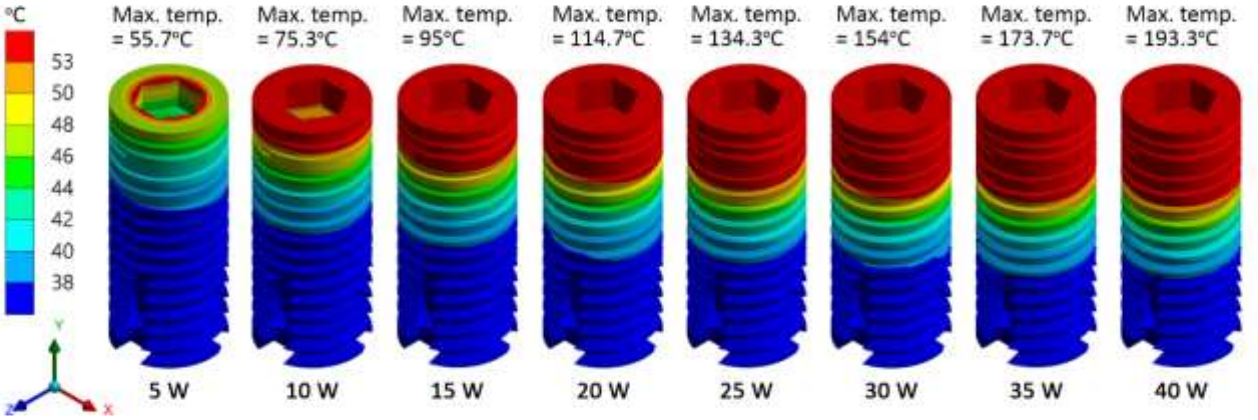

Fig. 5. The colour contour plot of temperature levels within the implant body in 4 s across all power levels

3.3 Temperature Distribution in the Implant-Bone Assembly

When plotting the temperature magnitudes in the implant-bone assembly, it became evident that the top region of the abutment body accumulated higher temperatures compared to other parts of the assembly. The degree of critical temperature distribution increased further with an escalation in power output. As depicted in Figure 6, the highest temperature in the assembly was generated at 40 W power (2646.5°C), while the lowest was recorded at 5 W power (362.3°C). Overall, there was a linear correlation between the level of heat supply and the generated temperature values.

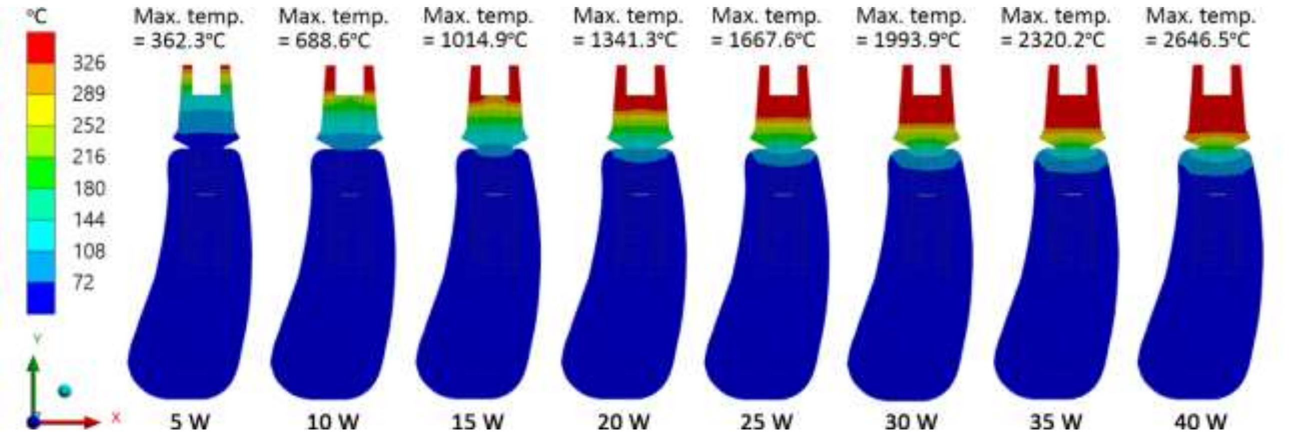

Fig. 6. The colour contour plot of temperature levels within the implant-bone assembly in cross-sectional view in 4s across all power levels

4. Discussion

In this study, we evaluated the variation in power outputs simulating the use of an electrosurgery tip for removing dental implants from the bone, employing finite element analysis. The increase in the temperature of the dental implant assembly is generally anticipated due to factors such as exposure to hot substances [15, 16], an electrocautery unit [17], and laser surgery [18]. However, the controlled increase in temperature, as examined in the present study, serves as a method for facilitating the extraction of failed osseointegrated dental implants. The proposed temperature increase is expected to result in a constrained osteonecrosis at the interface between the implant and bone.

In a previous technical report [19] and two case reports [9, 20], it has been documented that the removal of failed dental implants involved applying a reduced torque, achieved by heating the implant through contact with electrocautery tips or by applying laser. The primary focus of these reports; however, has been on the extraction torque of the implants and the ensuing clinical parameters, including complications in the healing process or further bone resorption. There was no exploration into the temperature increase within either the implants or the bone. As far as we know, the experimental investigation of temperature rise in dental implants due to contact with electrocautery tips has only been conducted by Wilcox *et al.,* [17]. They observed a temperature rise of 8.878°C as a result of a one-second contact with a unipolar electrocautery tip set at 5 W. Our findings demonstrated a marginal decrease in temperature elevation at an equivalent duration and power level, measuring 5.646°C. The absence of discussion regarding the size of the implant parts or surgical tips prevents us from establishing a more comprehensive correlation with the experimental results reported by Wilcox *et al.,* Moreover, Wilcox *et al.,* did not investigate the temperature rise within the bone [17].

The findings of the present study highlight the pivotal role of power output level in achieving controlled thermal necrosis. Generally, elevating the power output of the electrocautery unit reduces the time required for the bone and implant to reach 47°C. To put it simply, higher power output leads to a faster temperature increase. Our findings align with a previously published numerical study by Prabhu *et al.,* wherein the lower power level (5 W: > 4 s) exhibited a longer duration of heating compared to the higher one (40 W: 3.34 s) [13]. Another computational investigation conducted by Gungormus *et al.,* also observed a similar trend, indicating that an increase in power output could lead to a shorter duration for temperature rise [14]. However, the specific time duration values recorded in these studies differed slightly from ours, possibly due to variations in geometry, implant type, and the scope of the investigated problem.

Opting for a higher power level from the heating source may be preferable, facilitating a quicker heating of the bone-implant interface during the implant removal process. Nevertheless, close attention to the temperature rise within the implant body itself is crucial. Our observations reveal that the implant body reaches 47°C significantly faster than the bone across all power levels (see Figure 3). In essence, when the bone reaches 47°C, the implant temperature is considerably higher. Moreover, this temperature difference further amplifies with increasing power output. This is evident in the escalation of maximum temperature from 5 to 40 W models at the final analysis time (4 s), as illustrated in Figure 5. The rapid and substantial increase in implant temperature at higher power levels appears challenging to control and manage. Conversely, lower power outputs provide a more suitable and acceptable range of implant temperature for surgical operations. Thus, considering the balance between power output and temperature control is crucial for effective and safe procedures.

In terms of temperature distribution within the entire assembly (refer to Figure 6), the critical temperature value was initially detected in the abutment before diminishing radially and inferiorly. This observation can potentially be explained by the direct application of heat supply to the top surface of the abutment.

Several limitations were encountered in the current study, with the most notable being the lack of variation in the heat contact area or the size of the electrocautery tip. The mandibular bone was represented as a solid segment of bone, receiving heat exclusively from a single thermally loaded abutment. The resulting temperatures of the bone and implant could have been influenced by considering variations in heat sources applied to different contact areas or tip sizes. Furthermore, all components were assumed to have isotropic, linear, and homogeneous thermal and physical properties, despite the inherent inhomogeneity of biological tissues in reality.

5. Conclusion

The potential of thermal necrosis-aided implant removal as a promising method lies in its ability to assist clinical practitioners in minimising trauma and avoiding unnecessary bone loss during the extraction of failed osseointegrated dental fixtures. This study discovered that the bone and implant temperatures experienced a significant increase even at the low power settings commonly employed in electrosurgical procedures. Increasing the power level has resulted in a reduction of the time it takes for the bone and implant to reach 47°C, the initial temperature at which bone necrosis occurs. However, considering the substantial temperature increase in the implant body with higher power levels is critical, the application of lower power settings could be a viable approach for achieving controlled osteonecrosis. Further exploratory *in vitro* and *in vivo* studies are expected, utilising parameters derived from this study to examine and validate the applicability of thermal necrosis-aided implant removal.

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