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A Thermal and Biological Analysis of Bone Drilling

With the introduction of high-speed cutting tools, clinicians have recognized the potential for thermal damage to the material being cut. Here, we developed a mathematical model of heat transfer caused by drilling bones of different densities and validated it with respect to experimentally measured temperatures in bone. We then coupled these computational results with a biological assessment of cell death following osteotomy site preparation. Parameters under clinical control, e.g., drill diameter, rotational speed, and irrigation, along with patient-specific variables such as bone density were evaluated in order to understand their contributions to thermal damage. Predictions from our models provide insights into temperatures and thresholds that cause osteocyte death and that can ultimately compromise stability of an implant. [DOI: 10.1115/1.4040312]

Keywords: osteotomy, bone, drill, heat, model

Introduction

Bone has a microstructure that incorporates multiple interfaces, e.g., the juxtaposition of marrow spaces next to mineralized matrix, and mineralized matrix adjacent to fibrous tissues. These multiple interfaces have an impact on the thermal properties of bone, which, in turn, affect heat transfer during various procedures such as drilling. An understanding of heat transfer in bone as a composite material may lead to practical improvements in procedures such as osteotomy site preparation [1].

The first evidence of drilling bone dates from 800 CE [2]. Since then a variety of studies have focused on the thermal effects of drilling bone [3–7]. Drilling creates heat, and the temperature rise in bone is typically measured with a thermocouple. Due to practical issues, however, the thermocouple is usually positioned at a distance from the drill tip and therefore peak temperatures escape measurement.

To address such challenges, computational methods have been developed to investigate temperature distributions in bone caused by drilling. Some models take into account the relative density of bone but typically assume homogenous properties for the tissue [8]. This assumption belies a complex biological reality: trabecular bone is a porous, heterogeneous material, e.g., the jawbones, the growth plates, and in locations such as the iliac crest, whereas cortical bone is dense and homogenous, e.g., the outer shells of long bones. The differences between bone types are expressed by the ratio of bone volume (BV)/total volume (TV), where the numerator represents the mineralized BV and the denominator represents the TV in a given region of interest (ROI), which can include fluid-filled nonmineralized marrow spaces. Cortical bone typically has a high BV/TV value, on the order of 100%, because the majority of the region of interest is occupied by mineralized matrix. Because it includes marrow cavities, trabecular bone typically has a lower BV/TV, on the order of 50–70% in the jawbones [9]. The structural heterogeneity impacts both heat generation and heat dissipation during drilling.

In this work, we performed image-based computational modeling using composite material behavior for trabecular bone and coupled it with a biological assessment of the cellular responses to

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Fig. 1 Bone drilling, schematics, and heat generation: (a) an example of drilling in the rat maxillary alveolar bone; note the relative size of the drill to the anatomic location, e.g., the edentulous ridge, **(b)** schematic of bone cutting and the associated zones of heat generation, **(c)** the drill and parameters; adapted from Ref. [10], and **(d)** heat from drilling cortical bone and the linear relationships between heat and drill diameter for different rotational speeds

drilling. We validated the computational model and gained critical insights into two-dimensional (2D), transient temperature distributions in bone of various densities following clinical drilling protocols. Together, the results provide guidance on the biological consequences associated with osteotomy site preparation.

Methods

Estimation of Drilling Heat. Heat generated during drilling (Fig. 1(a)) predominantly arises from a primary deformation zone (Fig. 1(b)), where energy involved in shearing the bone material is converted to heat [10]. In a secondary zone, the heat generated dissipates into the bone chip and the tool and in a tertiary zone, heat results from friction between the tool and bone (Fig. 1(b)). Our computational model of heat transfer focused on periosteotomy bone.

To estimate heat generated by drilling bone as a function of drill diameter and rotational speed, analytical results from Ref. [8] were used, where heat from drilling cortical bone is given for different rotational speeds and shear areas of the chip (Table 1).

The shear area of a chip was approximated by

$$A_s = \frac{t_1(R - r_{ce})}{\sin(p)\sin\Phi} \quad (1)$$

where t_1 is the thickness of the cut chip, R is the radius of the drill bit, r_{ce} is the chisel edge radius, and p and Φ are the half point and the shear angles, respectively (Figs. 1(b) and 1(c)).

To calculate heat as a function of drill diameter, we identified the drill diameter corresponding to any given shear area by rearranging Eq. (1). Since it is difficult to measure during clinical use, we assumed a chip thickness of 20 μm based on values from Ref. [8], and half-point and shear angles of 45 deg based on measurements and approximations for commercially available Osteomed drills. Also, the chisel edge radius was neglected since it is small relative to the radius of the drill bit. Therefore, drill diameter was calculated for given shear areas, and curves were fitted to the heat values as a function of the drill diameter for each rotational speed (Fig. 1(d)). It was seen that the heat from drilling has a linear relationship with the drill diameter, and the slope of

Table 1 Analytical results of heat generation due to cortical drilling for different shear areas and rotational speeds [8].

Shear area (mm ²)	Heat (W)					
	350 rpm	500 rpm	650 rpm	800 rpm	950 rpm	1100 rpm
0.01761	0.06	0.087	0.11	0.138	0.161	0.189
0.02349	0.08	0.117	0.147	0.184	0.215	0.252
0.02936	0.1	0.146	0.184	0.23	0.268	0.315
0.03523	0.12	0.175	0.221	0.276	0.322	0.377
0.04111	0.14	0.204	0.258	0.322	0.376	0.44
0.04697	0.16	0.233	0.295	0.368	0.43	0.503
0.05286	0.18	0.262	0.331	0.414	0.483	0.566

Fig. 2 Computational domain. The majority of osteotomies in the oral cavity are generated in healed extraction sites; this condition was replicated in a model organism (rat): μ CT images: (a) osteotomy in alveolar bone, and (b) superimposed on which is a schematic of the heat transfer during bone drilling. (c) Computational model, which takes into account the appropriate bone density, and boundary conditions. The heat flux was applied at the drill tip, which was moved vertically.

the drill heat-diameter curves increased linearly with increase in the rotational speed. Thus, a linear fit between the drill heat, drill diameter, and rotational speed was proposed

$$Q_{\text{Drill}} \cong cND \quad (2)$$

where Q_{Drill} is the drill heat (in Watts) due to cutting a bone chip; N is the rotational speed (in rpm); D is the diameter of the drill bit (in mm); and c is a constant with a value of 2×10^{-4} W min/mm for cortical bone. The heat generated by cutting is conducted to the tool, the chip, and the intact bone. The fraction of heat that enters the bone was empirically determined by comparing computational and experimental temperatures [10]. It should be noted that this fraction does not include the convection heat removed by irrigation (described later). The heat flux that enters the bone during drilling was calculated as

$$q_{\text{Drill}} = \eta \frac{Q_{\text{Drill}}}{\pi D \Delta z} \quad (3)$$

where q_{Drill} is the heat flux (in W/m²); η is the fraction of the heat that enters the bone, found to be 0.5 [10]; and Δz is the height of the mesh covered by the drill node.

Computational Modeling of Temperature Distributions in Bone. Heat generated at the drill tip is partly conducted into surrounding bone and potentially results in thermal necrosis or apoptosis. The computational model's ROI was the periosteotomy bone as shown in microcomputed tomography (μ CT) images (Figs. 2(a) and 2(b)). A schematic of the two-dimensional axisymmetric computational model and mesh in cylindrical coordinates—and the boundary conditions—is presented in Fig. 2(c). $r_i = D/2$, $r_o = r_i + 2$ mm, and $H = 3$ mm represent, respectively, the radius of the drill hole (assumed equal to the drill radius), the outer radius of the bony sample, and the height of the annulus, i.e., the depth of the drill hole, with all values measured from the μ CT images as noted below. The domain size was chosen sufficiently wide, so the change in temperature results was

negligible with further extending the outer boundary in the radial direction.

The temperature distribution in bone was calculated by solving the transient heat conduction equation

$$\rho c_p \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) \quad (4)$$

where T is the bone temperature; and k , ρ , c_p are thermal conductivity, density, and heat capacity, respectively. Assuming that the temperature in bone varied with radial (r) and vertical (z) dimensions, the heat equation in cylindrical coordinates can be written as

$$\rho c_p \frac{\partial T}{\partial t} = \frac{1}{r} \frac{\partial}{\partial r} \left(kr \frac{\partial T}{\partial r} \right) + \frac{\partial}{\partial z} \left(k \frac{\partial T}{\partial z} \right) \quad (5)$$

The bottom and top boundaries ($z = 0$ and $z = H$) were assumed to be insulated

$$\frac{\partial T}{\partial z} = 0 \quad (6)$$

The temperature at the right boundary of the bone sample ($r = r_o$) was assumed to remain at body temperature

$$T(r_o, z, t) = T_o \quad (7)$$

where T_o is the body temperature equal to 37°C. The heat flux was applied to the drill hole's boundary ($r = r_i$) where the tip of the drill was located

$$-k \frac{\partial T}{\partial r} = q(z, t) \quad (8)$$

The tip was moved vertically with a feed rate assumed to be constant and equal to the depth covered by the drill inside the bone

divided by the drilling time. Below the drill tip, the value of the heat flux was set to zero. A convection boundary condition was applied above the drill tip

$$k \frac{\partial T}{\partial r} = h(T - T_{\infty}) \quad (9)$$

where h and T_{∞} are the ambient heat transfer coefficient and temperature, respectively. The initial temperature in the domain was set to 37 °C

$$T(r, z, 0) = T_o \quad (10)$$

To investigate the effect of BV/TV and to determine temperature variations and heat transfer characteristics in trabecular versus cortical bone, trabecular bone was assumed to be a composite material, with the ratio of solid/fluid being chosen to roughly match the BV/TV of alveolar bone. No cutting happens in fluid, marrow-filled spaces; consequently, the drill heat flux was set to zero at these layers.

The two-dimensional transient temperature distribution in bone with varying thermal properties was numerically calculated in MATLAB using finite difference method, and Euler scheme for time-stepping. The mesh was refined using the same height covered by the drill until there was ~1% change in the calculated temperatures with further mesh refinement.

Thermal parameters for cortical bone were as follows: $k_{\text{bone}} = 0.6$ (W/mK); $\rho_{\text{bone}} = 1800$ (kg/m³); $c_{p,\text{bone}} = 1260$ (J/kgK) [11]; and for bone marrow: $k_{\text{marrow}} = 0.3$ (W/mK); $\rho_{\text{marrow}} = 1029$ (kg/m³); $c_{p,\text{marrow}} = 2666$ (J/kgK) [12] were employed. Fluid convection coefficients were as follows: $h_{\text{control}} = 25$ (W/m²K) for ambient air [13], and $h_{\text{irrigation}} = 4000$ (W/m²K) to approximate the experimental irrigation rate [5]. The convective heat transfer coefficient has been calculated and is on the order of 10^1 W/(m² K) for air, and 10^3 W/(m²K) for irrigant [14].

Bone Tissue Collection From Human Subjects for Thermal Analysis. Six adult patients (three men, three women) who were undergoing osteotomies in preparation for subsequent maxillary dental implant placement volunteered to participate in the study. Informed consent was obtained from each participant and all procedures and materials were approved by local ethics committees and the Institutional Research Board (WIRB#20160967 for U.S. private practices, conformed to STROBE, strengthening the reporting of observational studies in epidemiology, guidelines). The drilling protocol involved a pilot drill diameter of 1.6 mm used at 1000 rpm with manual saline irrigation. The drilling depth was ~3 mm and duration was ~3 s. A BV/TV value of ~70% was used based on the μ CT analysis. Bone biopsies were collected using a trephine (Karl Schumacher, Linden, NJ) centered on the

pilot drill hole. The trephine and its contents were immediately transferred to 4% paraformaldehyde and after overnight fixation, samples were washed in ice-cold phosphate-buffered saline.

Apoptosis Analyzes and Quantification of the Zone of Death. TUNEL staining (#11684795910, Roche, Indianapolis, IN) was used to detect programmed cell death in osteocytes of the peri-osteotomy bone from patients. Histomorphometric measurements were performed using IMAGEJ software. Each trephined sample was sectioned through entire occluso-apical depth. Six samples (24 slides) were analyzed. Each slide was imaged at 10x and the distance from the edge of the osteotomy to TUNEL⁺ was measured. Multiple measurements were made in each quadrant around the osteotomy.

Microcomputed Tomography. For μ CT imaging, samples were fixed in 4% paraformaldehyde overnight at 4 °C, then transferred to 70% ethanol for scanning. The samples were scanned at 10.5 μ m resolution (70 kV, 115 μ A, 300 ms integration time) on a μ CT data-acquisition system (VivaCT 40, Scanco, Brüttisellen, Switzerland). Bone morphometry was performed using acquisition system's analysis software (Scanco). Reconstruction and volume rendering were carried out using Avizo (FEI, Hillsboro, OR) and ImageJ (NIH, Bethesda, MD) software.

Results

Validation of the Heat Transfer Model for Drilling in Bone.

We first conducted a validation test of the computational model and compared temperature predictions from the current analysis to experimentally measured temperatures in cortical bone during drilling. The same input parameters were employed [15], which included a 2.0 mm drill diameter, a rotational speed of 800 rpm, a cortical bone thickness of ~2 mm, and a drilling duration of ~8 s; temperatures were measured at a distance of 0.5 mm from the drill. Thermal simulation results (red line, Fig. 3(a)) matched experimentally measured temperatures as a function of time (black circles, Fig. 3(a)).

As an additional validation test, we compared simulation results with experimental data gained from tests, which were all performed in cortical bone, i.e., bone without marrow spaces, as a function of radial distance from an osteotomy for two different cases [5]: In one iteration, irrigation was excluded (black squares, Fig. 3(b)), and in a second iteration, irrigation was employed during drilling (green triangles, Fig. 3(b)), where the temperature in the bone was lowered by the use of irrigation. A 3.2 mm drill diameter, a rotational speed of 885 rpm, and a cortical bone thickness of 5 mm were used as input parameters, and the calculated

Fig. 3 Validation of the heat transfer model. Computational and experimental bone temperatures versus (a) drilling duration and (b) radial distance from the drill. Computational results agreed well with previous experimental data and demonstrated that peak temperatures occurred in the immediate vicinity of an osteotomy.

Fig. 4 Distribution of apoptotic osteocytes and calculated temperatures demonstrate a correlation between biological and thermal assessments. (a) Using TUNEL staining, a zone of apoptosis (arrow) is evident around the edge of the osteotomy (dotted line). The distribution of dying osteocytes corresponds to (b) calculated temperatures in the immediate vicinity of the drill. The threshold temperature for osteocyte apoptosis was determined as $\sim 47^{\circ}\text{C}$. Both the biological response (TUNEL staining) and the temperature contours are shown at the same relative magnification.

average maximum temperatures matched the experimental measurements (Fig. 3(b)).

From these analyses, we drew two conclusions: First, computational results agreed well with experimental data insofar as predicting temperatures versus radial distance from the osteotomy edge, and time since the initiation of drilling. Second, the results demonstrated that peak temperatures occurred in the immediate vicinity of an osteotomy (Fig. 3(b)), but thermocouples were only capable of measuring bone temperatures at a distance, e.g., $500\ \mu\text{m}$, from the osteotomy edge. We postulated, however, it was in this region that most important biological changes would occur. Our next experiments, therefore, set out to determine the biological sequelae of the calculated temperature distributions.

A Zone of Peri-Osteotomy Osteocyte Death Correlates With the Calculated Temperature Rise Caused by Drilling. At what temperature do osteocytes begin to die? To answer this question, we directly compared the distribution of temperature elevation due to drilling with the spatial distribution of apoptotic osteocytes. The input parameters used to calculate the temperature distributions matched the parameters used to produce the osteotomy (described in Methods).

The extent of dying osteocytes around the osteotomy was evaluated using TUNEL staining. Twelve hours after drilling, the majority of TUNEL⁺ cells were detected within a $207 \pm 19\ \mu\text{m}$ radial zone around the osteotomy (Fig. 4(a)). We correlated the calculated average maximum temperatures with the thermally affected zone and determined that the threshold temperature for osteocyte apoptosis was $\sim 47^{\circ}\text{C}$ (Fig. 4(b)). This finding is consistent with previously reported temperature thresholds [16].

We then calculated temperature distributions and thermally affected zones resulting from clinical drilling protocols (Table 2, and Figs. 5(a)–5(f)). BV/TV values reflecting the average dental patient population were employed [17] and rotational speed and

drill diameters mirrored standard procedures for implant site preparation, e.g., Ref. [18]. These data demonstrated that if some variables were held constant—such as drill rotational speed, the patient's BV/TV, and the use of irrigation—then the change in drill diameter had a minimal impact on the size of the thermally affected zone. For example, even if the diameter of the drill was increased by more than three-fold, the change in the size of thermally affected zone was approximately twofold, e.g., a maximum difference of $\sim 350\ \mu\text{m}$ (Fig. 6(a)).

Of the variables under a clinician's control, drill rotational speed and irrigation had the greatest impact on the size of the thermally affected zone. For example, when other variables were kept constant, a three-fold increase in rotational drill speed resulted in up to approximately eightfold change in the size of the thermally affected zone, e.g., a maximum difference of $\sim 800\ \mu\text{m}$ (Fig. 6(b)). Again, holding other variables constant, the use of irrigation caused up to approximately tenfold change in the size of the thermally affected

Table 2 Bone maximum temperature and thermally affected zone approximations for various BV/TV values and drilling protocols

BV/TV = 50%						
Rotational speed (rpm)	500		1000		1500	
Diameter (mm)	1.6	5	1.6	5	1.6	5
Maximum temperature ($^{\circ}\text{C}$)	58	60	79	82	100	105
Thermally affected zone (mm)	0.05	0.1	0.25	0.35	0.4	0.55
BV/TV = 70%						
Diameter (mm)	1.6	5	1.6	5	1.6	5
Maximum temperature ($^{\circ}\text{C}$)	68	71	99	104	129	138
Thermally affected zone (mm)	0.2	0.25	0.45	0.6	0.65	0.85
BV/TV = 100%						
Diameter (mm)	1.6	5	1.6	5	1.6	5
Maximum temperature ($^{\circ}\text{C}$)	82	88	128	139	174	190
Thermally affected zone (mm)	0.35	0.6	0.8	1.1	1.05	1.4

Fig. 5 Average maximum bone temperatures as a function of radial distance from the drill for various bone densities and clinical drilling protocols. (a) $N = 500$ rpm; $D = 1.6$ mm, (b) $N = 500$ rpm; $D = 5.0$ mm, (c) $N = 1000$ rpm; $D = 1.6$ mm, (d) $N = 1000$ rpm; $D = 5.0$ mm, (e) $N = 1500$ rpm; $D = 1.6$ mm, and (f) $N = 1500$ rpm; $D = 5.0$ mm. The gray horizontal line indicates the threshold for the thermally affected zone. Temperature distributions and thermally affected zones were calculated for clinical and patient-specific parameters.

zone, e.g., a maximum difference of $\sim 850 \mu\text{m}$ (Fig. 6(c)). Together these data indicated that changes in the drill rotational speed affect the bone temperatures and zones of osteocyte death more than the changes in the drill diameter and that rotational speed and irrigation are critically important in the extent of osteocyte cell death resulting from osteotomy site preparation.

Bone Density Critically Affects the Zone of Osteocyte Death. One variable that had a significant impact on the size of the thermally affected zone—but is not under the clinicians' control—was the BV/TV of the bone being drilled. Holding other variables constant, a two-fold increase in BV/TV significantly changed the size of the thermally affected zone; for example, altering the BV/TV from 50% to 100% increased the size of the thermally affected zone by $\sim 850 \mu\text{m}$ (Fig. 6(d)). Therefore, of all the variables affecting the “zone of death,” the critical variable of BV/TV was outside the clinician's control.

Discussion

A Computational Model That Predicts Biological Responses to Bone Cutting. We developed a predictive computational model and validated it with respect to experimentally measured

temperatures in bone. The computational model aligns with biological data, and therefore represents a powerful tool for assessing existing and new methods to cut bone. For example, limitations in temperature measurement techniques currently make it difficult to accurately measure peak temperatures and temperature gradients in the immediate vicinity, e.g., within $500 \mu\text{m}$, of an osteotomy edge. Our computational model revealed that the most important biological changes, e.g., programmed cell death and bone remodeling, would primarily take place within this zone. It is precisely this region that serves as the main support for an implant; therefore, a model that helps clarify how osteotomy site preparation impacts bone viability should prove very useful in assessing the likelihood of osseointegration under different bone cutting methods.

Which Drilling Parameters Contribute Most to Heat Generation? The biological response to bone cutting is a summation of multiple parameters that include drill diameter, rotational speed, irrigation, and BV/TV. Some of these variables are under control of the clinician. For example, compared to drill diameter, the rotational speed of the drill and irrigation—have a significant impact on the size of the thermally affected zone. The amount of heat produced by cutting bone is proportional to rotational speed,

Fig. 6 Relative contribution of drilling parameters to the size of thermally affected zone: (a) drill diameter, (b) rotational speed, (c) irrigation, and (d) bone density, i.e., BV/TV. The results indicated that rotational speed, irrigation, and bone density are critically important in the extent of cell death resulting from osteotomy site preparation.

whereas irrigation acts as a heat sink, dissipating the heat from accumulating in bone. These data form the basis for the generalized recommendation that bone should be cut at the lowest possible rotational speed, and using copious irrigation [19,20]. It must be noted, however, that while irrigation acts as a coolant, it also removes blood, fibrin clots, and osteogenic bone chips from the osteotomy [21,22] and these bone chips are robustly osteogenic [23].

What is needed is a cutting tool that, at a low rotational speed, can still efficiently remove bone. A lower rotational speed would generate significantly less heat and therefore reduce the necessity of irrigation. Consequently, osteogenic bone chips would be maintained in the osteotomy where they could contribute to new bone formation around the implant.

Patient-Specific Parameters That Influence Heat Generation. Most heat generated by drilling arises because of the deformation and cutting of the bone. This helps explain why the BV/TV of bone has such a robust effect on heat transfer. When the BV/TV is higher, more bone must be deformed and cut to create an osteotomy. These results add to our understanding of why drilling trabecular bone (versus drilling cortical bone) has a less detrimental effect on the zone of death.

In conclusion, the predictions from heat transfer models for bone drilling provide insights into temperatures and thresholds that can cause osteocyte death, which can ultimately compromise implant stability. The findings also identify the causes of bone thermal damage and can help improve clinical drilling protocols and new drill designs.

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