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Review

A review of heat transfer in human tooth—Experimental characterization and mathematical modeling

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ABSTRACT

With rapid advances in modern dentistry, high-energy output instruments (*e.g.*, dental lasers and light polymerizing units) are increasingly employed in dental surgery for applications such as laser assisted tooth ablation, bleaching, hypersensitivity treatment and polymerization of dental restorative materials. Extreme high temperature occurs within the tooth during these treatments, which may induce tooth thermal pain (TTP) sensation. Despite the wide application of these dental treatments, the underlying mechanisms are far from clear. Therefore, there is an urgent need to better understand heat transfer (HT) process in tooth, thermally induced damage of tooth, and the corresponding TTP. This will enhance the design and optimization of clinical treatment strategies. This paper presents the state-of-the-art of the current understanding on HT in tooth, with both experimental study and mathematical modeling reviewed. Limitations of the current experimental and mathematical methodologies are discussed and potential solutions are suggested. Interpretation of TTP in terms of thermally stimulated dentinal fluid flow is also discussed.

Contents

Abbreviations: DEJ, dentine–enamel junction; DFF, dentinal fluid flow; DRMs, dental restorative materials; DTs, dentinal tubules; FEM, finite element method; HSHPs, high-speed hand-pieces; HSPs, heat-shock proteins; HT, heat transfer; IPTR, intrapulpal temperature rise; LPUs, light polymerizing units; PBF, pulpal blood flow; TC, thermal conductivity; TD, thermal diffusivity; TPs, thermophysical properties; TTP, tooth thermal pain.

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1. Background

Heat transfer (HT) in human tooth is a common process in both daily life [1,2] and clinical dentistry [3–6]. With advances in modern dentistry, instruments with high-energy output (*e.g.*, dental lasers [7,8], light polymerizing units (LPUs) [9–11] and high-speed hand-pieces (HSHPs) [12,13]) have been widely employed in dental treatments. Typical clinical examples of these treatments are laser assisted tooth ablation [7,14], caries detection and prevention [15,16], bleaching [17–19] and hypersensitivity treatment [20,21]. Table 1 lists the currently available therapies and the corresponding intrapulpal temperature rise (IPTR).

Heat generated within the tooth during the above-mentioned treatments can cause thermally induced damage of tooth “hard” components (enamel and dentine) [2,29–31] and “soft” components (dental pulp) [32–34]. Temperature rise may also induce tooth thermal pain (TTP) [35–37]. The lack of detailed knowledge concerning these tooth thermal behaviors has hindered the further design and optimization of treatment strategies for clinical applications. Therefore, there is an urgent need to investigate the thermal behavior of human teeth.

1.1. *Structure of human tooth*

Human tooth is a hard tissue with layered composite structure (*e.g.*, enamel, dentine, cementum and dental pulp layers [38]) of complex geometry. Enamel is a highly mineralized layer consisting of 96% mineral, with water and organic material accounting for the remaining 4%. Dentine is a mineralized connective tissue layer with an organic matrix of collagenous proteins [39,40], composed of ∼70% inorganic materials, ∼20% organic materials, and ∼10% water by weight. There are dentinal tubules (DTs) radiating from the pulp cavity to the exterior cementum or dentine–enamel junction (DEJ) [40]; see Fig. 1.

Human tooth is also a sensory tissue. The pulp is a soft connective tissue containing nerve fibers with diameter varying from 1 to 10 μm [41]. There are also sensory nerve endings in DTs [42], which penetrate approximately 100–150 μm into the tubules from the wall of pulp chamber [43]. These pulpal nerve terminals play an important role in sensing thermal stimulus [44].

1.2. *What induces HT in tooth?*

HT in tooth occurs in both daily life and dental surgery. The thermal environment of teeth during daily life varies over a wide range of temperatures (−5 to 76.3 °C) [2,46], Table 2.

Since its first introduction as a tool for patient care in 1964 [48], laser has been found to be an exciting and rewarding technology in dental surgery (*e.g.*, dental ablation [7,49], cosmetic tasks [50,51], dental caries prevention [15] and hypersensitivity treatment [21–22]) due to advantages such as directivity, pulsed mode ability, and monochromaticity [52]. However, there is a major concern with the overheating of enamel and dentine during tooth-laser interaction, which causes carbonization, melting and cracking of the enamel and dentine, as well as inflammation and necrosis of the pulp [30,31].

Dental restorative processes are used to decrease the polymerization periods of resin composite and minimize its polymerization shrinkage. During these processes, teeth are exposed to the radiation of LUPs (*e.g.*, halogen polymerization lamps or semiconductor diode light) and experience substantial temperature increase of 10–18 °C within the resin and adjacent tooth [56–58], 7.5–29 °C at DEJ [53], and 2–9 °C at pulpal chamber wall [53].

Treatments with HSHPs are performed in dental clinics for applications such as cavity preparation and deceased tissue

Table 1 – Typical thermal therapies and relative intrapulpal temperature rise (IPTR).

	Temperature rise (°C)	References
Laser assisted tooth ablation	2.3–24.7	[22,23]
Laser assisted caries prevention	1.2–4.0	[15]
Bleaching (without light/laser assisted)	0.1–1.1	[24]
Bleaching (with light/laser assisted)	1.1–16.0	[17,25]
Polymerization of dental restorative materials (DRMs)	2.9–7.8	[10,26,27]
HSHPs cavity preparation (without water, high load)	16.4–19.7	[12,28]
HSHPs cavity preparation (without water, low load)	7.1–9.5	[12,28]
HSHPs cavity preparation (with water, high load)	2.2–5.9	[12,28]
	10.0–11.7	
HSHPs cavity preparation (with water, low load)	−1.8 (drop in temperature) to 5.0	[12,28]

removal. Heat is generated by friction between the machinery (*e.g.*, driller) and tooth, resulting in IPTR [12,28,13]. This temperature rise depends on the pressure and speed of the hand-pieces [54] as well as the cooling methods used [12,55].

1.3. *Specialties of tooth thermal behavior*

Unlike HT in engineering materials, the thermal behavior of tooth is mainly a heat conduction process coupled with tooth physiological processes (*e.g.*, dentinal fluid flow (DFF) [35], pulpal blood flow (PBF) [56,57]). The thermophysical properties (TPs) of teeth vary between different layers (*e.g.*, enamel and dentine) [58] and depend on their microstructures. For instance, the thermal conductivity (TC) of human dentine decreases with increasing volume fraction of dentine tubules [59]. The flow of dentinal fluid in the dentine tubules upon heating (or cooling) can also enhance HT within the pulp. In addition, PBF rate increases when the intrapupal temperature rises above 42 °C [56] and decreases during cooling [57]. The perfused blood plays an important role in the thermoregulation of pulpal soft tissue [56,60], working as a heat sink under heating and as a heating source when subjected to cooling.

1.4. *Thermally induced damage*

Abrupt changes in tooth temperature can cause TTP and damage in tooth hard components (enamel and dentine), soft pulp tissue [33,34,32,61] and restorative interface (*e.g.*, interfacial debonding between the tooth and the DRMs) [62,63].

Enamel and dentine have different thermal and mechanical properties. For example, the thermal diffusivity (TD) and Young's modulus of enamel are ~2.5 and ~4 times larger than that of dentine, respectively [64,65]. The difference in these properties may result in thermal stresses and cracking within tooth when subjected to thermal stimulus [2,29], Fig. 2. In addition, heat generated during laser treatment may result in carbonization, melting and cracking of the tooth structure [30,31]. Thermal denature of dentine collagen often occurs in endodontics treatments, where the tooth root canal dentine may be exposed to a temperature of ~300 °C [66]. The denaturation temperatures of demineralized dentine matrix have been reported to be 65.6 °C, 148.5 °C, and 166.8–172.7 °C for demineralized dentine saturated with water, saturated with methanol, ethanol or acetone, and bonded with resin, respectively [66].

Dental pulp, which is responsible for the maintenance of tooth vitality, is vulnerable to heat. When the IPTR exceeds ~5.5 °C, irreversible pulpal damage (*e.g.,* cell death in pulp) will be induced [33,34]. However, pulp cells may survive such injuries [73,74]. This may be due to the increased synthesis of heat-shock proteins (HSPs) [67]. Amano et al. [68] found that a heat stimulation of 42 °C decreases dental pulp cell viability, whereas the HSPs would help recovering the pulp cell vitality.

Current restorative techniques are based on the adhesive properties of DRMs [69], which work normally in the temperature range of −5 to 55 °C [46]. Differences in TPs between tooth and DRMs facilitate the development of thermal stresses [63,70,71], with maximum stress developed on the bonding interfaces [72]. Together with stresses caused by mastication, these stresses can induce degeneration of the bonding interface and hence reduce the life-span of dental restorations [73–75].

Table 2 – Intraoral temperatures measured *in vivo* and calculated extreme temperature during consumption of hot beverage and food [47].

Temperature (°C)	Hot beverage	Hot food
Highest[a]	76.3	53.6
Mean maximum[b]	46.4	41.6
Calculated extreme[c]	61.4	50.2

[a] Highest temperature measured in one volunteer between the lower incisors.
[b] Mean maximum temperature ± standard deviation recorded by each electrode for all volunteers.
[c] Calculated extreme temperature obtained by adding two standard deviations to the mean maximum temperature measured *in vivo*.

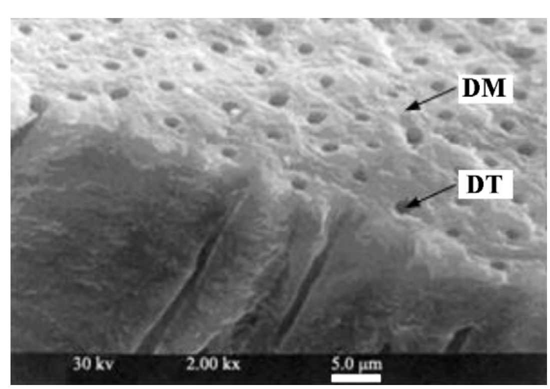

Fig. 1 – Scanning electron micrograph of dentine showing longitudinal and cross-sectional views of dentinal tubules (DTs) and dentinal matrix [45].

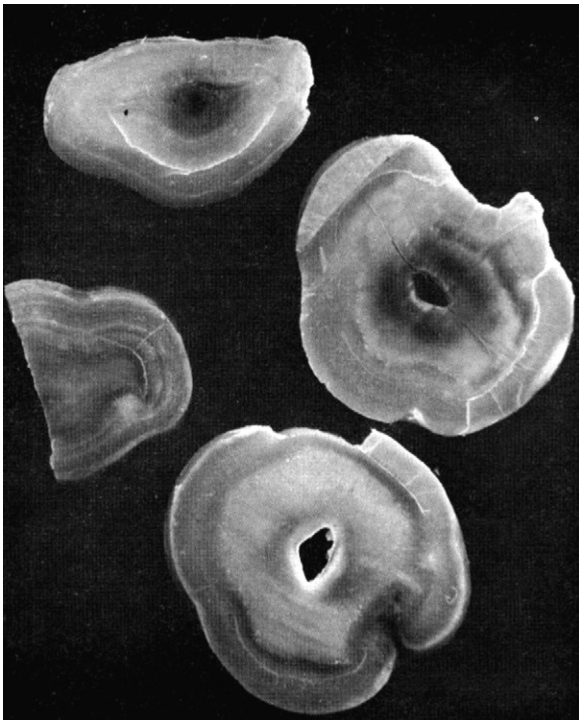

Fig. 2 – Sections of tooth after 7000 thermal cycles with crack depth and origin shown [29].

1.5. *Applications of tooth HT*

The study of HT in tooth and thermally related effects (*e.g.*, damage) is of great significance in modern dentistry, as summarized below.

(i) Precise temperature measurement in tooth may indicate the pulp vasculature and hence help in endodontic diagnosis [76].
(ii) *In vitro* investigation of HT in tooth under dental treatments can provide better approaches to evaluate the operation guidance, thereby reducing or preventing tooth damage so that a less empirical dental practice may be developed.
(iii) It can help in choosing and designing suitable DRMs with thermomechanical properties similar to toot, so that post-operative complications such as pain, marginal staining and caries may be avoided [77,78].
(iv) Tooth pain sensation is correlated with DFF [79].

A better understanding of HT phenomenon and thermally induced DFF in dental tubules will extend existing knowledge underlying TTP mechanisms and thus improve the relief methods for tooth pain sensation.

1.6. *Outline of this review*

In view of the lack of a thorough review of the current development in tooth HT, we present the state-of-the-art of existing studies on experimental characterization and mathematical modeling methodologies. The challenges associated with these methodologies and potential modifications essential for addressing these challenges are discussed. Interpretation of TTP in terms of heat stimulated DFF is also presented.

2. Experimental characterization of HT in tooth

Investigation of HT in tooth requires both experimental measurements [2,10,35,58,80,81] and mathematical modeling [53,82–84]. Mathematical models can supply estimates of information (*e.g.*, temperature and thermal stress) over the entire continuum of interest, while experimental characterizations can validate the mathematical models and determine the various material properties (*e.g.*, thermal, optical and mechanical) needed for theoretical modeling.

2.1. *Measurement of tooth TPs*

Knowledge of tooth TPs is crucial for accurate mathematical modeling and analysis of HT in tooth. The most important material properties for HT in tooth are tooth geometry, TC (k, $\mathrm{W\,m^{-1}\,K^{-1}}$), specific heat ($c_p$, $\mathrm{J\,kg^{-1}\,K^{-1}}$), and TD ($\alpha$, $\mathrm{m^2\,s^{-1}}$). In view of physics, TC indicates the ability of a material to conduct heat and is given as $k = (\Delta Q/\Delta t) \times (L/A \times \Delta T)$ where ΔQ (J) is the heat transmitted during time Δt (s) through a thickness L (m), in the direction normal to a surface of area A ($\mathrm{m^2}$), due to a temperature difference ΔT (°C). Specific heat is the energy required to increase a unit of temperature of a unit quantity of a substance. Thermal diffusivity $\alpha \equiv k/\rho c_p$ is the ratio of TC to volumetric heat capacity, ρ ($\mathrm{kg\,m^{-3}}$) being the density of the material.

Traditional HT instruments and methods have been widely used to characterize the TPs of tooth components, such as the thermal couples [29,65,85–88], infrared camera (IR) technique [89], flash laser method [60,90,91], differential scanning calorimeter (DSC) [59], and traditional calorimeter cup [65,92]. The reported data on human teeth are summarized in Table 3, which shows significant differences between reported results. For example, TC of human dentine was reported to be $\sim 0.11\,\mathrm{W\,m^{-1}\,K^{-1}}$ in one study [93], but to be $0.96–0.98\,\mathrm{W\,m^{-1}\,K^{-1}}$ in others [86,94]. Similarly, the reported TC of dentine covers a wide range of $0.36–0.88\,\mathrm{W\,m^{-1}\,K^{-1}}$ [59,64,85,95,96]. Whilst one study [91] reports the TD of enamel to be $2.27 \times 10^{-7}\,\mathrm{m^2\,s^{-1}}$, its was measured to be $\sim 4.20–4.69 \times 10^{-7}\,\mathrm{m^2\,s^{-1}}$ by others [64,85].

The significant discrepancy between the reported results of tooth TPs may be attributed to several challenges associated with the measurement: (i) The microstructure of tooth is heterogeneous and varies significantly from one sample to another (*e.g.*, teeth donators are of different ages, races, healthy conditions and of different kinds such as incisors and molars). For example, c_p of dentine is found to depend on the orientation of dentine tubules [59,64,92], with $873\,\mathrm{J\,kg^{-1}\,K^{-1}}$ and $1039–1181\,\mathrm{J\,kg^{-1}\,K^{-1}}$ in directions perpendicular and parallel to tubules, respectively. (ii) When thermal couples are used, the imperfect contact between the probe and tooth structure (*e.g.*, existence of air gap) increases the measurement error. (iii) The thermal, spatial and temporal resolutions of the IR camera and the radiation/reflectivity of irradiated

Table 3 – Thermophysical properties of human tooth.

Property	Tooth component	Values (method)	References
	Enamel[a]	4.69[f]	Brown et al. [64]
	Enamel	4.20[f]	Braden [85]
	Enamel[d]	2.27[g]	Panas et al. [91]
	Dentine[a]	1.87 ([b])[f]	Brown et al. [64]
		1.83 ([c])	
α Thermal diffusivity [$\times 10^{7}$ $m^2\,s^{-1}$]	Dentine[a]	1.99 ([b])[g]	Magalhães et al. [59]
		2.04–2.65 ([c])	
	Dentine	2.58[f]	Fukase et al. [97]
	Dentine	2.60[f]	Braden [85]
	Dentine[d]	1.92[g]	Panas et al. [91]
	Pulp[e]	1.5	de Vree et al. [98]
	Bulk value (enamel + detine)	3.49 (heating)[h]	Panas et al. [91]
		2.78 (cooling)	
	Enamel[a]	0.93[f]	Craig and Peyton [99]
	Enamel	0.92[f]	Braden [85]
	Dentine[a]	0.58 ([b])[f]	Craig and Peyton [99]
		0.57 ([c])	
	Dentine	0.11[f]	Johnson et al. [93]
	Dentine	0.96[f]	Lisanti and Zander [86]
	Dentine	0.98[f]	Simeral [94]
k Thermal conductivity [$W\,m^{-1}\,K^{-1}$]	Dentine	0.88[f]	Little et al. [96]
	Dentine[a]	0.36 ([b])[g]	Magalhães et al. [59]
		0.48–0.66 ([c])	
	Dentine	0.63[f]	Braden [85]
	Dentine	0.58 ([b])[f]	Craig and Peyton [99]
		0.57 ([c])	
	Dentine	0.40 (Root[c])[f]	Soyenkoff and Okun [95]
		0.45 (Crown[c])	
	Pulp[e]	0.63	de Vree et al. [98]
	Enamel[a]	0.71[i]	Brown et al. [64]
	Enamel	0.75[i]	Peyton and Simeral [92]
	Dentine[a]	1.59[i]	Brown et al. [64]
c_p Specific heat [$\times 10^{-3}$ $J\,kg^{-1}\,K^{-1}$]	Dentine[a]	0.87 ([b])[j]	Magalhães et al. [59]
		1.04–1.18 ([c])	
	Dentine	1.30[i]	Fukase et al. [97]
	Dentine	1.17[i]	Peyton and Simeral [92]
	Whole tooth[a]	1.26[i]	Brown et al. [64]
	Pulp[e]	4.2	de Vree et al. [98]
	Enamel(deciduous)	2.95	Berghash et al. [100]
	Enamel (permanent)	2.97	Manly et al. [101]
	Enamel[a]	2.80	Brown et al. [64]
	Dentine (deciduous)	2.18	Berghash et al. [100]
ρ Density [$\times 10^{-3}$ $kg\,m^{-3}$]	Dentine (permanent)	2.14	Manly et al. [101]
	Dentine[a]	1.96	Brown et al. [64]
	Dentine[a]	2.09–2.40	Magalhães et al. [59]
	Cementum	2.03	Berghash et al. [100]
	Whole tooth[a]	2.20	Brown et al. [64]
	Pulp[e]	1.00	de Vree et al. [98]

[a] From human third molar.
[b] Perpendicular to tubules.
[c] Parallel to tubules.
[d] From human premolar.
[e] Values taken from water according to de Vree et al. [99].
[f] Data acquired using thermal couple method.
[g] Data acquired using flash laser method.
[h] Data acquired using monotonic heating regime method.
[i] Data acquired using traditional calorimeter cup method.
[j] Data acquired using DSC method..

sample surface for flash laser are the main restrictions in acquiring precise TPs with these measurement techniques.

2.2. In vitro measurement of HT in tooth

Experimental characterization of tooth HT can be divided into two main types: *in vivo* and *in vitro*. *In vivo* characterization is desirable since it can reflect the active processes (*e.g.*, blood circulation in pulp chamber and fluid motion in DTs [28,35,32]) not captured *in vitro* [89,102–104]. However, there are several challenges associated with *in vivo* experimental investigations [89,103]. (i) Some internal features are difficult to capture. For example, experimental measurement of *in vivo* temperature change within tooth pulp is impractical [81]. (ii) Experimental procedures are restricted by patient tolerance especially when extreme thermal loading conditions as used in clinical pulp vitality testing [35,105,106] are involved. (iii) *In vivo* HT investigations of tooth are usually performed in animals [107–109]. The extension of these results to human teeth is questionable.

In view of the challenges of *in vivo* tests, *in vitro* tests are also widely used, *e.g.*, for studying HT during pulp sensitivity testing [81] and for safety evaluation of light-activated bleaching, HSHPs cavity preparation and laser irradiation [3,17,28]. Existing *in vitro* characterization of tooth HT can be divided into three categories depending on how the contribution of pulpal tissue to the tooth thermal behavior is modeled (*e.g.,* with empty pulpal chamber [25,81], with simulated pupal tissue [10,23,80], with consideration of microcirculation [27]).

In vitro HT studies have been carried out in tooth with empty pulp chamber (pulp soft tissue was extracted [25,54,81]). However, in this case, the measured temperature change on pulpal wall does not necessarily indicate the temperature change in pulp soft tissue [112], as the TC and heat capacity of teeth with empty pulp chambers are significantly different from those with filled chambers.

To simulate the transfer of heat between dentine–pulp junction (DPJ) and soft pulp tissue, Preiskorn et al. [80] placed a copper wire within the pulpal chamber through tooth root canal, with its protruding part spirally shaped to intensify the HT between the wire and a water bath. IPTR were measured both with and without considering the influence of pulp HT, with significant difference in the measured temperature profiles observed. Similarly, to mimic the vital dental pulp, Attrill et al. placed a 'pulp phantom' into the pulp chamber [23], whilst Hannig et al. [10] filled the pulp chamber with warm water to mimic HT through soft tissue in the pulp chamber.

Although *in vitro* measurements with simulated pulp tissue are more reliable than those carried out with an empty pulp, both approaches ignore the influence of PBF and the dense capillary network adjacent to the odontoblast layer in the vital pulp. This deficiency limits the direct application of the measurement data to the *in vivo* situation [27,76]. Daronch et al. [27] proposed an alternative way to measure IPTR by connecting one of the tooth roots via a small diameter tube to an infusion pump, with the pump designed to deliver water at a speed of 0.0125 ml/min that mimics the PBF rate. The tooth was then immersed into a water bath up to the cement–enamel junction. As blood supply in the pulp chamber and HT through pulp soft tissue and periodontal ligament are both considered, the measurements, although still *in vitro*, are more applicable in clinical situations.

2.3. Response of tooth to heat stimulus in terms of blood circulation and DFF

The pulp is highly vascularized [43], with arterial vessels entering through the apex of tooth roots and passing into the tooth crown. The microcirculation of blood in the dental pulp plays an important role in tooth thermal behavior, besides providing nutrition to the peripheral tissue. Histophysiological studies reveal that the PBF rate is almost constant when the IPTR lies within the range of 33° to 42 °C [111], increases significantly when the temperature rises above 42 °C [56], and decreases with decreasing temperature under cooling [57]. This feature contributes significantly to the thermoregulation of pulpal soft tissue [56,60]: the PBF works as a heat sink under heating and as a heating source under cooling. However, the influence of PBF on HT is considered negligible in clinical settings due to the low blood volume in the pulpal vasculature [110].

Thermal stimulation (hot or cold) of tooth causes temperature change within tooth [10,35,112] and may agitate tooth pain response [36]. Three theories for pulp nerves activation have been proposed, *i.e.*, the neural theory, the odontoblastic transduction theory and the hydrodynamic theory [113]. The neural theory assumes that direct thermal stimulus on the axons extending to the DEJ are responsible for TTP sensation. However, this assumption has been shown experimentally to be questionable, as the sensory response of teeth to thermal agitation occurred before a temperature change was detected in the pulpal wall (the locations of most sensory structures) [37]. Based on the observation that the human odontoblasts *in vitro* produce action potentials [114], the odontoblastic transduction theory proposes that odontoblasts may work as sensor cells that initiate tooth pain transmission. A strong argument against the transduction theory is the observation that dentin remains sensitive after local destruction of the odontoblasts [115,116]. At present, the most widely accepted theory is the hydrodynamic theory [115,117] which suggests that the free nerve endings located in the inner dentine and around the odontoblasts are activated by thermally induced DFF [36]. The DFF may be induced by the thermal expansion or contraction of fluid, pushing the fluid away from the pulp under cooling and toward the pulp under heating [118,119], or by the driving force of thermal expansion of enamel and dentine [35].

2.4. Influence of different factors on tooth HT

HT in tooth is affected by a number of factors such as the temperature dependent TPs, geometry of teeth components, types of DRMs, pulp circulation and blood perfusion, as discussed below.

2.4.1. Temperature dependent TPs

When the temperature change exceeds 10 °C, it is necessary to account for the temperature dependence of tooth TPs in modeling. The dependence of the tooth TC and TD on temperature may be approximated as:

$$k = k_0 + k_1 T, \quad \alpha = \alpha_0 + \alpha_1 T \tag{1}$$

where k_0, k_1, α_0 and α_1 represent the linear fit values to test data. However, k_1 and α_1 usually have values close to zero, implying that k and α are only slightly influenced by temperature, especially when the temperature is relatively low. Consequently, existing studies of tooth HT [120–123] often assume that the TPs are independent of temperature to simplify the mathematical analysis.

2.4.2. *Geometry of tooth*

Enamel has a unique microstructure consisting of aligned prisms/rods, which run approximately perpendicular from DEJ towards tooth surface [124]. This characteristic arrangement of prisms in enamel should have a significant influence on heat excursion in tooth. Besides, the presence of DTs is strongly correlated with the porosity of human dentine as well as its density and TC [59]. Furthermore, Pashley [125] and Komabayashi et al. [126] demonstrated that there is a marked increase in the number of DT, the tubular radius, and the total physical surface area available for diffusion in regions near the pulpal chamber wall, whereas a reverse situation was found in regions close to DEJ. This characteristic geometry of dentine facilitates a better transfer of heat into the pulp. Therefore, it is conceivable that the dentine material is adapting not only to enhance its endurance to thermal alterations but also to dissipate more heat toward the pulp [127].

2.4.3. *Influence of the thickness of dentine and enamel layers*

Enamel and dentine have relatively lower values in TC and diffusivity and thus protect the pulp from noxious thermal agitation. White et al. [128] and Paghdiwala et al. [129] proposed that dentine layer should be considered when analyzing tooth HT despite of its small thickness and that the prepared tooth with small amounts of remaining dentine thickness were more prone to temperature increases [59]. The effect of thermal stimulation on DFF has been investigated in tooth with exposed dentine [130], and it was shown that tooth with exposed dentine responded more quickly to thermal stimulation than intact tooth, due to direct transfer of heat to dentine in the absence of enamel.

2.4.4. *Dental restorative materials*

The influence of DRMs on tooth HT has been studied extensively due to its significance in clinical applications [53,122,123]. It has been established that the thermal performance of restored tooth differs significantly from that of intact ones, due to the different TPs of DRMs and tooth.

2.4.5. *Heat transfer due to PBF and DFF*

The PBF rate in the highly vascularized pulpal chamber [131,132] is correlated with peripheral temperature. For example, the PBF rate increases significantly when the temperature rises above 42 °C [56] and decreases under cooling [57]. Heat exchange between the perfused blood and peripheral tissue may significantly influence the thermal behavior of pulpal soft tissue [56,60]. In addition, heat can also stimulate DFF due to thermal deformation of DTs and the tissue fluid [35]. The outward (away from the pulp) or inward (toward the pulp) fluid movement under cold or hot stimulus [35] would inevitably cause the redistribution of pulp chamber temperature.

3. Mathematical modeling of HT in tooth

HT in restored or unrestored human tooth has been studied extensively, with emphasis placed on assessing the risk of unexpected IPTR during dental treatments. Tooth temperature has been measured invasively, which however only provides information of tissue temperature at discrete points [10,35,80]. Mathematical modeling, in this aspect, can address this issue by predicting temperature distribution and its evolution within tooth.

3.1. *Theoretical analysis*

Based on Pennes' bioheat transfer equation, various heat conduction problems solved by analytical means have been presented under different thermal boundary conditions [133–135]. The Pennes model is given as [136]:

$$\rho c \frac{\partial T}{\partial t} = \nabla(k\nabla T) + \rho_b c_b \omega_b (T_a - T) + q_{met} + q_{ext} \tag{2}$$

where ρ (kg m^{-3}) is tissue density; c (J kg^{-1} K^{-1}) is tissue specific heat; k (W m^{-1} K^{-1}) is tissue TC; ρ_b (kg m^{-3}) is blood density; c_b (J kg^{-1} K^{-1}) is blood specific heat; T_a (K) is blood temperature; ω_b (ml s^{-1}) is blood perfusion rate; T (K) is temperature of surrounding tissue; q_{met} (J) is metabolism heat generation; and q_{ext} (J) is heat generation due to external heating sources.

In clinics, prediction of heat distribution in tooth is vital for optimizing parameters of dental treatments. Analytical solutions to the heat conduction equation for laser interaction with material and biological tissues have been well established [137,138]. However, the solutions are usually obtained by simplifying the problem (*e.g.*, surface energy processes such as irradiation and convection are neglected), which put restrictions on modeling the spatial and temporal profiles of the incident pulse.

3.2. *Numerical simulation*

HT in tooth dependents on its geometry, material properties and *in vivo* biological function. Due to the complicated topological structure of tooth, introducing more than one of its components into an analytical model leads to intractable solutions, while numerical methods such as the finite element method (FEM) and the finite difference method (FDM) offer an alternative approach [53,82,139,140]. In mathematical modeling, the tooth is generally simplified as a system of geometrical areas containing only enamel, dentine (in intact tooth) and DRMs (in restored tooth), with HT between pulpal wall and pulp tissue ignored [122]. However, these simplifications lead to approximate results [110], as HT in dental pulp is considered important [53,121,141].

Based on the Fourier law of heat conduction, various models with or without the consideration of soft tissue (dental pulp) have been proposed for HT in restored/unrestored tooth. The temperature and thermal stress fields have been quantitatively calculated using 2D or three-dimensional (3D) FEM.

3.2.1. Modeling of normal heating

The thermal behavior of restored and unrestored tooth caused by food intake in daily life can be modeled as a cylindrical layered structure (*e.g.*, for human first molar during hot water drinking process [98]). With tooth taken as homogeneous and isotropic, one has [98]:

$$k_i\left[\frac{\partial}{\partial r}\left(r\frac{\partial T}{\partial r}\right)+\frac{\partial}{\partial z}\left(r\frac{\partial T}{\partial z}\right)\right]=\rho_i c_i r\frac{\partial T}{\partial t} \tag{3}$$

where i is the index of sub-layers ($i=1, 2, 3, \ldots, n$); r and z are the radial and axial coordinates; T is the temperature (which is a function of time t and location); k_i, c_i and ρ_i are the TC, specific heat and mass density in sub-layer i, respectively. To mimic convective loading, warm water has been used, with its temperature decreasing linearly from 60 °C to 35.2 °C [98]. The temperature predicted by the above model matched well with the clinical experimental measurement of Trowbridge et al. [37].

3.2.2. Modeling of heat distribution in tooth under irradiation of LPUs

The need of tooth restorations leads to the use of light-activated resin composites [9,142]. However, rapid polymerization necessitates the usage of increased LPUs output energy which may damage the pulp tissue [10,11]. Tunc [53] and Preiskorn et al. [80] developed models to analyze the potentially overheating of pulp when tooth is irradiated with clinical relevant LPUs. However, a constant heat flux thermal boundary condition was assumed, which limits the universality of applying the mathematical model from one type of LPU to another. Jakubinek et al. [82] have recently developed a mathematical model that considers the light intensity of LPUs. It was assumed that heat generated within the nth layer of restored tooth due to curing light, Q_n (J), can be described by the difference between the light intensity entering and leaving a slice of material of thickness d divided by that thickness, as:

$$Q_n=\begin{cases}\dfrac{I_0}{d}\left(e^{-(n-1)d/x}-e^{-(nd/x)}\right), & \text{for } 0\le t\le t_L\\ 0, & \text{otherwise}\end{cases} \tag{4}$$

where t_L (s) is the time duration of light exposure, I_0 (W/m^2) is the light intensity on tooth surface, and δ (m) is the 1/e penetration depth. The numerical predictions of Jakubinek et al. [82] appear to agree well with experimental measurements.

3.2.3. Modeling of laser interaction with tooth hard tissue

To analyze the heating of tooth hard tissue by laser, Zuerlein et al. proposed a one-dimensional (1D) heat conduction equation based on the Fourier conduction law and energy conservation equation [143,144], as:

$$\frac{\partial T(z,t)}{\partial t}=\frac{k}{\rho c_p}\frac{\partial^2 T(z,t)}{\partial z^2}+\frac{Q(z,t)}{\rho c_p} \tag{5}$$

where the source term $Q(z, t)$ is the temporally and spatially varying laser power input:

$$Q(z,t)=(1-R)F(t)\mu_a e^{-\mu_a z} \tag{6}$$

Here, z (m) is the depth from tooth front surface; R is the reflectance of laser beam of a specific wavelength on a specific tissue surface; and μ_a is the absorption coefficient; $F(t)$ (J m^{-2}) is the time dependent incident fluence of the laser pulse (assuming that absorption is consistent with the Beer–Lambert law). When a specific type of laser (*e.g.*, CO_2, Er:YSGG and Er:YAG lasers) is used, the important optical properties are the reflectance and the absorption coefficient, while scattering is negligible in highly absorbing materials (*e.g.*, tooth) [144,145]. Combination of Eqs. (5) and (6) leads to:

$$\frac{\partial T(z,t)}{\partial t}=\frac{k}{\rho c_p}\frac{\partial^2 T(z,t)}{\partial z^2}+\frac{(1-R)F(t)\mu_a e^{-\mu_a z}}{\rho c_p} \tag{7}$$

The distribution and evolution of tooth temperature can be determined from numerical solutions to Eq. (7). For thermal boundary conditions, the front and back surfaces of the tooth sample may be assumed thermally insulated so that the transient heat conduction equation can be solved in the axial direction only, since the absorption depth is much smaller than the laser beam diameter.

Clinically, the use of Nd:YAG laser ($\lambda = 1064$ nm) offers a series of advantages compared with CO_2 laser, attributed mainly to the limited crack formation when the former is used [146]. However, when the wavelength of the Nd:YAG laser lies within about 300–1000 nm, the scattering of non-pigmented tissue dominates over absorption. To account for the scattering and its anisotropy, Moriyama et al. [140,147] modified Eq. (6) as:

$$Q(z,t)=\mu_a I_o(t)\exp(-\mu_{eff} z) \tag{8}$$

where z is the depth from tooth front surface; I_o is the power density; and μ_{eff} is the effective attenuation coefficient defined as:

$$\mu_{eff}=\sqrt{3\mu_a[\mu_a+(1-g)]\mu_s} \tag{9}$$

In the above equation, μ_s is the scattering coefficient and g is the anisotropy parameter.

The reflectivity of enamel and dentine accounts for about 8.8–49.4% of the energy loss on the irradiated surface [145]. However, Moriyama et al.'s model neglected the reflectance of laser beam on tissue surface and hence its clinical application is limited. Notice also that in both studies of Zuerlein et al. [143,144] and Moriyama et al. [140,147], heat radiation and convection from tissue surface were neglected which may be true in many cases. However, when the tissues are heated to high temperatures (*e.g.*, dental ablation and caries treatment by laser), heat radiation and convection are particularly important as these are the processes allowing for cooling during and after the absorption and diffusion of laser energy. According to Sagi-Dolev et al. [148], neglecting the surface processes would lead to over-estimated temperature rises in tooth. Hence, they proposed a model that accounts for surface heat dissipation through convection and radiation by introducing two extra boundary conditions:

$$q_h=H(T-T_\infty),\quad q_r=\varepsilon\sigma(T^4-T_\infty^4) \tag{10}$$

where q_h is the convection heat flux on tissue surface; H is the surface convection coefficient; T is the temperature of tissue surface; T_∞ is the surrounding temperature; q_r is the heat radiation of tissue surface; ε is the emissivity; and σ is the Stephan-Boltzmann constant.

Nevertheless, the reflectivity of laser beam on tooth enamel surface was ignored by Sagi-Dolev et al. [148]. As previously mentioned, the reflectivity of enamel and dentine accounts for about 8.8–49.4% of energy loss on the laser irradiated surface [145]. This may explain why the model predictions of Sagi-Dolev et al. [148] deviate from the *in vitro* measurements.

Therefore, a mathematical model that accounts for all the important surface energy processes (reflection, radiation and convection) is needed for more accurate predictions.

3.3. *Limitations with existing models*

In spite of the extensive research thus far, significant discrepancy still exists between experimental measurements and mathematical modeling. For example, Tunc [53] reported that the temperature predicted by existing numerical models could not be used directly as temperature changes *in vivo* because the models did not consider the cooling effect of blood perfusion in pulp chamber and DFF. The discrepancy indicates that some of the assumptions made when developing these models are not realistic. The factors that prohibit the development of a sound and conclusive mathematical model include: (1) the physical properties of tooth vary in different teeth (incisor, canine, molar) and different donors including ages, sex and races. Even for a single tooth, its physical properties differ from one layer to another and are anisotropic as well as inhomogeneous in each layer [45,149]. While, in the existing models, teeth are treated as layered materials and assumed to be homogenous, isotropic and linearly elastic in each layer. Such simplifications lead to an inaccurate presentation of tooth HT and hence prohibit the predictions to be directly used as guidance for clinical applications. (2) The current models assume that heat conduction occurs only between the pulpal wall and the pulp, with the influence of PBF and DFF neglected. (3) The thermal boundary conditions in tooth especially the part that under the gum is too complex to be determined. In the exiting models, these boundary conditions have been much simplified (*e.g.*, using oral temperature as the constant temperature) [60,138].

4. Future work

Characterizing how heat is transferred across tooth is of great importance which can contribute to a variety of dental medical applications as well as the fundamental understanding of tooth thermal pain. In this respect, the present review identifies the following topics for further research.

The lack of reliable and comprehensive data concerning tooth TPs causes great difficulties in the mathematical modeling of HT in tooth, which also impedes the development of DRMs possessing “tooth like” TPs to avoid post complications. It is therefore of great importance to systematically re-evaluate the TPs of teeth, where a number of factors should be properly considered, including the gender and age of the donors, the type of tooth (incisor, canine and molar), *etc*.

When studying the thermal behavior of biological tissue, thermomechanical coupling should be taken into account as mechanical loadings often accompany thermal stimuli [150,151]. For example, excessive heat deposition in tooth will cause significant tooth thermal deformation [129,152] and DFF [35]. The resulting thermal stresses may cause cracking in tooth [2,29,63] as well as marginal leakage of dental restorations interfaces [63]. More importantly, the thermally induced DFF and tooth deformation is critical for the study of TTP mechanisms [36,108].

Current mathematical models, although very helpful, still need further improvement. Previous models [55,59,106,137,138] have assumed that each layer of a tooth is made of isotropic material, and have neglected its biological features (both structure and function) such as: (1) biologically organized enamel prisms running from DEJ to enamel surface; (2) organized system of DTs filled with tissue fluid that run complexly through the tissue; and (3) cooling/heating effect of PBF in the capillarity. Improved modeling with consideration of the above-mentioned factors should provide a more accurate prediction of tooth HT which, eventually, may be practically used as guidance for clinical applications. Furthermore, experimental measurements are needed to check the validity of model predictions.

How do the external thermal stimuli (cold or hot) cause tooth pain sensation? Also, why outward fluid flow in DTs (away from the pulp that can be triggered by cold stimulus) is more efficient in evoking axonal activation than inward flow (toward the pulp that can be triggered by hot stimulus)? [108]. The answer may be obtained through a comprehensive study of thermal-relative DFF in DTs [36]. Due to the presence of mechanosensitive ion channels on pulpal neurons [153], the shearing force induced by DFF may be responsible for the activation of iron channels and consequently induce tooth pain sensation [40,153]. A holistic mathematical model capable of describing the complicated processes of heat stimulated pain sensation in biological tissue has been established [154–156], which may be applicable in the study of TTP.

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REFERENCES

[1] Boehm RF. Thermal environment of teeth during open-mouth respiration. J Dent Res 1972;51:75–8.
[2] Jacobs HR, Thompson RE, Brown WS. Heat transfer in teeth. J Dent Res 1973;52:248–52.
[3] Ana PA, Blay A, Miyakawa W, Zezell DM. Thermal analysis of teeth irradiated with Er,Cr YSGG at low fluences. Laser Phys Lett 2007;4:827–34.

[4] Uhl A, Volpel A, Sigusch BW. Influence of heat from light curing units and dental composite polymerization on cells in vitro. J Dent 2006;34:298–306.
[5] Martins GR, Cavalcanti BN, Rode SM. Increases in intrapulpal temperature during polymerization of composite resin. J Prosthet Dent 2006;96:328–31.
[6] Vanderlei AD, Borges ALS, Cavalcanti BN, Rode SM. Ultrasonic versus high-speed cavity preparation: analysis of increases in pulpal temperature and time to complete preparation. J Prosthet Dent 2008;100:107–9.
[7] Fahey M, Onyejekwe O, Lawrence Mason H, Mitra K. Precise dental ablation using ultra-short-pulsed 1552 nm laser. Int J Heat Mass Trans 2008;51:5732–9.
[8] Spitz SD. Lasers in prosthodontics: clinical realities of a dental laser in a prosthodontic practice. Alpha Omegan 2008;101:188–94.
[9] Uhl A, Mills RW, Jandt KD. Polymerization and light-induced heat of dental composites cured with LED and halogen technology. Biomaterials 2003;24:1809–20.
[10] Hannig M, Bott B. In-vitro pulp chamber temperature rise during composite resin polymerization with various light-curing sources. Dent Mater 1999;15:275–81.
[11] Bouillaguet S, Caillot G, Forchelet J, Cattani-Lorente M, Wataha JC, Krejci I. Thermal risks from LED- and high-intensity QTH-curing units during polymerization of dental resins. J Biomed Mater Res B: Appl Biomater 2005;72B:260–7.
[12] Cavalcanti BN, Otani C, Rode SM. High-speed cavity preparation techniques with different water flows. J Prosthet Dent 2002;87:158–61.
[13] Cavalcanti BN, Lage-Marques JL, Rode SM. Pulpal temperature increases with Er:YAG laser and high-speed handpieces. J Prosthet Dent 2003;90:447–51.
[14] Rode AV, Gamaly EG, Luther-Davies B, Taylor BT, Graessel M, Dawes JM, et al. Precision ablation of dental enamel using a subpicosecond pulsed laser. Aust Dent J 2003;48: 233–9.
[15] Zezell DM, da Ana PA, Ribeiro AC, Bachmann L, Freitas AZ, Vieira ND. Lasers in caries diagnosis and prevention. Int J Appl Electromagn Mech 2007;25:627–33.
[16] Rodrigues LKA, dos Santos MN, Pereira D, Assaf AV, Pardi V. Carbon dioxide laser in dental caries prevention. J Dent 2004;32:531–40.
[17] Eldeniz AU, Usumez A, Usumez S, Ozturk N. Pulpal temperature rise during light-activated bleaching. J Biomed Mater Res B: Appl Biomater 2005;72B:254–9.
[18] Joiner A. The bleaching of teeth: a review of the literature. J Dent 2006;34:412–9.
[19] Buchalla W, Attin T. External bleaching therapy with activation by heat, light or laser—a systematic review. Dent Mater 2007;23:586–96.
[20] Lee BS, Chang CW, Chen WP, Lan WH, Lin CP. In vitro study of dentin hypersensitivity treated by Nd:YAP laser and bioglass. Dent Mater 2005;21:511–9.
[21] Kimura Y, Wilder-Smith P, Yonaga K, Matsumoto K. Treatment of dentine hypersensitivity by lasers: a review. J Clin Periodontol 2000;27:715–21.
[22] Geraldo-Martins VR, Tanji EY, Wetter NU, Nogueira RD, Eduardo CP. Intrapulpal temperature during preparation with the Er:YAG laser: an in vitro study. Photomed Laser Surg 2005;23:182–6.
[23] Attrill DC, Davies RM, King TA, Dickinson MR, Blinkhorn AS. Thermal effects of the Er:YAG laser on a simulated dental pulp: a quantitative evaluation of the effects of a water spray. J Dent 2004;32:35–40.
[24] Luk K, Tam L, Hubert M. Effect of light energy on peroxide tooth bleaching. J Am Dent Assoc 2004;135: 194–201.
[25] Sulieman M, Addy M, Rees JS. Surface and intra-pulpal temperature rises during tooth bleaching: an in vitro study. Br Dent J 2005;199:37–40.
[26] Hussey DL, Biagioni PA, Lamey PJ. Thermographic measurement of temperature change during resin composite polymerization in vivo. J Dent 1995;23:267–71.
[27] Daronch M, Rueggeberg FA, Hall G, De Goes MF. Effect of composite temperature on in vitro intrapulpal temperature rise. Dent Mater 2007;23:1283–8.
[28] Ozturk B, Usumez A, Ozturk AN, Ozer F. In vitro assessment of temperature change in the pulp chamber during cavity preparation. J Prosthet Dent 2004;91:436–40.
[29] Brown WS, Jacobs HR, Thompson RE. Thermal fatigue in teeth. J Dent Res 1972;51:461–7.
[30] Wigdor HA, Walsh JT, Featherstone JDB, Visuri SR, Fried D, Waldvogel JL. Lasers in dentistry. Lasers Surg Med 1995;16:103–33.
[31] Frentzen M, Koort HJ. Lasers in dentistry: new possibilities with advancing laser technology? Int Dent J 1990;40:323–32.
[32] Sulieman M, Rees JS, Addy M. Surface and pulp chamber temperature rises during tooth bleaching using a diode laser: a study in vitro. Br Dent J 2006;200:631–4.
[33] Pohto M, Scheinin A. Microscopic observations on living dental pulp. II. The effect of thermal irritants on the circulation of the pulp in the lower rat incisor. Acta Odontol Scand 1958;16:315–27.
[34] Zach L, Cohen G. Pulp response to externally applied heat. Oral Surg Oral Med Oral Pathol 1965;19:515–30.
[35] Linsuwanont P, Palamara JEA, Messer HH. An investigation of thermal stimulation in intact teeth. Arch Oral Biol 2007;52:218–27.
[36] Linsuwanont P, Versluis A, Palamara JE, Messer HH. Thermal stimulation causes tooth deformation: a possible alternative to the hydrodynamic theory? Arch Oral Biol 2008;53:261–72.
[37] Trowbridge HO, Franks M, Korostoff E, Emling R. Sensory response to thermal stimulation in human teeth. J Endod 1980;6:405–12.
[38] Sumikawa DA, Marshall GW, Gee L, Marshall SJ. Microstructure of primary tooth dentin. Pediatr Dent 1999;21:439–44.
[39] Carda C, Peydro A. Ultrastructural patterns of human dentinal tubules, odontoblasts processes and nerve fibres. Tissue Cell 2006;38:141–50.
[40] Pashley DH. Dynamics of the pulpo-dentin complex. Crit Rev Oral Biol Med 1996;7:104–33.
[41] Graf W, Bjorlin G. Diameters of nerve fibers in human tooth pulps. J Am Dent Assoc 1951;43:186–93.
[42] Frank RM. Ultrastructure of human dentine 40 years ago: progress and perspectives. Arch Oral Biol 1999;44:979–84.
[43] Hildebrand C, Fried K, Tuisku F, Johansson CS. Teeth and tooth nerves. Prog Neurobiol 1995;45:165–7.
[44] Anderson DJ, Hannam AG, Matthews B. Sensory mechanisms in mammalian teeth and their supporting structures. Physiol Rev 1970;50:171–95.
[45] Grayson W, Marshall J. Dentin: microstructure and characterization. Ouintmence Int 1993;24:606–17.
[46] Haskan H, Boyraz T, Kilicarslan MA. Investigation of thermal stresses in dental restoration by mathematical method. J Eur Ceram Soc 2007;27:899–902.
[47] Osnat F. Temperature changes in dental implants following exposure to hot substances in an ex vivo model. Clin Oral Implants Res 2008;19:629–33.
[48] Goldman L, Goldman B, Hornby P, Meyer R. Impact of the laser on dental caries. Nature 1964;203:417.
[49] Majaron B. Heat diffusion and debris screening in Er:YAG laser ablation of hard biological tissues. Appl Phys B: Lasers Opt 1998;66:479–87.

[50] Convissar RA. Lasers and light amplification in dentistry. Dent Clin North Am 2000;44:717–994.
[51] Keller OR, Weber FE, Gratz KW, Baltensperger MM, Eyrich GK. Laser-induced temperature changes in dentine. J Clin Laser Med Surg 2003;21:75–381.
[52] Ishikawa I, Aoki A, Takasaki AA. Potential applications of erbium:YAG laser in periodontics. J Periodontal Res 2004;39:275–85.
[53] Tunc EP. Finite element analysis of heat generation from different light-polymerization sources during cementation of all-ceramic crowns. J Prosthet Dent 2007;97: 366–74.
[54] Hatton JF, Holtzmann DJ, Ferrillo PJ, Stewart GP. Effect of handpiece pressure and speed on intrapulpal temperature rise. Am J Dent 1994;7:108–10.
[55] Lloyd BA, Rich JA, Brown WS. Effect of cooling techniques on temperature control and cutting rate for high-speed dental drills. J Dent Res 1978;57:675–84.
[56] Raab WH. Temperature related changes in pulpal microcirculation. Proc Finn Dent Soc 1992;88(Suppl. 1):469–79.
[57] Goodis HE, Winthrop V, White JM. Pulpal responses to cooling tooth temperatures. J Endod 2000;26:263–7.
[58] Kishen A, Ramamurty U, Asundi A. Experimental studies on the nature of property gradients in the human dentine. J Biomed Mater Res A 2000;51:650–9.
[59] Magalhães MF, Ferreira RAN, Grossi PA, Andrade RM. Measurement of thermophysical properties of human dentin: effect of open porosity. J Dent 2008;36:588–94.
[60] Raab WH, Muller H. Temperature-dependent changes in the microcirculation of the dental pulp. Dtsch Zahnarztl Z 1989;44:496–7.
[61] Llyod BA, McGinley MB, Brown WS. Thermal stress in teeth. J Dent Res 1978;57:571–82.
[62] Li H, Burrow MF, Tyas MJ. The effect of thermocycling regimens on the nanoleakage of dentin bonding systems. Dent Mater 2002;18:189–96.
[63] Lee SY, Chiang H-C, Huang H-M, Shih Y-H, Chen H-C, Dong D-R, et al. Thermo-debonding mechanisms in dentin bonding systems using finite element analysis. Biomaterials 2001;22:113–23.
[64] Brown WS, Dewey WA, Jacobs HR. Thermal properties of teeth. J Dent Res 1970;49:752–5.
[65] Xu HH, Smith DT, Jahanmir S, Romberg E, Kelly JR, Thompson VP, et al. Indentation damage and mechanical properties of human enamel and dentin. J Dent Res 1998;77:472–80.
[66] Armstrong SR, Jessop JLP, Winn E, Tay FR, Pashley DH. Effects of polar solvents and adhesive resin on the denaturation temperatures of demineralised dentine matrices. J Dent 2008;36:8–14.
[67] Szymanska Z, Zylicz M. Mathematical modeling of heat shock protein synthesis in response to temperature change. J Theor Biol 2009;259:562–9.
[68] Amano T, Muramatsu T, Amemiya K, Kubo K, Shimono M. Responses of rat pulp cells to heat stress in vitro. J Dent Res 2006;85:432–5.
[69] De Munck J, Van Landuyt K, Peumans M, Poitevin A, Lambrechts P, Braem M, et al. A critical review of the durability of adhesion to tooth tissue: methods and results. J Dent Res 2005;84:118–32.
[70] Sidhu SK, Carrick TE, McCabe JF. Temperature mediated coefficient of dimensional change of dental tooth-colored restorative materials. Dent Mater 2004;20: 435–40.
[71] Sideridou I, Achilias DS, Kyrikou E. Thermal expansion characteristics of light-cured dental resins and resin composites. Biomaterials 2004;25:3087–97.
[72] Price RB, Derand T, Andreou P, Murphy D. The effect of two configuration factors, time, and thermal cycling on resin to dentin bond strengths. Biomaterials 2003;24:1013–21.
[73] Arola D, Huang MP. The influence of simultaneous mechanical and thermal loads on the stress distribution in molars with amalgam restorations. J Mater Sci: Mater Med 2000;11:133–40.
[74] Nikaido T, Kunzelmann KH, Chen H, Ogata M, Harada N, Yamaguchi S, et al. Evaluation of thermal cycling and mechanical loading on bond strength of a self-etching primer system to dentin. Dent Mater 2002;18:269–75.
[75] Cardoso PE, Placido E, Moura SK. Microleakage of four simplified adhesive systems under thermal and mechanical stresses. Am J Dent 2002;15:164–8.
[76] Jafarzadeh H, Udoye CI, Kinoshita J-I. The application of tooth temperature measurement in endodontic diagnosis: a review. J Endod 2008;34:1435–40.
[77] Murray PE, Windsor LJ, Smyth TW, Hafez AA, Cox CF. Analysis of pulpal reactions to restorative procedures, materials, pulp capping, and further therapies. Crit Rev Oral Biol Med 2002;13:509–20.
[78] Torstenson B, Brannstrom M. Contraction gap under composite resin restorations: effect of hygroscopic expansion and thermal stress. Oper Dent 1988;13:24–31.
[79] Charoenlarp P, Wanachantararak S, Vongsavan N, Matthews B. Pain and the rate of dentinal fluid flow produced by hydrostatic pressure stimulation of exposed dentine in man. Arch Oral Biol 2007;52:625–31.
[80] Preiskorn M, Zmuda S, Trykowski J. In vitro investigations of the heat transfer phenomena in human tooth. Acta Bioeng Biomech 2003;5:23–36.
[81] Linsuwanont P, Palamara JE, Messer HH. Thermal transfer in extracted incisors during thermal pulp sensitivity testing. Int Endod J 2008;41:204–10.
[82] Ana PA, Velloso WF, Zezell DM. Three-dimensional finite element thermal analysis of dental tissues irradiated with Er,Cr:YSGG laser. Rev Sci Instrum 2008;79:9.
[83] Jakubinek MB, O'Neill C, Felix C, Price RB, White MA. Temperature excursions at the pulp–dentin junction during the curing of light-activated dental restorations. Dent Mater 2008;24:1468–76.
[84] Mihai O, Scarlat F, Mihailescu IN. Temperature field modeling during multi-modes CO_2 laser irradiation of human enamel. Optics Laser Technol 2007;39:537–40.
[85] Braden M. Heat conduction in teeth and the effect of lining materials. J Dent Res 1964;43:315–22.
[86] Lisanti VF, Zander HA. Thermal conductivity of dentin. J Dent Res 1950;29:493–7.
[87] Phillips R, Reinking RH, Phillips LJ. Thermal conductivity of dental cement-a method and determinations for three commercial materials. J Dent Res 1954;33:511–8.
[88] Tibbetts VR, Schnell RJ, Swartz ML, Phillips RW. Thermal diffusion through amalgam and cement bases: comparison of in vitro and in vivo measurements. J Dent Res 1976;55:441–51.
[89] Panas AJ, Preiskorn M, Dabrowski M, Zmuda S. Validation of hard tooth tissue thermal diffusivity measurements applying an infrared camera. Infrared Phys Technol 2007;49:302–5.
[90] Parker WJ, Jenkins RJ, Butler CP, Abbott GL. Flash method of determining thermal diffusivity, heat capacity and thermal conductivity. J Appl Phys 1961;32:1679–84.
[91] Panas AJ, Żmuda S, Terpiłowski J, Preiskorn M. Investigation of the thermal diffusivity of human tooth hard tissue. Int J Thermophys 2003;24:837–48.
[92] Peyton FA, Simeral WG. The specific heat of tooth structure. University of Michigan School of Dentistry, Alumni Bull 1954:33.

[93] Johnson RJ, Phillips LJ, Phillips RW. An improved method for measuring the coefficient of thermal conductivity of dental cement. Am Dent Assoc 1956;53:577–83.
[94] Simeral WG. Thermal conductivity of dental materials, microfilm of papers presented at the annual IADR. Dental materials group meeting. J Dent Res 1951;30:499.
[95] Soyenkoff BC, Okun JH. Thermal conductivity measurements of dental tissues with the aid of thermistors. J Am Dent Assoc 1958;57:23–30.
[96] Little PAG, Wood DJ, Bubb NL, Maskill SA, Mair LH, Youngson CC. Thermal conductivity through various restorative lining materials. J Dent 2005;33:585–91.
[97] Fukase Y, Saitoh M, Kaketani M, Ohashi M, Nishiyama M. Thermal coefficients of paste-paste type pulp capping cements. Dent Mater J 1992;11:189–96.
[98] de Vree JH, Spierings TA, Plasschaert AJ. A simulation model for transient thermal analysis of restored teeth. J Dent Res 1983;62:756–9.
[99] Craig RG, Peyton FA. Thermal conductivity of tooth structure, dental cements and amalgam. J Dent Res 1961;40:411–7.
[100] Berghash SR, Hodge HC. Density and refractive index studies of dental hard tissues. III. Density distribution of deciduous enamel and dentin. J Dent Res 1940;19:487–95.
[101] Manly RS, Hodge HC, Ange LE. Density and refractive index studies of dental hard tissues. II. Density distribution curves. J Dent Res 1939;18:203–11.
[102] Laurell KA, Carpenter W, Daugherty D, Beck M. Histopathologic effects of kinetic cavity preparation for the removal of enamel and dentin. An in vivo animal study. Oral Surg Oral Med Oral Pathol Oral Radiol Endod 1995;80:214–25.
[103] Spierings TA, Peters MC, Bosman F, Plasschaert AJ. Verification of theoretical modeling of heat transmission in teeth by in vivo experiments. J Dent Res 1987;66:1336–9.
[104] Ciucchi B, Bouillaguet S, Holz J, Pashley D. Dentinal fluid dynamics in human teeth, in vivo. J Endod 1995;21:191–4.
[105] White JH, Cooley RL. A quantitative evaluation of thermal pulp testing. J Endod 1977;3:453–7.
[106] Fuss Z, Trowbridge HO, Bender IB, Rickoff B, Sorin S. Assessment of reliability of electrical and thermal pulp testing agents. J Endod 1986;12:301–5.
[107] Jyvasjarvi E, Kniffki KD. Cold stimulation of teeth: a comparison between the responses of cat intradental A delta and C fibres and human sensation. J Physiol: Lond 1987;391:193–207.
[108] Vongsavan N, Matthews B. The relationship between the discharge of intradental nerves and the rate of fluid flow through dentine in the cat. Arch Oral Biol 2007;52:640–7.
[109] Rizoiu I, Kohanghadosh F, Kimmel AI, Eversole LR. Pulpal thermal responses to an erbium, chromium:YSGG pulsed laser hydrokinetic system. Oral Surg Oral Med Oral Pathol Oral Radiol Endodont 1998;86:220–3.
[110] Chang JC, Wilder-Smith P. Laser-induced thermal events in empty and pulp-filled dental pulp chambers. Lasers Surg Med 1998:22.
[111] Newton AV, Mumford JM. Thermal flow through human teeth. Dent Pract Dent Rec 1970;21:84–6.
[112] Brown AC, Goldberg MP. Surface temperature and temperature gradients of human teeth in situ. Arch Oral Biol 1966;11:973–82.
[113] Sessle BJ. Invited review: the neurobiology of facial and dental pain: present knowledge, future directions. J Dent Res 1987;66:962–81.
[114] Allard B, Magloire H, Couble ML, Maurin JC, Bleicher F. Voltage-gated sodium channels confer excitability to human odontoblasts: possible role in tooth pain transmission. J Biol Chem 2006;281:29002–10.
[115] Brännström M. The hydrodynamic theory of dentinal pain: sensation in preparations, caries, and the dentinal crack syndrome. J Endod 1986;12:453–7.
[116] Hirvonen TJ, Narhi MV. The effect of dentinal stimulation on pulp nerve function and pulp morphology in the dog. J Dent Res 1986;65:1290–3.
[117] Brännström M, Linden LA, Astrom A. The hydrodynamics of the dental tubule and of pulp fluid. A discussion of its significance in relation to dentinal sensitivity. Caries Res 1967;1:310–7.
[118] Brännström M, Johnson G. Movements of the dentine and pulp liquids on application of thermal stimuli. An in vitro study. Acta Odontol Scand 1970;28:59–70.
[119] Hashimoto M, Ito S, Tay FR, Svizero NR, Sano H, Kaga M. Fluid movement across the resin-dentin interface during and after bonding. J Dent Res 2004;83:843–8.
[120] Motawea I. 3-D finite element analysis of transient heat transfer and thermal stress in a crowned mandibular first molar tooth. In: 2007 international conference on thermal issues in emergingtechnologies – theory and applications. 2007. p. 131–5.
[121] Toparli M, Aykul H, Sasaki S. Temperature and thermal stress analysis of a crowned maxillary second premolar tooth using three-dimensional finite element method. J Oral Rehabil 2003;30:99–105.
[122] Toparli M, Sasaki S. Finite element analysis of the temperature and thermal stress in a postrestored tooth. J Oral Rehabil 2003;30:921–6.
[123] Gungor MA, Kucuk M, Dundar M, Karaoglu C, Artunc C. Effect of temperature and stress distribution on all-ceramic restorations by using a three-dimensional finite element analysis. J Oral Rehabil 2004;31:172–8.
[124] Hoffman R, Gross LEO. Microstructure of dental enamel. I. Organization and contour of prisms. J Dent Res 1967;46:1444–55.
[125] Pashley DH. Smear layer: physiological considerations. Oper Dent 1984;3:13–29.
[126] Komabayashi T, Nonomura G, Watanabe LG, Marshall Jr GW, Marshall SJ. Dentin tubule numerical density variations below the CEJ. J Dent 2008;36:953–8.
[127] Kishen A, Asundi A. Investigations of thermal property gradients in the human dentine. J Biomed Mater Res 2001;55:121–30.
[128] White JM, Goodis HE, Rose CM, Daniels TE. Effects of Nd:YAG laser on pulps of extracted human teeth. J Dent Res 1991.
[129] Paghdiwala AF, Vaidyanathan TK, Paghdiwala MF. Evaluation of erbium:YAG laser radiation of hard dental tissues: analysis of temperature changes, depth cuts and structural effects. Scanning Microsc 1993;7: 989–97.
[130] Chidchuangchai W, Vongsavan N, Matthews B. Sensory transduction mechanisms responsible for pain caused by cold stimulation of dentine in man. Arch Oral Biol 2007;52:154–60.
[131] Boyer CC, Neptune CM. Patterns of blood supply to teeth and adjacent tissues. J Dent Res 1962;41:158–71.
[132] Provenza DV. The blood vascular supply of the dental pulp with emphasis on capillary circulation. Circ Res 1958;6:213–8.
[133] Xu F, Lu TJ, Seffen KA. Biothermomechanics of skin tissues. J Mech Phys Solids 2008;56:1852–84.
[134] Deng Z-S, Liu J. Analytical study on bioheat transfer problems with spatial or transient heating on skin surface or inside biological bodies. J Biomech Eng 2002;124:638–49.
[135] Xu F, Seffen KA, Lu TJ. Non-Fourier analysis of skin biothermomechanics. Int J Heat Mass Trans 2008;51:2237–59.

[136] Pennes HH. Analysis of tissue and arterial blood temperatures in the resting human forearm. J Appl Physiol 1948;1:93–122.
[137] Bechtel JH. Heating of solid targets with laser pulses. J Appl Phys 1975;46:1585–93.
[138] McKenzie AL. Physics of thermal processes in laser-tissue interaction. Phys Med Biol 1990;35:1175–209.
[139] Romeed SA, Fok SL, Wilson NA. Comparison of 2D and 3D finite element analysis of a restored tooth. J Oral Rehabil 2006;33:209–15.
[140] Moriyama EH, Zangaro RA, Lobo PDdC, Villaverde AGJB, Watanabe I-S, Pacheco MTT, et al. Mathematical simulation of the thermal diffusion in dentine irradiated with Nd:YAG laser using finite difference method. In: Lasers in dentistry VIII. San Jose, CA, USA: SPIE; 2002.
[141] Spierings TA, de Vree JH, Peters MC, Plasschaert AJ. The influence of restorative dental materials on heat transmission in human teeth. J Dent Res 1984;63:1096–100.
[142] Bagis B, Bagis Y, Ertas E, Ustaomer S. Comparison of the heat generation of light curing units. J Contemp Dent Pract 2008;9:65–72.
[143] Zuerlein MJ. Interactions of infrared lasers with dental hard tissues at clinically relevant wave lengths. University of California San Francisco/University of California Berkeley; 1999. p. 257.
[144] Zuerlein MJ, Fried D, Seka W, Featherstone JDB. Modeling thermal emission in dental enamel induced by 9–11 [mu]m laser light. Appl Surf Sci 1998;127–129:863–8.
[145] Fried D, Glena RE, Featherstone JDB, Seka W. Permanent and transient changes in the reflectance of CO_2 laser-irradiated dental hard tissues at lambda = 9.3, 9.6, 10.3, and 10.6 mu m and at fluences of 1–20 J/cm^2. Lasers Surg Med 1997;20:22–31.
[146] A comparison of the morphological changes after Nd-YAG and CO_2 laser irradiation of dentin surfaces. J Endod 2000;26:450–3.
[147] Moriyama EH, Zangaro RA, Lobo PDC, Villaverde AB, Pacheco MT, Watanabe I-S, et al. Optothermal transfer simulation in laser-irradiated human dentin. J Biomed Opt 2003;8:298–302.
[148] Sagi-Dolev AM, Shitzer A, Katzir A, Akselrod S. Heating of biological tissue by laser irradiation: theoretical model. Opt Eng 1992;31:1417–24.
[149] Sakae T. Variations in dental enamel crystallites and micro structure. J Oral Biosci 2006;48:85–93.
[150] Xu F, Wen T, Seffen KA, Lu TJ. Biothermomechanics of skin tissue. J Mech Phys Solids 2008;56:1852–84.
[151] Xu F, Wen T, Lu TJ, Seffen KA. Skin biothermomechanics for medical treatments. J Mech Behav Biomed Mater 2008;1:172–87.
[152] Kishen A, Murukeshan VM, Krishnakumar V, Asundi A. Analysis on the nature of thermally induced deformation in human dentine by electronic speckle pattern interferometry (ESPI). J Dent 2001;29:531–7.
[153] Hermanstyne TO, Markowitz K, Fan L, Gold MS. Mechanotransducers in rat pulpal afferents. J Dent Res 2008;87:834–8.
[154] Xu F, Wen T, Lu TJ, Seffen KA. Modeling of skin thermal pain: a preliminary study. J Appl Math Comput 2008;205:37–46.
[155] Xu F, Lu TJ, Seffen KA. Skin thermal pain modeling: a holistic method. J Thermal Biol 2008;33:223–37.
[156] Xu F, Wen T, Lu TJ, Seffen KA. Modelling of nociceptor transduction in skin thermal pain sensation. J Biomech Eng 2008;130:1–13.