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Heat Generation and Transfer on Biological Tissues Due to High-Intensity Laser Irradiation

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1. Introduction

The use of high intensity laser irradiation is already used clinically for many dental procedures, since the literature findings confirm its efficacy. However, for the establishment of a laser procedure for any clinical application, it is strongly necessary to study the thermal effects on irradiated tissues, considering that most of biological tissues is highly connective and can easily transmit the received heat (Nyborg & Brännström, 1968; Baldissara et al, 1997; Brown et al, 1970).

The interactions of high intensity laser irradiation on dental hard tissues are commonly induced by thermal action (Seka et al, 1996; Niemz, 1997). For the clinical application of lasers, a precise irradiation parameter must be chosen in order to avoid morphological damage, such as surface carbonization or cracking, which could produce structural, esthetic damages and post-operative complaints. Moreover, the energy densities used must be safe as regards pulp and periodontal tissue vitality (Ana et al, 2006; Nammour et al, 2004; Ana et al, 2007).

Considering a clinical application of high intensity lasers, parameters such as wavelength, energy density, intensity, peak power, average power, repetition rate and pulse length are extremely important to heat generation due to irradiation on any biological tissue. The amount of heat inside the tissue is highly dependent of its optical properties, such as absorption and scattering coefficients. Also, the heat transfer is dependent of some properties of biological tissues, such as thermal diffusivity, thermal conductivity and others (Brown et al, 1970; Seka et al., 1996; Yu et al., 1993).

The knowledge of some basic concepts of temperature and heat are necessary to understand and to evaluate the effects of lasers on biological tissues. These concepts are described below.

1.1 Temperature

The thermal state of a body normally is described by subjective words like "hot", "cold" or "warm". Therefore, the temperature is a physical parameter which is used to characterize the thermal state of the body in a less subjective and more quantitative way; in other words, assigning values.
The temperature is defined as the measurement of the thermal vibration (average kinetic energy) of the body particles. Considering that the thermal vibration of atoms and molecules is difficult to measure, it is necessary to assign numeric values to some macroscopic properties of the body. For example, the Celsius scale was assigned 100 for water boiling temperature and zero to water solidification temperature at the sea level. From this scale it is possible to graduate and assign values to the different thermal states.

1.2 Heat

Heat is the energy associated to the temperature variation. When a hot body gets together with a cold body, the hot body transfers heat (thermal energy) to the cold one. The energy flow stops when both bodies have the same temperature, reaching thermal equilibrium. Microscopically the hot body molecules vibrate more intensely than the cold body, transferring this vibration (energy) to the cold body molecules, until both bodies reach the equilibrium temperature which means they have the same mean kinetic energy. The equation that related the heat with temperature variation is:

$$\Delta Q = m.c.\Delta T \tag{1}$$

where ΔQ is the heat amount of energy (cal) acquired or lost. In this equation, $\Delta T = \left(T_f - T_i\right)$ is the temperature variation (T_f = final temperature, T_i = initial temperature); m is the mass (g) and c is the specific heat (cal/g°C).

1.3 Heat propagation

The heat is energy transferred in association to a temperature variation. The heat propagation occurs spontaneously from the hot body to cold body. There are three different mechanisms for heat propagation into materials:

a. **Conduction:** this is the typical heat propagation mechanism that occurs in solids. The vibrating particles inside the body transmit part of their kinetic energy to their neighbors. As a consequence, he heat flows from the higher temperature regions (where molecules have higher mean kinetic energy) to the low temperature ones (where molecules have lower mean kinetic energy). This process, defined as heat flow, is directly proportional to the cross-sectional area, to the temperature difference between the two points and inversely proportional to the thickness (Figure 1).
   The heat conduction equation is also known as Fourier´s law. For the one-dimensional plane wall shown in Figure 1, the heat flow can be defined as:

$$heat\ flow = \frac{\Delta Q}{\Delta t} = k \cdot \frac{A \cdot (T_i - T_f)}{\Delta z} \tag{2}$$

   where ΔQ is the heat amount that propagates through the area A and thickness Δz, Δt is the time interval, T_i e T_f are temperatures initial and final and k is a proportionality constant, called thermal conductivity coefficient. This coefficient is a characteristic of each material and expresses the facility that the heat flows through the material. The materials with a high *k* value are called good thermal conductors and with low *k* is a good thermal insulating.

b. **Convection:** this mechanism describes how the heat propagates in fluids (liquids or gases). When heating a container with a fluid, the part closer to the heat source became warmer, decreasing its density. Whence hot fluid goes up and the colder one goes down, and these movements are called convection currents.
c. **Irradiation:** this heat propagation mechanism corresponds to the emission or absorption of electromagnetic heat waves; for instance, the same way that the Sun heats the Earth.

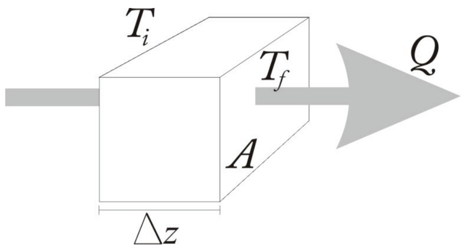


Fig. 1. Schematic diagram of heat propagation in a block having cross section area A and thickness Δz

1.4 Heat generation in a laser-tissue interaction

Considering the interaction of a laser source with the biological tissue, there are some models that should be described: the microscopic and the macroscopic model.

a. Microscopic model

When a photon with energy *hv* reach a molecule *A*, the energy could be absorbed, bringing *A* to excited state *A**:

$$hv + A \rightarrow A^* \tag{3}$$

In this way, the molecule A* suffers an inelastic collision (equal energy loss) with the neighbor B (an electron, atom or a molecule), transferring part of its energy and decay to an energy lower state:

$$A^* + B(\varepsilon_{cin}) \rightarrow A + B(\varepsilon_{cin} + \Delta\varepsilon) \tag{4}$$

This process changes part of the laser energy (high frequency electromagnetic waves) into thermal energy (low frequency electromagnetic waves).
The amount $(\varepsilon_{cin} + \Delta\varepsilon)$ represents the increase of the thermal vibration of the particle B and, as a consequence, there is a microscopic temperature increment by the energy absorbed from the photon.
The macroscopic effect of the temperature increase is observed due to the high number of type-A molecules, which absorbed the energy from a high number of photons presented in a laser beam, and transform this energy in thermal vibration. Figure 2 represents the phenomenon described above.

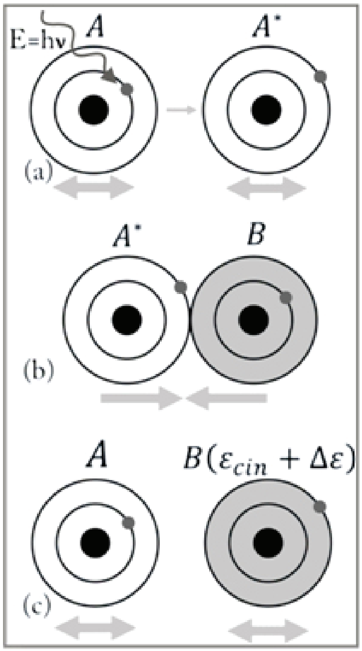


Fig. 2. Schematic diagram of the heat generation in a biological tissue: (a) a photon with the energy *hv* is applied in the molecule *A*, then goes to an excited state *A**. (b) The molecule A* collides with B. (c) A* transfers its energy to B; B becomes $B(\varepsilon_{cin} + \Delta\varepsilon)$ and starts to vibrate more intensely

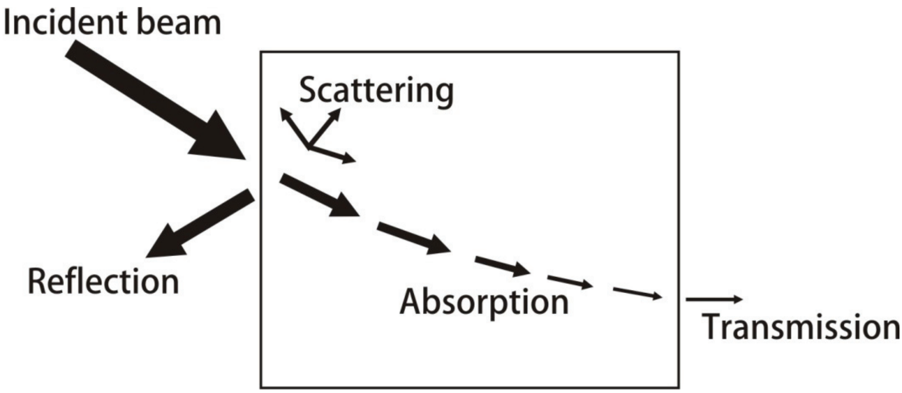


Fig. 3. Schematic diagram showing the laser-tissue interaction: reflection, scattering, absorption and transmission

b. Macroscopic model

In a macroscopic approach it could be observed that the heat generated is directly related with the laser propagation in the tissue. For that, it is convenient to remember how the heat propagation occurs. When a laser beam irradiates a sample (Figure 3), a part of the beam is reflected and the other part penetrates in the surface. That part which penetrates is attenuated mainly in two different ways: the absorption and scattering - as long as the beam penetrated in the sample.

The absorption and the scattering are characterized for absorption coefficient (μ_a) and scattering coefficient (μ_s), which represents, respectively, the rate of radiation energy loss per penetration length unit, due the absorption and the photons scattering. These two coefficients are specific to each tissue and depend on the laser wavelength.

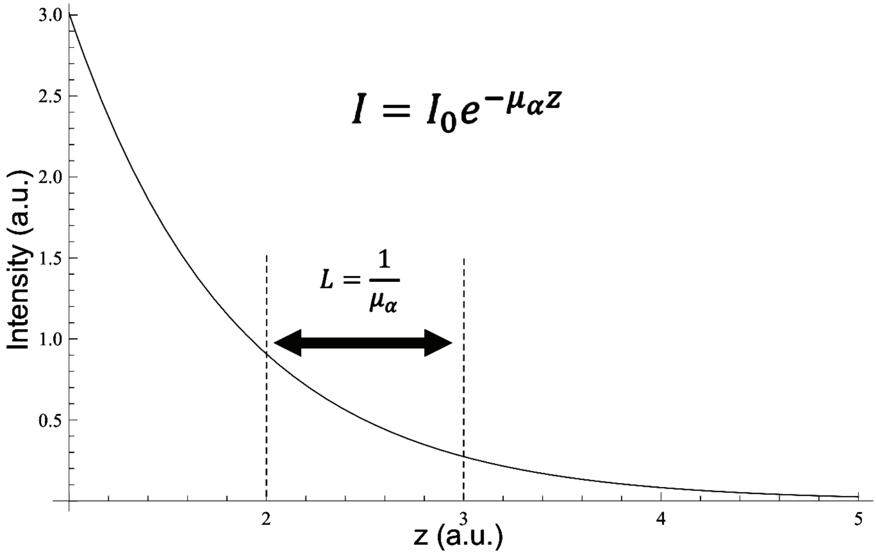


Fig. 4. Laser beam attenuation as a function of penetration length

To simplify, initially consider an absorber and not scattering sample. In this case, the beam attenuation is described by the Beer's law (Figure 4):

$$I(z) = I_0 \cdot e^{-\mu_a \cdot z} \tag{5}$$

where I is the beam intensity that depends on the penetration length z and I_0 is the intensity for z = 0. The inverse of the absorption coefficient is defined as the optical absorption length (L) (Figure 4):

$$L = \frac{1}{\mu_a} \tag{6}$$

The generated heat per area unit and per time unit, in a very small thickness Δz, is given by:

$$S(z) = \frac{I(z) - I(z + \Delta z)}{\Delta z} = -\frac{\partial I(z)}{\partial z} = \mu_a \cdot I(z) \tag{7}$$

The equation 7 expresses that the generated heat in the tissue is equal to the absorbed energy and can be described as the absorption coefficient multiplied by the local intensity.

In the most cases, the light is both absorbed and scattered into the sample simultaneously. The beam attenuation continues to be described by a similar law from the Beer's Law, but now the attenuation coefficient is the sum of the absorption and scattering coefficients, which is called total attenuation coefficient $(\mu_T = \mu_a + \mu_s)$.

1.5 Heat propagation in biological tissues

The heat conduction equation in a material medium is given by:

$$\frac{\partial T}{\partial t} = \frac{k}{\rho \cdot c}\nabla^2 T + \frac{S}{\rho \cdot c} \tag{8}$$

where T is the temperature (°C), t is the time (s), k is the thermal conductivity, ρ is the tissue density (g/cm^3), c is the specific heat (cal/g.°C) and S is the generated heat per area and per time (cal/s.cm^2).

This equation can be deduced from the diffusion general equation, but it requires a specific Physics and Mathematical knowledge. Therefore it is important to know that it describes a strong correlation among the temperature temporal variation $\left(\frac{\partial T}{\partial t}\right)$, the temperature spatial variation $(\nabla^2 T)$ and the laser source S. It is also important to say that this same equation also works when the sample is not being irradiated. In order to calculate how the heat propagates after an exposure time, when the laser beam is off, it is only necessary to solve the equation 8 with $S = 0$

There are some other thermal parameters related to the heat propagation. The thermal penetration length is a parameter that describes the propagation extension per time, and it is given by:

$$z_{thermal}(t) = \sqrt{4 \cdot \alpha \cdot t} \tag{9}$$

where $\alpha = \frac{k}{\rho \cdot c}$ is the tissue thermal diffusivity and t is the time. For instance, the thermal diffusivity of water is $\alpha = 1.4 \times 10^{-7}\, m^2/s$.

Other important parameter is the thermal relaxation time, which is obtained mathematically correlating the optical penetration length with thermal penetration length:

$$L = z_{thermal}$$

$$\frac{1}{\mu_a} = \sqrt{4 \cdot \alpha \cdot \tau_{thermal}}$$

$$\tau_{thermal} = \frac{1}{\mu_a^2 \cdot 4 \cdot \alpha} \tag{10}$$

The thermal relaxation time (equation 10) describes the necessary time to the heat propagates from the surface of irradiation until the optical penetration length and is particularly important when the intention is to cause a localized thermal damage, with minimal effect in adjacent structures. This parameter can be interpreted as follows: if the time of the laser pulse is smaller than the relaxation time, the heat would not propagate until a distance given by the optical penetration length *L*. So the thermal damage will happen only in the first layer where the heat is generated. On the other hand, if the time of the laser pulse is higher than the relaxation time, the heat would propagate for multiple of the optical penetration length, resulting in a thermal damage in a bigger volume to the adjacent structures.

2. Characteristics of dental tissues and their influence on heat propagation

The tooth is composed basically for enamel, dentin, pulp and cementum. Enamel, dentin and cementum are called "dental hard tissues", and the main constituent is represented by the hydroxyapatite (Chadwick, 1997; Gwinnett, 1992) (Figure 5). Dentin and cementum have higher water and organic compound percentage when compared to the enamel and, due to this composition, they are more susceptible to heat storage than the enamel. Dentin has low thermal conductivity values and offers more risk when lasers irradiate in deeper regions, considering that dentinal tubules area and density increase at deepest regions, and subsequently, can easily propagate the generated heat (Srimaneepong et al., 2002). As an example, considering the use of CO_2 lasers in dentistry (wavelength of 9.6 µm or 10.6 µm), the absorption coefficient for dentin tissue is lower than enamel due to its low inorganic content; also, the thermal diffusivity is approximately three times smaller, which can lead a less heat dissipation amount and, as a consequence, can induce higher pulp heating (Fried et al., 1997).

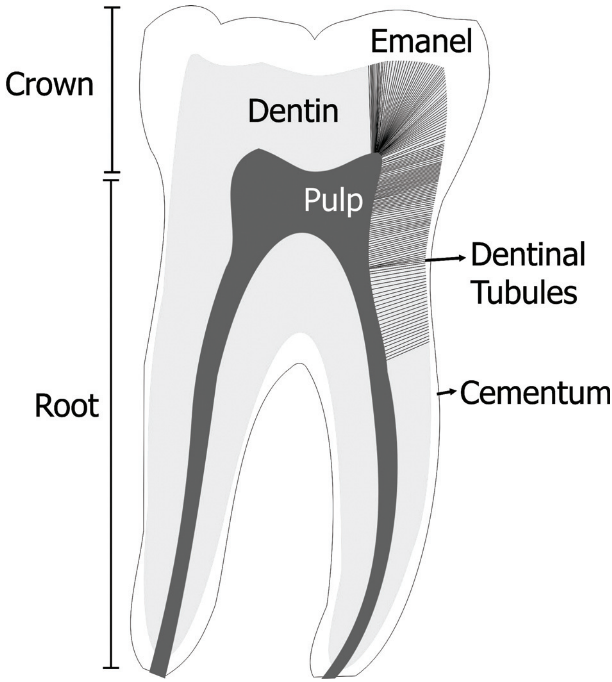


Fig. 5. Representation of a molar tooth, evidencing the macroscopic structures

Dental pulp is a connective and vital tissue, and the higher vascularization makes this tissue strong susceptible to thermal changes. The minimal change in pulp temperature ($\Delta T \leq 5$ °C) is sufficient to alter the microvascularization, the cellular activation and their capacity of hydratation and defense (Nyborg & Brännström, 1968; Zach & Cohen, 1965).
The majority of high intensity lasers used for dental hard tissues cause photothermical and photomechanical effects. Photons emitted at wavelength of visible and near infrared regions of electromagnetic spectrum are poorly absorbed by dental hard tissues (Seka et al., 1996) and, due to this fact, the heat diffusion to the pulp is easy. In this way, in order to choose a parameter of laser irradiation for a clinical application, it is necessary to establish limit energy densities that promote a significant temperature increment on enamel and dentin surface, in order to produce mechanical and/or thermal effects on these structures (Ana et al., 2007). Also, the temperature increment inside the pulp tissue must be bellowing a temperature threshold.
Previous studies have indicated that temperature increments above 5.6 °C can be considered potentially threatening to the vitality of the pulp (Zach & Cohen, 1965) and increments in excess of 16 °C can result in complete pulpal necrosis (Baldissara et al., 1997). Further studies showed levels of 60% and 100% of pulp necrosis when pulp tissue was heated about 11 °C and 17 °C, respectively (Powell et al., 1993). The pulpal temperature rise due to laser-tissue interaction has also been investigated and most of lasers systems promoted an increase in pulpal temperature dependent on the power setting (Ana et al., 2007; Yu et al., 1993; Zezell et al., 1996; Boari et al., 2009).
As well as the knowledge of laser wavelength, energy density and pulse duration, another point to be considered in heat transfer is the tissue characteristics and the influence of the oral environment. Although the calculation of heat transmission and dissipation is performed using hole sound teeth at *in vitro* studies, in clinical situations several characteristics of tissue can change, such as the type of teeth, the remaining thickness, the presence of saliva and the presence of demineralization (Ana et al., 2007; Powell et al., 1993). For instance, due to the great amount of water in carious lesions, the heat transfer to the pulp can be more excessive in decayed teeth. Relating the influence of tissue thickness, White *et al.(1994)* determined that Nd:YAG laser irradiation with a power output of 0.7 W (approximately 87 J/cm^2) induces an increase of 43.2 °C in a remaining dentin thickness of 0.2 mm and induces an increment of 5.8 °C in a dentin thickness of 2.0 mm. Considering that the human teeth present a big variation in volume and weight, and taking into account the low thermal conductivity of dentin, the operator must judge the physical conditions of dental hard tissue in order to adequate the exposition time to avoid dangerous thermal effect on pulp.

3. Changes in tissue thermal characteristics during laser irradiation

Considering the laser irradiation in dental hard tissues, it is necessary to know and to understand the thermal behavior of these tissues when submitted to heating. For that, the evaluation of the heat conduction phenomenon is extremely necessary.
Teeth are mainly composed by hydroxyapatite that, in principle, has high heat capacity value and low heat conduction value (Pereira et al., 2008). The main reason of the changes of thermal parameters of hydroxyapatite can be explained by the complexity of the photon diffusion into the material due to the ionic bond between the chemical elements.
Several studies about thermal parameters measurement in hard dental tissues have been published (Brown et al, 1970; Incropera et al., 2006). Results of these studies are summarized in table 1.

Thermal parameter	Enamel	Dentin	Water
Specific Heat (J/g°C)	0.71 (Brown et al, 1970)	1.59 (Brown et al, 1970)	4.18 (Incropera et al., 2006)
Thermal conductivity (10^{-3}W/cm °C)	9.34 (Brown et al, 1970)	5.69 (Brown et al, 1970)	6.1 (Incropera et al., 2006)
Thermal diffusivity (10^{-3}cm²/s)	4.69 (Brown et al, 1970)	1.86 (Brown et al, 1970)	1.3 (Incropera et al., 2006)

Table 1. Thermal parameters of dental hard tissues (enamel and dentin) and water

Although these thermal values are well-established in literature and can be used for supporting clinical applications, it is important to consider that all parameters were measured at room temperatures. In the moment of laser irradiation of dental hard tissues, the temperature increase can lead several chemical and ultra-structural changes on enamel and dentin (Bachmann et al., 2009; Fowler & Kuroda, 1986); as a consequence, the tissue thermal characteristics of tissue may change during laser irradiation.
Several studies have been developed in order to propose theoretical models of heat propagation in dental hard tissues (Craig R.G & Peyton, 1961; Braden et al., 1964). These models assumed that thermal parameters are constant in function of temperature, which seems to be not true according to the discussed above. Thus, we have to assume that the determination of laser irradiation parameters based only by theoretical calculation that consider thermal properties as constant can be wrong. Figure 6 shows experimental data (Pereira et al., 2008), obtained by infrared thermography, of the thermal diffusivity changes as function of temperature changes.

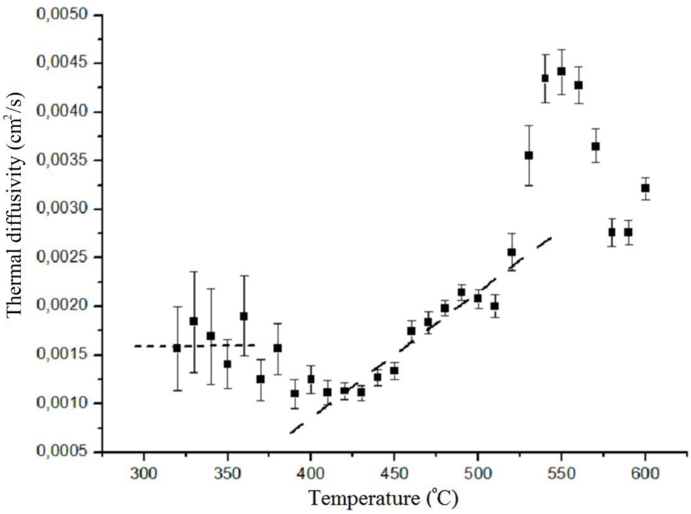


Fig. 6. Thermal diffusivity of dentin as function of temperature (Pereira et al., 2008)

Figure 7 shows the changes on heat penetration on dentin in function of time of exposure. It can be seen that data obtained vary among the related studies due to the fact that some of them consider the thermal diffusivity values always constant, while the present study (Pereira et al., 2008) consider the changes in thermal diffusivity according to the temperature (Figure 7). This fact has significant relevance mainly for clinical procedures using laser irradiation, when it is necessary temperature increases up to 800 °C for cutting dental hard tissues and for caries prevention, for example (Fried et al., 1996; Ana et al., 2007). When a tooth is submitted to this temperature elevation, the heat spreads more quickly than calculated by theoretical models that considered thermal diffusivity values as constant, which can represent a problem mainly for the deeper tissues (pulp tissue).

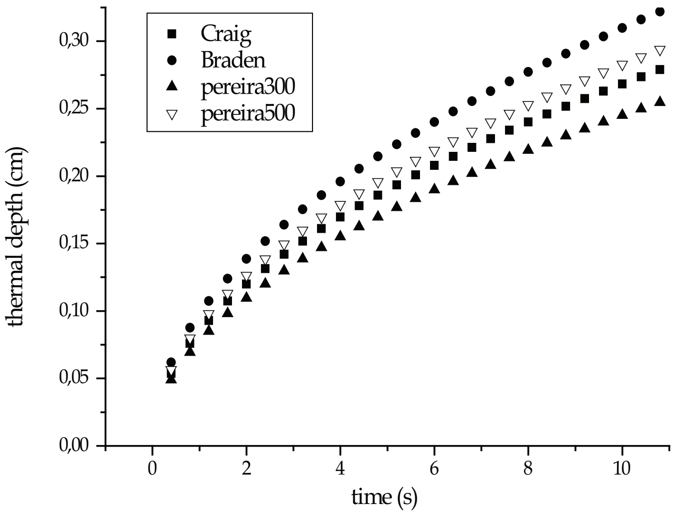


Fig. 7. Calculated thermal depth (cm) in function of time (t) for dentin tissue, obtained by four different literature studies (Pereira et al., 2008)

4. Considering temperature to determine clinical protocols using lasers

As it was stated previously, for the determination of clinical protocols it is demanding to consider the safety and efficacy of lasers, also the characteristics and properties of target tissues. Besides that, literature studies clearly show that laser features, such as wavelength, mode of operation (continuous *versus* pulsed modes), temporal pulse length and repetition rate are characteristics directly related with pulp heating. In this way, among the optical properties, the transmission is the most important property to be considered for preserving pulp vitality.

Among high intensity lasers with high absorption and low transmission through enamel and dentin, erbium lasers seems to be the most appropriated wavelength to be used in dentistry. However, some studies point out that, even with this laser, the repetition rate and pulse duration are decisive on determining clinical parameters; for example, the longer pulse duration is, the higher is the heat generated in pulp (Yu et al., 1993).

Taking into account the clinical application of high intensity lasers on dental hard tissues, some strategies may be useful to control the heat generation and transmission on these tissues. In order to restrict the heat dissipation through the teeth tissues, the application of a photosensitizer is frequently applied over the enamel and dentin surfaces before laser irradiation, and this application can avoid pulpal damages even when laser irradiation occurs with high energy densities (Tagomori & Morioka, 1989; Jennett et al., 1994). The application of a photosensitizer before laser irradiation is commonly used in order to enhance surface tissue absorption in the near-infrared range for ablation and caries prevention actions in dental tissues, considering that some lasers, such as Nd:YAG and Ho:YAG, are poorly absorbed by enamel and dentin. The absorption of the laser beam is increased at the surface of the enamel and the heat produced due to laser absorption in the coating material is transmitted into the adjacent enamel. This technique certifies the deposit of a short laser pulse energy to a small volume of tissue, avoiding the excessive laser beam penetration in deeper dental structures and consequently with less risk of damages in dental pulp (Boari et al., 2009).

The use of Indian Ink is a well-recognized and efficient technique to reduce beam transmission on dental hard tissues. However, because of the difficulty in its removal, which can prejudice the aesthetics of remaining teeth, it has been suggested the application of a coal paste, a mixture of triturated vegetal coal in 50% ethanol, which is biocompatible, easy to remove and presented important results in previous *in vitro* (Boari et al., 2009) and *in vivo* (Zezell et al., 2009) studies. In an *in vitro* study performed by our group, it was demonstrated that the enamel recovering with the coal paste promoted an increase of surface temperatures, which confirmed the absorption of laser beam at the surface (Ana et al., 2007) (Figure 8). Also, the coal paste significantly decreased the heat transfer into the teeth when enamel was irradiated with Nd:YAG and Er,Cr:YSGG lasers, and can assure the pulpal safety when laser irradiation is performed for a long period of time. The

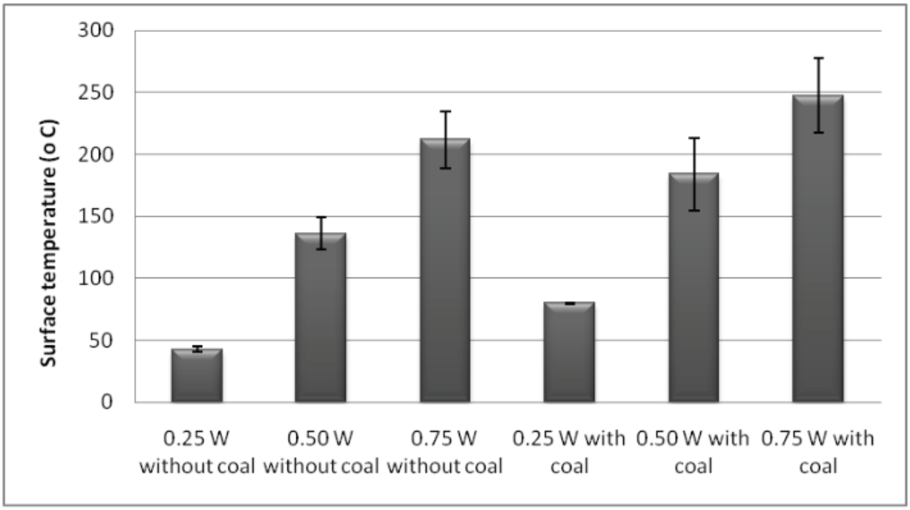
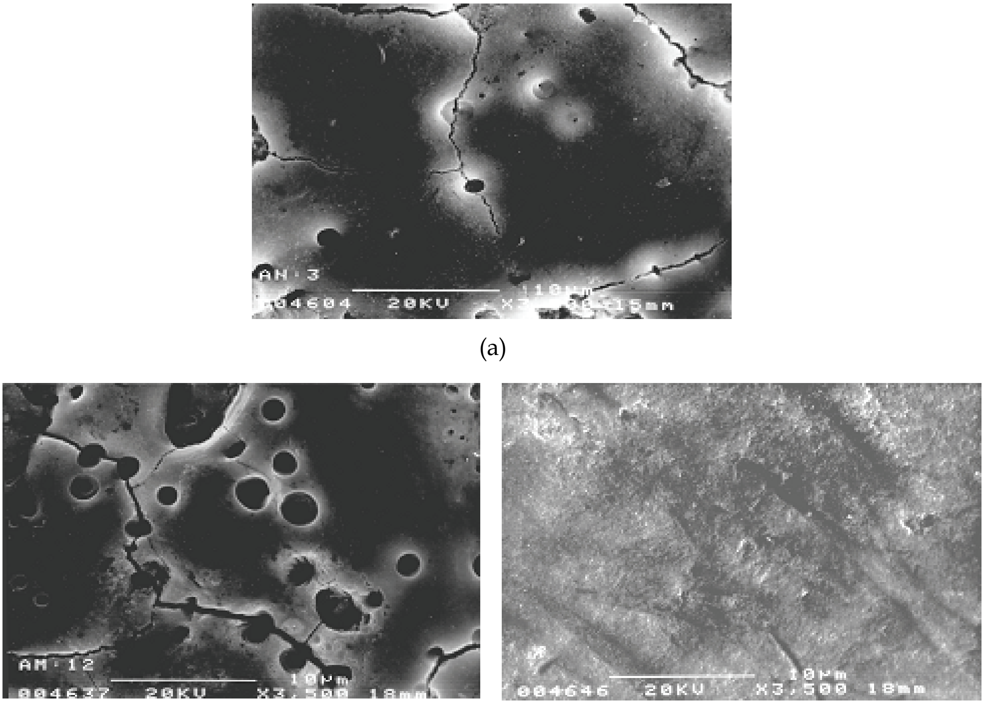


Fig. 8. Surface temperature increase on enamel surface during Er,Cr:YSGG (λ = 2078 nm) laser irradiation with and without the application of coal paste (Ana et al., 2007). It can be noted that, even at three different average powers, the presence of the photosensitizer significantly increased the surface temperature during laser irradiation. Bars mean standard deviation

morphological changes promoted on enamel surface are similar than those promoted by the recovering with Indian Ink, showing evidences of surface heating that promoted melting and recrystallization of enamel (Boari et al., 2009) (Figure 9).



(a)

(b) (c)

Fig. 9. Scanning electron micrography of dental enamel after irradiation with Nd:YAG (λ = 1064 nm) laser irradiation at energy density of 84.9 J/cm^2 after surface recovering with Indian Ink (a) or coal paste (b) or no recovering (c) (Boari et al., 2009). It is possible to note the presence of melting and recrystallization of enamel after recovering with coal paste and Indian Ink. These characteristics are not observed when enamel is irradiated without the presence of a photosensitizer. Original magnification = 3500 X. (a) enamel + Indian Ink; (b) enamel + coal paste; (c) sound enamel

The presence of air-water spray during laser irradiation is another strategy used for clinicians to avoid excessive heat generation on the pulp. The water coolant allows the cleaning of surfaces to be irradiated and increases the efficacy of ablation phenomenon, in a process called “water augmentation” (Fried et al., 2002). When dental hard tissues are irradiated with Er:YAG in addiction to a thin water layer, studies relate that the cutting efficiency increases at the same time that the pulp temperature decreases. However, the thickness of water layer should be well-controlled, considering that erbium lasers interacts primary with water and an thick water layer over the tissue can restrict the laser interaction with the enamel bellow it and, as a consequence, the absorption by the target tissue can decrease.

5. How to determine temperature variations in biological tissues?

Among physical methods to determine the temperatures on materials, the thermocouples (Ana et al., 2007; Boari et al., 2009), elliptical mirrors, HgCdZnTe detectors (Fried et al., 1996) and infrared cameras (Ana et al., 2007) are the most used ones to measure temperature changes in pulp, periodontal tissues and dental hard tissues surfaces. These techniques present good accuracy and efficacy, and can be easily adapted to experimental conditions. However, it should be considered that all experimental methods present some difficulties, such as sample standardization (considering the large variation in volume, size, thickness, and hydratation degree of tissues), the exact duplication of the thermal load, accuracy, availability and cost of equipments (Ana et al., 2008).

The finite element method model (FEM model) is another method that has been popular among researchers, taking into account that this technique is a good analytical tool to model and simulate the thermal or mechanical behavior of dental structures (Toparli et al., 2003). The FEM model can be used to simulate the effects of laser on enamel and dentin, but not on gums or inside the pulp cavity, which is filled with blood vessels and innervated tissues, since these materials are soft and highly inhomogeneous. However, the effects of laser irradiation could be difficult to simulate even on the dental hard tissues (only enamel and dentine), since the thermal characteristics of these materials may not have been well determined.

It must be pointed out that all the *in vitro* methods do not reproduce exactly all the interferences of the oral environment in the photothermal response of enamel and dentin tissues, such as the influence of surrounding saliva, pulpal and periodontal tissues, presence of biofilm, body temperature and other factors (Ana et al., 2008).

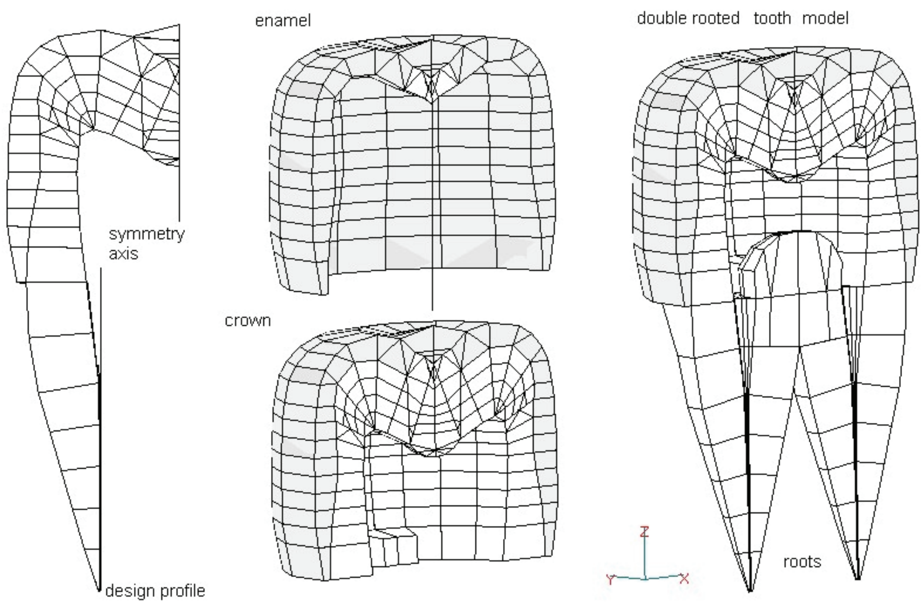


Fig. 10. FEM model used to simulate the heat generation and transmission at dental pulp during Er,Cr:YSGG laser irradiation (Ana et al., 2008)

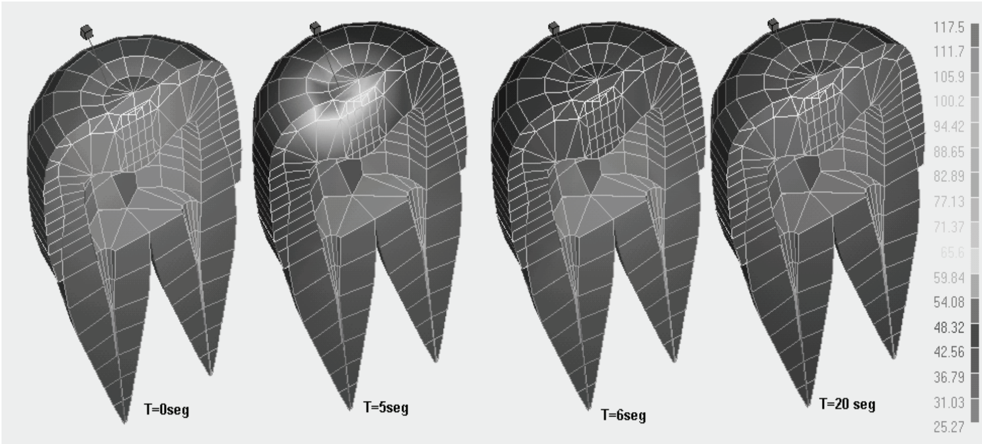


Fig. 11. Calculated temperature distribution: The gray scale represents the temperature in Celsius. The sequence of images illustrates the effects of irradiation with a laser beam and the propagation of heat through the tooth after the beam is turned off

6. Pre-clinical studies

As affirmed previously, the pre-clinical studies are necessary to predict the biological effects of high-intensity lasers and to establish possible parameters and conditions for a future application in dental practice. For preventing dental caries, for example, infrared lasers such as Nd:YAG and Er,Cr:YSGG can be indicated.

The Er,Cr:YSGG laser is emitted in 2.78 μm wavelength, which is better absorbed by water and OH^- contents of hydroxyapatite (Seka et al., 1996), and promotes surface temperatures up to 800 °C at the ablation threshold (Fried et al., 1996). Due to this fact, Er,Cr:YSGG laser is applied for cutting of enamel, dentin and root surfaces, and also for caries prevention.

In a first *in vitro* study, surface temperature measurements were performed in order to verify if this laser had potential to promote chemical and crystalline changes on dental enamel, which can occur in temperatures above 100 °C (Bachmann et al., 2009). For that, the temperature changes in enamel surface during and immediately after laser irradiation were monitored using an infrared high resolution fast thermographic camera (ThermaCam FLIR SC 3000 Systems, USA), which stores infrared images and data at rates up to 900 Hz. This experiment was performed at a controlled room temperature of 24.6 °C, 47 % air relative humidity and considering teeth emissivity as 0.91. The thermographic camera was positioned at 0.1 m distance of samples and the obtained infrared images were recorded at rates of 900 Hz for later analysis (Ana et al, 2007).

For laser irradiation, laser handpiece was positioned at focused beam, at 1 mm distance from the enamel surface. This assembly was kept in optical supports and the area of interest was isolated at a focal length of 0.1 m using an internal macro lens.

The results of surface temperature obtained in this study (Ana et al., 2007) are shown in Figure 12. It is possible to evidence that the surface temperature rises with the increase of energy density, and the presence of photosensitizer (coal paste) propitiated higher temperature values when compared to the surfaces in which were not previously recovered with the coal paste. In this way, even using laser wavelengths highly absorbed by dental enamel, the application of a photosensitizer in the enamel surface can potentiate the absorption phenomenon. This can reflect on temperature rise and, in this way, crystalline changes at this superficial enamel may occur and can favor the caries preventive effect.

Another point to be considered is that the temperature rise of 247.6 °C found when teeth were irradiated with 8.5 J/cm² is lower than that temperature reported by literature studies, who found approximately 400 °C measured by an elliptical mirror and a HgCdZnTe detector with a time resolution of 1 μs (Fried et al., 2006). In this way, the temperature elevation during Er,Cr:YSGG laser irradiation could be higher than those detected by infrared camera. Taking into account that the pulse width of Er,Cr:YSGG laser is 140 μs, even the 900 Hz recording rate of the infrared thermographic method seems to be unable to detect the highest temperature peaks during laser irradiation. In this way, the infrared thermographic camera gives an idea of average temperature changes when teeth are irradiated with high intensity lasers. However more accurate systems are required to precisely determine the maximum temperature peaks, such as the use of integrating sphere.

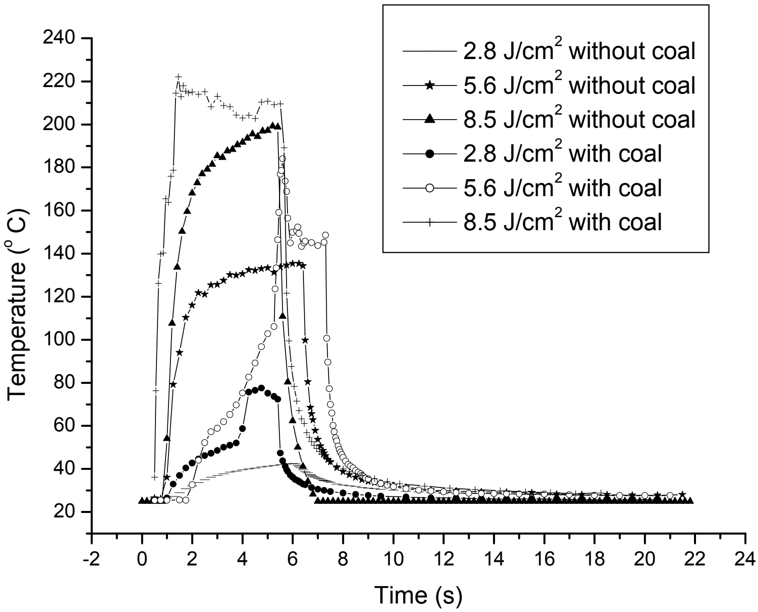


Fig. 12. Dental enamel surface temperature measurement, by infrared thermography, during Er,Cr:YSGG laser irradiation at parameters aimed at caries prevention (Ana et al., 2007)

For measurement of heat transfer to the pulp chamber, the use of fast-response thermocouples seems to be more accurate since it is not possible to see the heat transfer from enamel to pulp by infrared thermography, unless the teeth are half-sectioned.

In this way, calibrated K-type chromel-alumel thermocouples (Omega Engineering, Stanford, USA) were inserted inside the pulp chamber of sound molar human teeth, which were previously filled with a thermally-conductive paste (thermal conductivity of 0.4 cal s^{-1} m^{-1} K^{-1} – Implastec, Votorantim, Brazil) in order to keep thermal contact between the probe end and dentin surface. These thermocouples had 0.05 mm diameter probe and were sensitive to temperature variations between 0.1 °C and 100 °C. The temperature sensitive end of the probe was placed at the closest distance to the area to be irradiated, and its location was controlled radiographically for each sample (Romano et al., 2011). The thermocouple apparatus was connected to an analogue-to-digital converter (SR lock-in amplifier, Stanford Research System, USA) linked to a computer, and time and temperature data were recorded at sampling rate of 20 Hz, with temperature resolution of 0.1° C. During laser irradiation, samples were fixed and immersed in a water-filled heating circulator at standardized temperature of 37 °C, with only the coronal part of the tooth not being submerged in order to simulate body temperature in the oral environment.

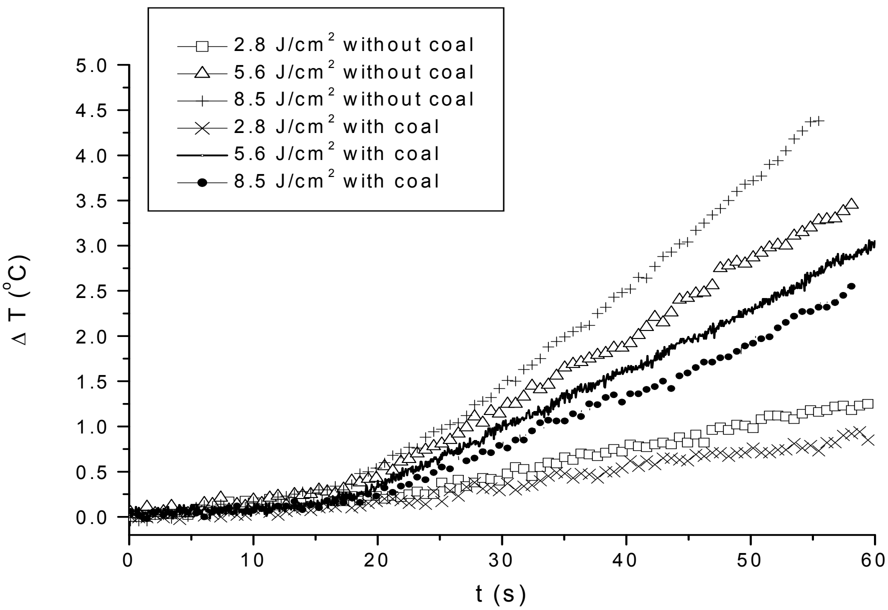


Fig. 13. Pulp chamber temperature variation during enamel surface irradiation with Er,Cr:YSGG laser, measured by fast-response thermocouples (Ana et al., 2007)

Figure 13 shows the results of temperature evaluation inside the pulp chamber during enamel irradiation by Er,Cr:YSGG laser. It is possible to observe that the device can detect minimal temperature variations and, although surface temperatures detected increased up to 230 °C, the pulp temperature variations were up to 4.5 °C. This fact evidences that dental enamel and dentin are good thermal insulating tissues. Also, the presence of the photosensitizer on enamel surface was important to effectively reduce the heat transfer through the pulp chamber, increasing the safety of a future clinical procedure. It must to be emphasized that the time of exposure is important and further histological *in vivo* studies are also necessary to confirm this hypothesis.

All studies described previously were performed to evaluate just one possible clinical application of Er,Cr:YSGG laser. Considering that this laser can be used for multiple applications, the development of a FEM model could be a faster tool to evaluate the heat generation and transfer to dental hard tissues. In this way, a further study was performed to develop this model and to compare the simulation results with those obtained experimentally (Ana et al., 2008).

For that, a geometric FEM model of a half-sectioned double rooted molar tooth was constructed using a typical profile of a tooth root. Since the goal was to calculate the temperature distribution on the surface and inside the tooth, in each element of the model the total heat is given by the internal heat, determined by the material density (ρ) and specific heat (**c**), and the heat flux, determined by the element material thermal conductivity (α). In this study, the external heat source is due to almost instantaneous light absorption converted into heat. Moreover, in spite of not considering wavelength dependence with absorption, reflection, transmission and scattering, the FEM model predicts accurate values for temperatures inside the teeth, mainly because 2.78 µm is strongly absorbed by the dental hard tissue. The optical penetration is very small (to the order of few micrometers) since the optical absorption coefficient of enamel is about 7000 cm^{-1}.

The *in vitro* experiment aimed to compare the results of FEM simulation was performed using half-sectioned teeth, irradiated with Er,Cr:YSGG laser on enamel surface and monitored by infrared imaging. This set-up allowed the visual evaluation of temperature changes and heat diffusion from enamel into dentin and the pulp chamber. Taking into account that the variation in temperature is also dependent on the volume and weight of a tooth, the use of sectioned teeth does not correspond to an *in vivo* condition inside the mouth, but gives a reasonable idea of heat transfer inside the tooth and gives the exact value of surface temperature during laser irradiation. Moreover, in a clinical protocol, laser irradiation is performed by scanning all over the enamel surface, and it was not possible to reproduce this situation in the present study. However, in a clinical application of lasers, the time of irradiation in just one region of tissue is always less than the time considered during experiments, which increases the safety of evaluated parameters.

Energy Density	**Presence of photosensitizer**	**Infrared thermographic camera**		**Finite element method model**	
		Surface	Pulp	Surface	Pulp
	yes	79.6 °C	0.5 °C	94.2 °C	0.95 °C
2.8 J/cm²	no	42.7 °C	1.0 °C	66.0 °C	0.9 °C
	yes	184.1 °C	1.5 °C	178.0 °C	1.9 °C
5.6 J/cm²	no	136.1 °C	1.2 °C	136.0 °C	1.4 °C
	yes	247.6 °C	2.1 °C	231.0 °C	2.1 °C
8.5 J/cm²	no	211.8 °C	1.9 °C	188.0 °C	2.3 °C

Table 2. Comparison of means of temperature rise on surface and in pulp obtained by FEM model and by thermographic camera (Ana et al., 2008)

Table 2 shows the results of temperature monitoring by infrared thermography and the simulation by FEM model. It is possible to observe a good correlation between the *in vitro* and computational method, indicating that FEM model can be used as an alternative to determine heat generation on the enamel surface as well as inside the pulp chamber according to the several laser energies and the presence or not of a photosensitizer. Although it is not possible to simulate the influence of oral environment on FEM model, under the given conditions the simulated model was shown to have a good approximation to the physical reality.

7. Conclusion

The effect of high intensity lasers irradiation on biological tissues and consequently their clinical uses are based on heat generation, which is necessary to assure effective clinical procedures such as faster cutting, good homeostasis and desired chemical changes on target tissues. The comprehension of heating generation and transmission through these tissues is essential to determine safe irradiation parameters of lasers and, for that, the knowledge of optical and thermal properties of tissues and their changes due to heat are strongly necessary. Also, the interaction of laser wavelength with the tissues is necessary to avoid deleterious effects in target tissues, as well in surrounding ones.

The pre-clinical experiments give us important information about laser-tissue interaction, and help to suggest laser parameters and conditions for development of a further clinical protocol. For that, methods such as thermocouples, infrared thermography and finite element simulation are good tools that demonstrated to be useful on predicting the clinical results.

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