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Influence of FBG sensors length on temperature measures in laser-irradiated pancreas: theoretical and experimental evaluation

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Abstract— **Temperature distribution *T(x,y,z,t)* in tissue undergoing Laser-induced Interstitial Thermotherapy (LITT) plays a crucial role on treatment outcome. Theoretical and experimental assessment of temperature on *ex vivo* laser-irradiated pancreas is presented. The aim of this work is to assess the influence of thermometers dimensions on temperature measures during LITT. *T(x,y,z,t)* inside tissue is monitored by optical sensors, i.e., Fiber Bragg Gratings (FBGs): three FBGs with lengths of 10 mm and nine FBGs of 1 mm, at different distances (2 mm, 5 mm and 10 mm) and different quotes (0 mm, 2 mm and 4 mm) from the laser fiber tip are used. Theoretical punctual *T(x,y,z,t)* is averaged out on both 10 mm and 1 mm in order to compare numerical predictions with experimental data. Results demonstrate the influence of FBG length on *T(x,y,z,t)* measures. This phenomenon depends on the distance between sensor and applicator: it is particularly significant close to the applicator tip (2 mm) because of the high spatial *T(x,y,z,t)* gradient within the tissue. Both theoretical results and experimental ones show that just at a distance of 10 mm from the tip, differences between *T(x,y,z,t)* provided by FBGs of 10 mm and 1 mm are negligible.**

I. INTRODUCTION

Local hyperthermia is a technique for cancer treatment: it employs increase of temperature within a body district for tumor ablation [1]. The increment of temperature inside the target tissue is achieved by a heat source guided within the target. When the heat source is a laser and the outcomes depend on laser-tissue interaction, this technique is known as Laser-induced Interstitial Thermotherapy (LITT). The aim of this treatment is the complete removal of tumor mass, and the preservation of healthy tissue surrounding the cancer. Hence, the monitoring of tissue temperature distribution -*T(x,y,z,t)*- during LITT in both tumor target and surrounding tissue is pivotal for the clinical evaluation of hyperthermia efficacy [2], and to optimize laser settings.

LITT is routinely used for treatment of tumors in several organs such as liver, brain, prostate [3]. Recently, its feasibility on pancreas has been investigated and first trials on pig model showed encouraging outcomes [4]. The final aim of the clinician employing LITT is the estimation of volumes of tissue ablated (coagulated and vaporized) due to laser light absorption [5]: since thermal damage strongly depends on temperature (up than 60 °C [6]) and time, the monitoring of its spatial and temporal evolution is fundamental in order to establish the effectiveness of LITT.

Besides numerical model implemented for *T(x,y,z,t)* prediction, several methods can be used for monitoring temperature in tissue undergoing LITT: infrared thermography, thermocouples [7], and fiber optic sensors, which are widespread in medicine [8, 9, 10, 11]. Among fiber optic sensors for thermometry, fluoroptic probes and Fiber Bragg Gratings (FBGs) sensors [12] are the most employed. Infrared thermography measures only the surface temperature. Invasive measurements can be obtained with sensors inserted inside the tissue; anyway, thermocouples and fluoroptic sensors may show artefacts due to self-heating and direct absorption of laser light [13]. Respect to the abovementioned sensors, FBGs have the advantages of immunity from electromagnetic interferences and small dimension (e.g., 1 mm of length), and can be embedded in an array of multiple sensing elements [14].

This study presents the theoretical prediction of *T(x,y,z,t)* within pancreas undergoing LITT and an original comparison with experimental data. In fact, *T(x,y,z,t)* measurement inside two *ex vivo* swine pancreases are performed with two modalities: 1) three FBGs with 10 mm of length placed at different distances from the laser fiber applicator tip (2 mm, 5 mm and 10 mm), 2) nine FBGs 1 mm-long (three embedded in three different fibers) placed at the same distances from tip, but at different quotes (0 mm, and 2 mm and 4 mm downward the applicator). We evaluate the sensitivity of the mathematical model to predict response of sensors with different length, by averaging out on 10 mm and 1 mm the theoretical punctual *T(x,y,z,t)*. The main aim of this work is to assess the difference in *T(x,y,z,t)* measure and prediction due to different sensor length, and the advantage to employ small sensors (ideally punctual).

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Pancreases are irradiated with Nd:YAG laser beam (1064 nm) with power (P) of 6 W and energy (E) of 1000 J, guided within a quartz optical applicator (150 μm core radius, r_f).

II. Theoretical Background

A. Pancreas thermal response model

The thermal response of pancreas is related to P deposition in tissue and can be theoretically predicted by Bioheat Equation [3, 5, 12], as shown in (1). The metabolic heat generation and the heat contribution due to blood perfusion are absent, because *ex vivo* pancreases are treated.

$$\rho \cdot c \frac{\partial T(x,y,z,t)}{\partial t} = \nabla(k\nabla T(x,y,z,t)) + Q_{laser} - Q_e \tag{1}$$

where ρ is the tissue density [kg·m^{-3}], c is the tissue specific heat [J·kg^{-1}·K^{-1}] and k is the tissue heat conductivity [W·m^{-1}·K^{-1}]. $T(x,y,z,t)$ is the tissue temperature, expressed as function of spatial coordinates x, y, z and of time, t [s].

As described in previous publications [5, 12], laser beam irradiance, $I(x,y)$ [W·m^{-2}], is modeled using a 2D Gaussian distribution with standard deviation σ equal to $r_f/3$, in order to obtain the 99% of P contained within the fiber core:

$$I(x,y) = I_0 \cdot e^{-\frac{x^2+y^2}{2\sigma^2}} \tag{2}$$

where $I_0 = \dfrac{P}{2\cdot\pi\cdot\sigma^2}$ is the collimated irradiance [W·m^{-2}]. Lambert-Beer's Law is used to express the relation between laser heat source term Q_{laser} [W·m^{-3}] as a function of tissue optical properties:

$$Q_{laser} = \mu_{eff} \cdot I(x,y) \cdot e^{-\mu_{eff}\cdot z} \tag{3}$$

Scattering and absorption phenomena in the tissue are modeled by the effective attenuation coefficient μ_{eff} [m^{-1}] in (4), that determines the amount of laser light penetrating into the tissue and, therefore, the amount of laser energy converted into heat.

$$\mu_{eff} = \sqrt{3\mu_a(\mu_a + \mu_s(1-g))} \tag{4}$$

where μ_a [m^{-1}] is the absorption coefficient, μ_s [m^{-1}] the scattering coefficient and g the anisotropy coefficient.

Q_e [W·m^{-3}] is the power absorption due to water evaporation:

$$Q_e = -\lambda \cdot \frac{d\rho_w}{dt} \tag{5}$$

where λ is the water's latent heat [J·kg^{-1}] and ρ_w is the water density [kg·m^{-3}], that depends on temperature [15]:

$$\rho_w(T) = \begin{cases} 0.778\cdot\left(1-e^{\frac{T-106}{3.42}}\right) & T \le 103°C \\ 0.0289\cdot T^3 - 8.924\cdot T^2 + 919.6\cdot T - 31573 & 103°C < T < 104°C \\ 0.778\cdot e^{\left(-\frac{T-80}{34.37}\right)} & T \ge 104°C \end{cases} \tag{6}$$

At 100 °C water boils and induces lysis, causing necrosis and the loss of physiological activity of cells. Equations (5-6) are applied under the following hypotheses: the whole water steam does not leave the system, steam fills the tissue region at lower temperature and condenses uniformly.

Constant values are: for pancreas, ρ=1040 kg·m^{-3}, c=3590 J·kg^{-1}·K^{-1}, k=0.5417 W·m^{-1}·K^{-1}; λ=2260000 J·kg^{-1}. For the quartz fibers: ρ=2600 kg·m^{-3}, c=820 J·kg^{-1}·K^{-1}, k=3 W·m^{-1}·K^{-1}. For some constants, e.g., c and optical properties, pancreas values have been replaced with liver ones [2], due to the absence of data in literature. FBGs properties were not included in calculus, aiming to simplify the simulation.

Theoretical model was implemented in COMSOL Multiphysics 3.5a, considering a 3D geometry in order to model the pancreas and the bare fiber applicator. Initial tissue temperature and boundary conditions are $T_0=T_\infty$=298 K (room temperature).

B. Fiber Bragg Grating: working principle

Temperature distribution was monitored by FBG sensors. FBGs are optical elements embedded in a short segment of optical fiber [14]. They consist of a periodic variation in the reflective index of the fiber core. Grating is characterized by its spatial period, Λ [m], and the effective reflective index, n_{eff}. When interrogated with a polychromatic radiation, a narrow range of wavelengths are reflected, the other ones are transmitted. The reflected range of wavelengths are centered around a characteristic value, i.e., Bragg wavelength (λ_B), expressed as follows:

$$\lambda_B = 2\cdot \Lambda \cdot neff \tag{7}$$

The variation of λ_B ($\Delta\lambda_B$) depends on the variation of temperature, ΔT, or mechanical stress, ε, as reported in (8):

$$\frac{\Delta\lambda_B}{\lambda_B} = P_e \cdot \varepsilon + [P_e \cdot (\alpha_s - \alpha_f) + \zeta] \cdot \Delta T \tag{8}$$

where P_e is the strain-optic coefficient, α_s [K^{-1}] and α_f [K^{-1}] are the thermal expansion coefficient of the fiber bonding material and of fiber respectively, and ζ is the thermal-optic coefficient [K^{-1}]. In order to employ FBG as temperature sensors in a pancreas undergoing LITT, in a previous work we conducted preliminary trials allowing to consider negligible ε of the tissue [12]. Therefore (8) can be reduced and simplified as follows:

$$\Delta\lambda_B = \alpha_T \cdot \Delta T \tag{9}$$

where α_T [nm·°C^{-1}] is a global thermal coefficient.

III. Experimental Trials

A. Static Calibration of Fiber Bragg Grating

A static calibration is performed to verify the linear relationship between $\Delta\lambda_B$ and ΔT in (9). Temperature is adjusted and changed in the range between 20 °C and 80 °C inside a thermostatic chamber (B.E. 77, BICASA), and two silicon bandgap sensors are used as temperature reference (LASCAR ELECTRONICS EL USB-2-LCD, accuracy of ±0.5 °C). The calibration curve is obtained by a linear fit of the data using the least mean square error algorithm. α_T is estimated by fitting experimental data with a level of confidence of 95%, resulting in α_T = 0.008417±0.000004 nm·°C^{-1}.

B. Experimental Set Up

The increase of temperature (ΔT) in *ex vivo* porcine pancreases during LITT is monitored. Experimental trials on animal pancreatic tissues are carried out with P=6 W and E=1000 J.

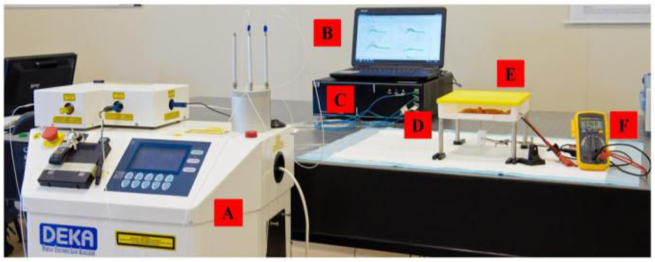

Figure 1. Experimental setup: laser (A), PC (B), OSA (C), mechanical support containing pancreas (D), FBGs (E), thermistor connected to a multimeter (F).

The Nd:YAG laser beam (Fig. 1A) with wavelength of 1064 nm (DEKA smart 10634) is guided within an optical quartz applicator (150 µm core radius). A mechanical support is *ad hoc* realized to contain tissue, FBGs and applicator, and to control relative distances among them (Fig. 1D). An optical spectrum analyzer, OSA (Fig. 1C), with four channels (Optical Sensing Interrogator Si 425, MICRON OPTICS) is used to record $\Delta\lambda_B$ with frequency of 250 Hz. It permits to simultaneously monitor each grating output. Moreover, we use a PC (Fig. 1B) to save data registered by the OSA. A thermistor (NTC) placed far from the applicator (about 10 cm) and connected to a multimeter (FLUKE 179) is used to monitor initial pancreas temperature, equal to room temperature (Fig. 1F).

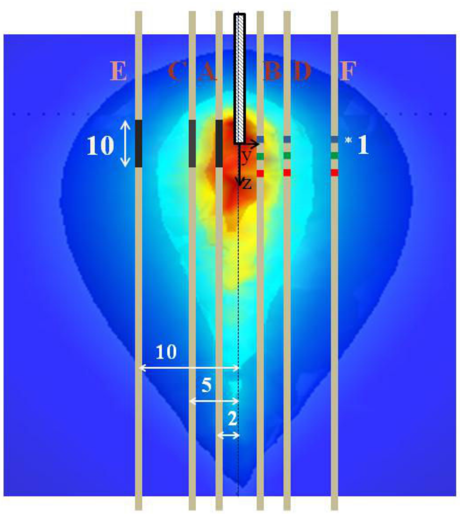

Figure 2. Schematic of FBGs arrangement in pancreas: FBGs 10 mm-long (left side) and FBGs 1 mm-long (right side). Lengths are expressed in mm.

The polymeric mask houses sensors at known distances from the applicator tip (2 mm, 5 mm and 10 mm, for both experiments, Fig. 2). Applicator and sensors are parallel each other. Measurements are performed by FBGs (Technica SA, Fig. 1E), arranged according to two configurations: 1) three optical fibers each equipped with one FBG 10 mm-long (Fig. 2, left side) and 2) three optical fibers each equipped with three FBGs 1 mm-long at different quotes (0 mm, and 2 mm and 4 mm upward the applicator, Fig. 2 right side).

IV. Results and Discussion

A. Theoretical results

Theoretical $T(x,y,z,t)$ provided by numerical simulations is obtained by averaging out on 10 mm and 1 mm the theoretical temperatures in correspondence of gratings position in experiments. Results are depicted in Fig. 3. As expected, ΔT predicted by averaging the theoretical punctual $T(x,y,z,t)$ on 10 mm is lower than the one obtained by averaging on 1 mm. It happens because the temperature gradient inside a tissue undergoing LITT is high. In fact, the tissue can experience a steep variation of ΔT along 10 mm as demonstrated by temperature modeling on 1 mm. At 2 mm from the applicator, theoretical maximum ΔT (ΔTmax) registered at the end of irradiation are the following: predicted value on length of 10 mm (Fig. 3A) reaches about 145 °C, ΔTmax predicted on length of 1 mm and three quotes, ranges from 118 °C to 200 °C (Fig. 3B). The difference between the two configurations drops with the increase of the distance between the tip and the sensors, because temperature gradient decreases. As a matter of fact, at 5 mm the estimated ΔTmax averaged out on 10 mm (Fig. 3C) reaches 50 °C, whereas other configuration shows ΔTmax ranging from 46 °C to 63 °C (Fig. 3D). At 10 mm distance from the tip, the difference is equal to a few degrees (8 °C *vs* a range of 9-11 °C, Fig. 3E-F).

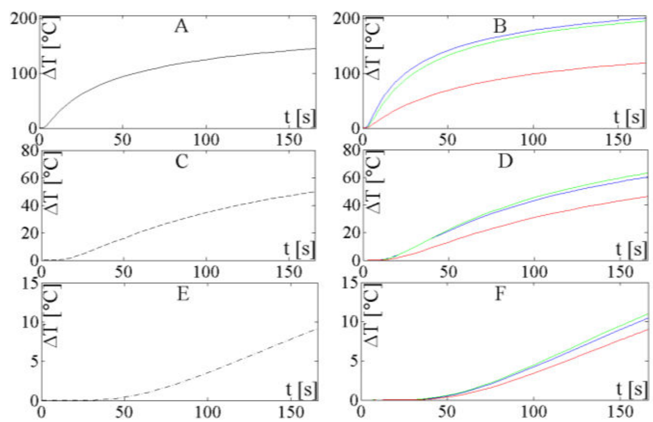

Figure 3. Theoretical ΔT in ex vivo pancreas during LITT. ΔT averaged out on 10 mm of length at 2 mm (A), 5 mm (C) and 10 mm (E) from applicator tip; ΔT averaged out on 1 mm of length at 2 mm (B), 5 mm (D) and 10 mm (F) from applicator tip, at three quotes: 0 mm (blue line), 2 mm (green line) and 4 mm (red line).

B. Experimental results

Theoretical trends are confirmed by experimental data (Fig. 4). FBG 10 mm-long at 2 mm from the applicator measures a ΔTmax of 62 °C (Fig. 4A), whereas ΔTmax monitored by three FBGs 1 mm-long ranges from 40 °C to about 74 °C (Fig. 4B). At 5 mm, FBG 10 mm-long records 48 °C (Fig. 4C) *vs* the range 35-48 °C showed by smaller sensors (Fig. 4D). Lastly, at 10 mm from the applicator tip, in configuration 1 (Fig. 2, left side) sensor measures less than 8 °C (Fig. 4E), whereas in configuration 2 (Fig. 2, right side) FBGs record ΔTmax ranging from about 4 °C up to 8 °C (Fig. 4F).

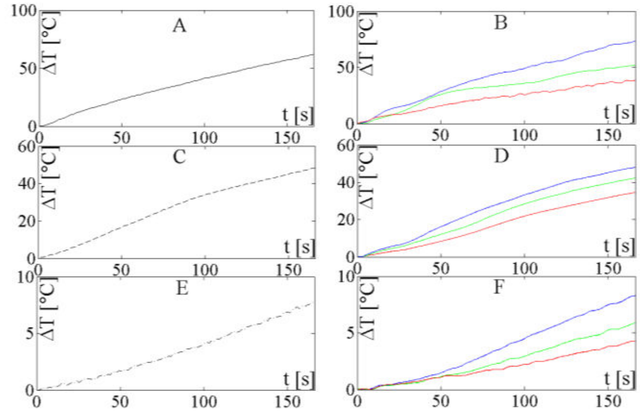

Figure 4. Experimental ΔT in ex vivo pancreas during LITT. ΔT averaged out on 10 mm of length at 2 mm (A), 5 mm (C) and 10 mm (E) from applicator tip; ΔT averaged out on 1 mm of length at 2 mm (B), 5 mm (D) and 10 mm (F) from applicator tip, at three quotes: 0 mm (blue line), 2 mm (green line) and 4 mm (red line).

Since a laser-irradiated tissue presents a high spatial temperature gradient in a limited region sorrounding the applicator (diameter up to 3 cm [2]), the ideal sensor for temperature estimation should be punctual. This study demonstrates that a small sensor (i.e., FBG 1 mm-long) is more suitable when the region of interest is close to the applicator tip (e.g., 2 mm). As a matter of fact, differences between sensors with length of 10 mm and 1 mm reduce at higher distance (e.g., at 10 mm, both FBGs measure ΔTmax up to 8 °C), as also predicted by the theoretical model.

At the best of our knowledge, this is the first study which theoretically and experimentally assesses the influence of sensors size on T measures during LITT. Studies about distributed temperature sensing system using chirped FBG have been presented; for example, Li *et al.* [16] propose a small size fully distributed FBG sensor with a spatial resolution of 0.25 mm to monitor temperature inside the tissue irradiated with P=1 W, but there are not experimental results in tissue application. More studies have been presented considering T measurement at several positions [7, 12]; for example, Lippert *et al.* [17] use thermocouples at four distances from applicator tip. They measure a ΔTmax on tongue tissue of 44 °C at 6 mm from tip and P=5 W (*vs* our 50 °C at 5 mm and 6 W [12]).

V. Conclusions

Herein T assessment of pancreas undergoing LITT is carried out. Measures are performed, within two laser-irradiated *ex vivo* pancreases, at different distances from the applicator.

Theoretical and experimental results show the strongly influence of FBG dimensions on T. In particular, the lower is the distance from the applicator the higher is the influence of sensor size on temperature measures.

Further trials could be useful to monitor temperature with FBGs at other distances from the tip, at different laser settings and with different dimensions of sensors. These tests could allow to evaluate the sensitivity of the numerical model and to tune it for the prediction of thermal response of pancreas undergoing LITT at laser settings of clinical interest. Moreover, the use of optical fibers with three embedded FBGs could be pivotal in *in vivo* applications, because this configuration provides T measurements in three positions by introducing only one fiber.

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