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Temperature profile reconstruction during tumor laser ablation using percutaneous all-fiber smart applicators

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ABSTRACT

The paper discusses all-fiber applicators for the percutaneous laser ablation of tumors, which integrate very dense fiber Bragg grating arrays to add quasi-distributed sensing capabilities. First an assessment of the temperature map distribution reconstruction from the measurements is presented and the impact of some non-idealities is studied; then the developed probes are used to analyze different laser operating conditions, comparing the measurements in ex-vivo porcine livers with modeling expectations.

Keywords: Tumor laser ablation, Fiber Bragg grating sensors, Quasi-distributed fiber sensing, Characterization of fiber optic temperature sensors

1. INTRODUCTION

Cancer incidence is increasing worldwide and its prognosis is still strongly dependent on the affected organ, the tumor type, and its stage and dimensions. Indeed, while efficient treatments have been developed for some neoplastic forms, other forms have still quite low survival rates. An example of the latter is the liver cancer, which is predicted to become the sixth most diagnosed cancer and the fourth cause of death. The most diffused liver cancer is the HepatoCellular Carcinoma (HCC), which can be originated by hepatitis B and C, alcohol abuse, obesity and type 2 diabetes.[1–3]

HCC is difficult to be treated with traditional surgical approaches because of the complex structure of the liver and of its distinctive features, such as the friability and its venous reticulum; hence, the increasing interest for alternative mini-invasive procedures to locally destroy malignant cells by raising the temperature above the cytotoxic level (i.e, above around 50 °C). Among them, stands Laser Ablation (LA), which exploits the absorption of laser light by the tissue to increase the temperature and thus to induce the cellular necrosis.[4,5] In the specific case of liver or of other deep-laying organs, minimally invasive LA procedures can be implemented by delivering the laser beam through a miniaturized optical fiber applicator inserted in the skin via a needle.

Despite the advantages, the deployment of LA still lags behind the more popular RadioFrequency Ablation (RFA) and MicroWave Ablation (MWA), both for historical reasons – LA is the most recently introduced – and because, given the more localized effect, it requires a tighter control of the induced temperature. For example, in the case of HCC, the large variability in the light absorption and scattering coefficients and the possible presence of blood vessels make very difficult to predict the laser parameters to obtain a certain temperature distribution. Moreover, as the temperature raises, the tissue undergoes a dehydration that further changes the optical and thermal parameters, not only making more critical to avoid carbonization, but also shrinking the treated area with respect to the expectations. It is therefore evident that a real-time monitoring of induced temperature is fundamental not only to avoid such carbonization, but more in general to optimize the treatment result.[6] However, measuring the temperature during LA procedures is quite complicated because it can be done only either through imaging techniques (e.g., Magnetic Resonance Imaging, MRI, Computerized Tomography, CT, or Ultra-Sounds, US) or with all-fiber sensors since metallic temperature sensors, such as thermocouples, would produce artifacts due to their interaction with the laser beam.

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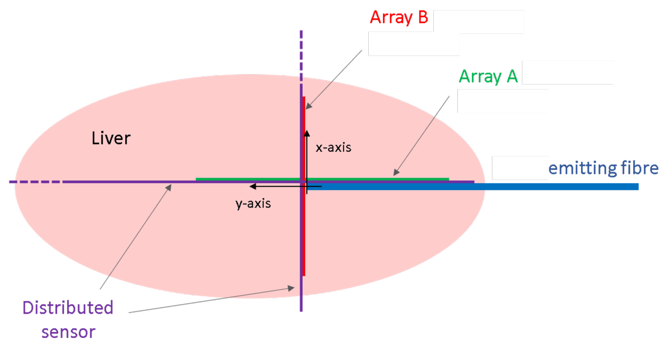

Figure 1. Scheme of the position of the beam delivery and of sensing optical fibers on a liver slice.

In order to overcome these limitations, over the years the authors have developed all-fiber applicators with embedded sensing capabilities, specific for the laser ablation of tumors in deep-laying organs.[7,8] These applicators are capable of measuring the temperature in some points along the laser delivery fiber thanks to the integration of Fiber Bragg Grating (FBG) sensors. Full treatment planning tools, however, would require an estimation of the temperature over the entire tumor mass and this, in turn, would require using multiple applicators of the proposed type, which is not possible without worsening the invasive impact. Moreover, even accepting this drawback, it would be practically impossible to ensure the relative positions among the applicators as defined during the treatment planning phase, and this would lead to non-negligible errors due to the large gradients typically present during LA procedures. Therefore, since the only easy access to temperature measurements is along the delivery fiber axis, the challenge becomes how to estimate the temperature in the entire tumor area from some measurements taken along a single section only.

A possible solution is to complement the temperature measurements with a suitable thermal model to reconstruct the entire temperature map by matching the predictions along the cut defined by the delivery fiber direction with the measured values. Clearly, an accurate temperature map reconstruction requires a very good estimation of the optical and thermal properties of the considered organ tissue and the measurement of the temperature in as many points as possible. The latter requirement translates into the necessity of a dense array of thermometers to implement a quasi-distributed sensing system with small separation between two consecutive sensing points, if not in the use of a truly distributed sensing system. The latter may be more attractive because of the very high number of sensing points; however, it is not the optimal choice since typical high spatial resolution distributed sensing systems have responses in the order of seconds[9] and this severely limits the applicability of the heat pulse method,[10] which is the approach we devised to evaluate the tissue thermal parameters. Indeed, given the variability of biological tissues, optical and thermal parameters cannot be estimated only from literature data, but must be measured before any ablation procedure. This can be practically implemented using a modification of the cited heat pulse method. These aspects are not analyzed in this paper, which instead is focused on the evaluation of the possibility to measure the temperature using fiber-based temperature sensors based on dense FBG arrays and on the investigation of the main sources of errors.

2. MATERIALS AND METHODS

As previously mentioned, the objective of the work here described is to precisely measure the temperature distribution across the applicator during simulated LA procedures in internal organs. These data will be later used to estimate the optical and thermal parameters of the tissue and, from them, to predict the temperature distribution in the entire tumor mass through a proper model that describes the evolution of the LA procedure. In particular, as a preliminary phase, the paper focuses on the possibility to reconstruct the temperature distribution from the measurements obtained with FBG arrays, using as reference the readings from a distributed sensing system made by a span of single-mode fiber interrogated with a coherent OFDR instrument. To better analyze the impact of the position of the sensors in the temperature map reconstruction, some parallel and perpendicular positions around the applicator have been tested, as shown in Fig. 1.

FBGs are periodic structures inscribed in the core of optical fibers, resulting in a notch filter behavior, for which the spectral portion around a specific wavelength, the Bragg wavelength λ_B, is back-reflected, while all the other wavelengths are transmitted through the fiber. FBGs can be used for sensing applications because the Bragg wavelength shifts both with temperature (T) and strain (ε). As a fist approximation the dependency of the Bragg wavelength shift with temperature and strain can be expressed using a linear relation:

$$\Delta\lambda_B = K_\varepsilon \cdot \varepsilon + K_T \cdot T \tag{1}$$

where K_ε is the strain sensitivity, while K_T is the temperature sensitivity. The cross-sensitivity between temperature and strain is a limiting aspect because the tissue deformations with heat can induce strain and thus introduce artifacts in the temperature estimation. These effects can be mitigated by inserting the gratings into capillaries, although this changes also the heat distribution; thus this approach requires a careful analysis to assess its impact on the actual temperature map reconstruction.[11–13] For the measurements described in this paper, different arrays with up to 20 multiplexed along a single fiber FBGs have been written in single-mode telecom-grade optical fibers with a femtosecond laser using a point-by point technique.[9] The FBGs in the arrays are densely packed, allowing a spatial resolution that can be as low as 1 mm. Prior to their use in ex-vivo experiments, all the FBG arrays have been fully characterized in a climatic chamber to relate the temperature changes with the Bragg wavelength shift of each FBG. Then, the FBG arrays have been read using a LUNA-MicronOptics HYPERION interrogator, which is a multi-channel FBG measurement instrument based on a narrow-band laser swept over the C-band of the optical communication windows.

The distributed measurement system used as a reference is based on a LUNA OBR4600, an instrument that employs coherent OFDR to detect variations in the Rayleigh scattering due to local temperature changes with a spatial resolution down to few tens of micrometers and a temperature resolution of about 0.1 °C.[14] Different types of G-652 compliant single-mode fibers with only the primary coating (i.e., with a total diameter of 250 µm) to minimize the invasive impact and ensure a good sensitivity to temperature have been considered; this is because through tests in climatic chamber it has been found that the polymeric coating can induce stresses that result in hysteresis and deviations from linearity. Therefore, not all the fibers are suited for the intended applications. Than, tests in climatic chamber are necessary also to find the specific coefficient that relates the spectral shift due to the variations in the Rayleigh scattering with the local temperature variations.

The temperature map distribution around the heating laser delivery fiber has been sampled forming a grid of parallel lines (spacing of 3 mm) with the fiber read by the LUNA OBR4600, as shown in the setup in Fig. 2.

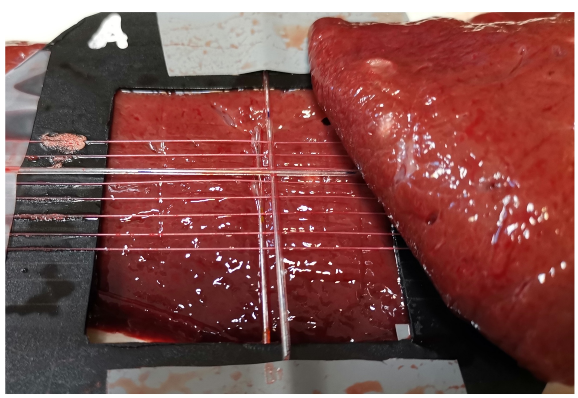

Figure 2. Practical implementation of the scheme in Fig. 1.

3. EXPERIMENTAL ACTIVITIES

The experiments have been carried out on ex-vivo bovine livers. In order to ease the positioning of the sensors around the applicator, the livers have been sliced, the sensing fibers accurately positioned on one of the surfaces, and then covered with another slice, paying attention not to move the fibers (Fig. 2). The use of slices has also allowed acquiring a thermal map with a camera for further validations, albeit only at the end of each ablation procedure by quickly removing the upper slice.

In all the experiments, the laser beam has been provided by a home made laser source built around a high-power laser diode (emission up to 20 W) emitting at 915 nm in a 200 µm fiber. Since the goal is to evaluate the possibility to use the measurement to recover the thermal parameters and not to conduct an actual laser ablation procedure, a maximum power of 2 W has been used in most of the experiments.

Fig. 3 reports the comparison of the readings of an FBG array positioned on one side of the delivery fiber with those of the fully-distributed method. As the quasi-distributed sensing system supports a faster acquisition rate than the distributed system, four readings could have been acquired in the time frame corresponding to the distributed sensor acquisition rate. The temperature profiles obtained from the two types of sensors are very similar, within the uncertainties characteristic of each sensing approach and the discretization forced by the FBG array. As expected, a certain asymmetry in the induced temperature profile between the front and the back of the applicator can be seen. Fig. 4 is similar to Fig. 3, but for one of the sensors perpendicular to the delivery fiber. Again the curves are very similar, although it is evident the error in locating the temperature peak in the quasi-distributed sensing system due to the spatial discretization of the used FBG array. Denser FBG arrays, such as those with only 1 mm spacing, can solve this problem, but a figure obtained during a test with a less dense array is reported on purpose to highlight the importance of properly choosing the FBG array parameters.

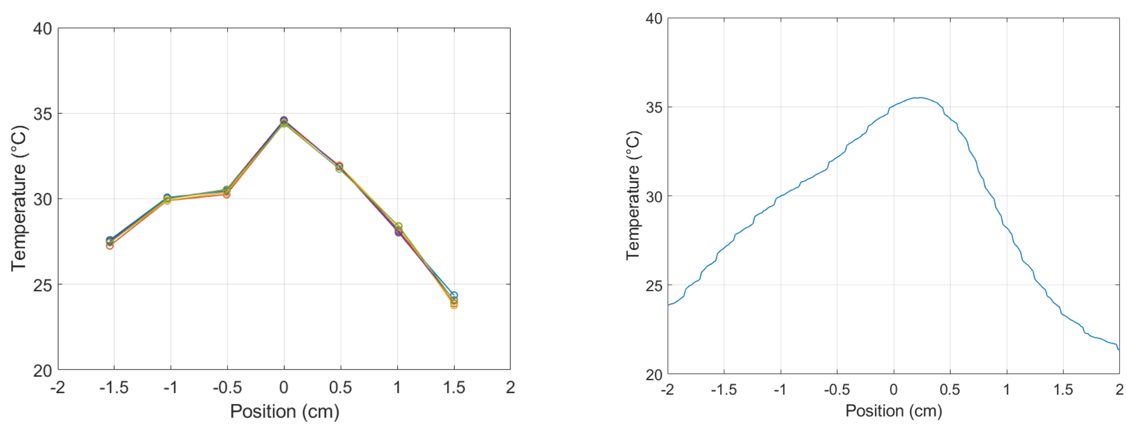

Figure 3. Temperature of quasi-distributed (left) and fully distributed (right) sensors in close contact with the delivery fiber. The origin of the ordinates is at the delivery fiber tip.

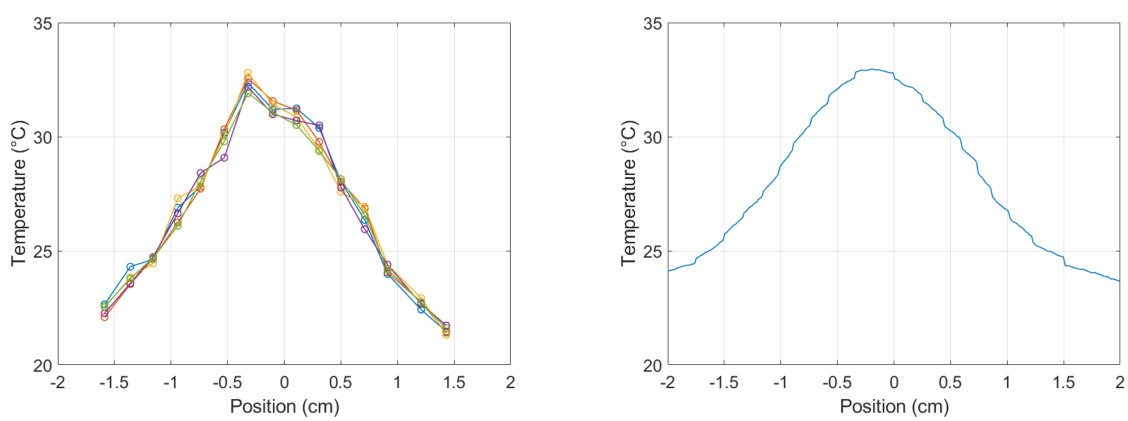

Figure 4. Temperature of quasi distributed (left) and fully distributed (right) sensors positioned perpendicular to the delivery fiber at 1 mm. The origin of the ordinates is at the delivery fiber tip.

Fig. 5 shows an example of reconstructed temperature maps using the measurements taken with the quasi-distributed fiber sensors.

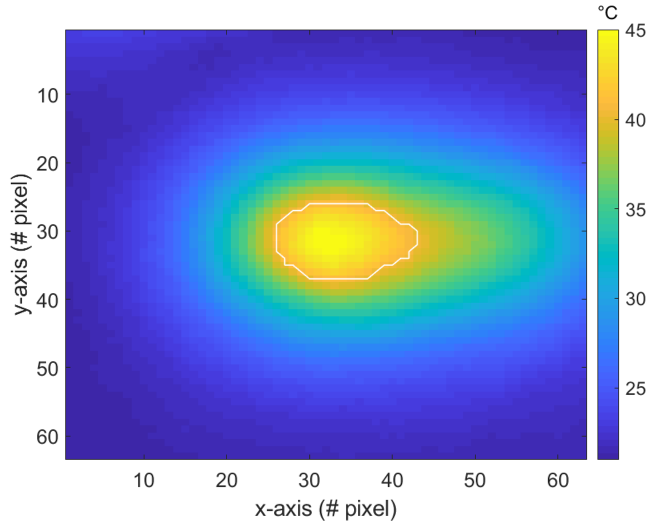

Figure 5. Reconstructed thermal map of a laser ablation. The reported temperatures are the increase with respect to the ambient temperature and the white contour identifies the region corresponding to higher than cytotoxic values.

As a preliminary demonstration of the utilization of the measurements, Fig. 6 shows the comparison between the actual measured temperature decay along the delivery fiber axis and the prediction of a model based on the solution with a finite element method of the heat distribution in circular rod of liver, having the thermal parameters obtained from the measured temperature decay along the delivery fiber direction. Having used ex-vivo sample, for the moment the model is simplified and does not take into account perfusion; however, the very good agreement between the curves in Fig. 6 is encouraging for the possibility to find a treatment planning procedure in which the model parameters can be extracted for a specific case from a preliminary heating with sub-cytotoxic temperatures.

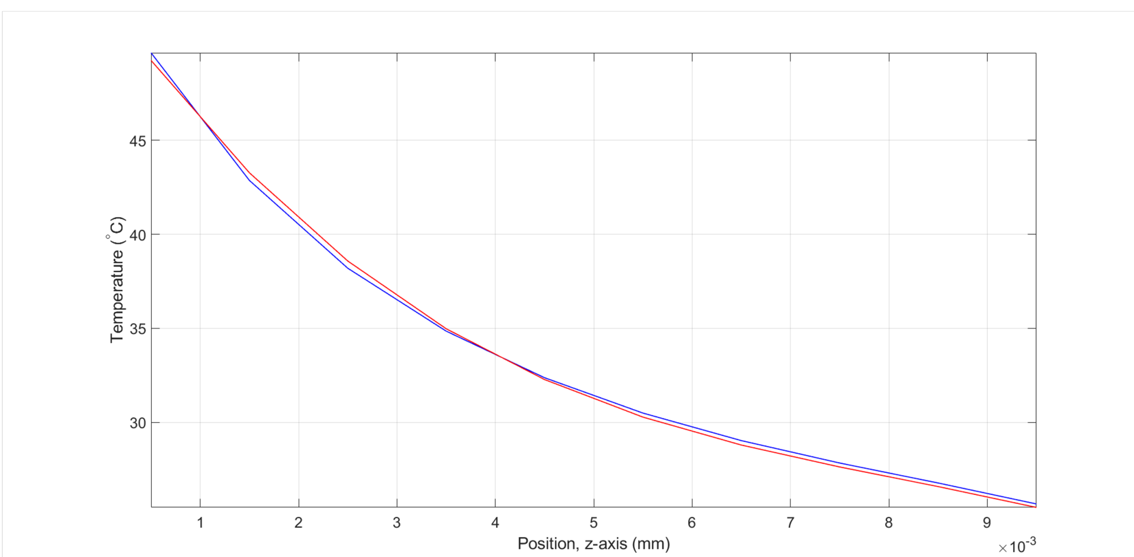

Figure 6. Comparison of the actual measured temperature decay along the delivery fiber axis and that obtained from simulations.

The accuracy of the reconstructed temperature profile is strongly affected by the errors in the temperature measurements and decreases as the distance from the heat source increases. For example, numerical experiments carried out assuming Gaussian noise with mean equal to 0 and variance of 1, have shown that when the distance exceeds 3.5 mm the error becomes larger than 0.1 °C, as shown in Fig. 7. This may still seem an acceptable value, but instead it becomes critical in actual LA procedures when there is the necessity of an accurate estimation of the treated area margins.

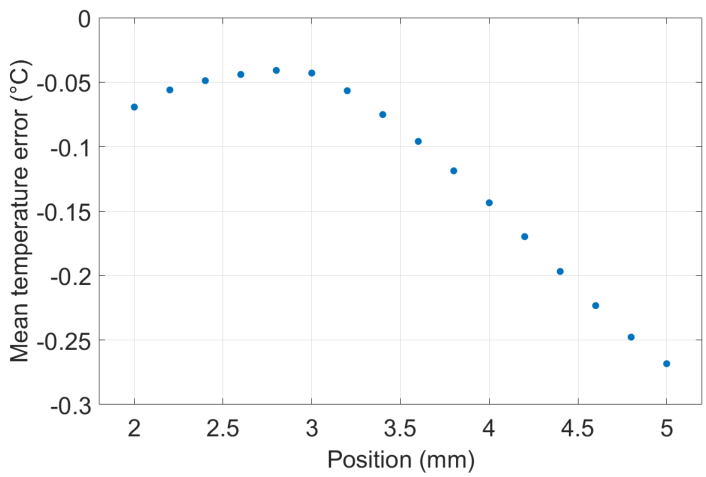

Figure 7. Temperature error varying the distance from the laser source, with power 2 W.

4. CONCLUSIONS

The paper has analyzed the possibility to reconstruct a temperature map during laser ablation in internal organs using quasi-distributed thermometers based on FBG arrays. The comparison with truly fiber-based distributed sensing systems has confirmed the validity of the proposed approach, although care is needed to reduce the impact of the errors, which otherwise can lead to large errors in the estimation of the margins of the area subjected to treatment.

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