Experimental Qualification of Fiber Bragg Grating Sensors for Temperature Monitoring in Laser Ablation

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***Abstract* – The paper discusses the main issues related to the usage of Fiber Bragg Gratings as temperature sensors when they are employed to monitor percutaneous laser ablation of tumors. After a description of the main problems related to this specific application, two different setups for the characterization and the qualification of these sensors are described and preliminary results are presented.**

Keywords* – *Biomedical measurements, thermal ablation, fiber Bragg gratings, temperature sensors

I. INTRODUCTION

Different treatments for primary and, especially, secondary malignancies in patients affected by non-operable tumors have attracted large attention in the last decades; among them, the ablation of malignant tissues is particularly interesting because of the effectiveness and minimally-invasive impact [1]. The majority of these procedures involves the delivery of thermal energy that locally increases the cell temperature above cytotoxic levels, thus causing the cell death. The treatment effectiveness depends on both the temperature and the exposition time, being the two quantities inversely related. As an example, the same cytotoxic effect can be obtained maintaining the cell temperature at about 45 °C for a couple of hours or setting a higher temperature, such as 100 °C, for few seconds only. Temperatures above 100 °C should be avoided since they cause tissue carbonizations and this is not appropriate, besides for changing the thermal properties. On the other hand, temperatures below 43 °C do not produce irreversible damages [2].

Different technologies can be employed to deliver the required thermal energy for tissue ablation. The most common approaches take advantage of electromagnetic waves at different wavelengths ranging from radio-frequency (Radio Frequency Ablation, RFA) to microwave (MicroWave Ablation, MWA) and to optical radiation produced by lasers (Laser Ablation, LA). For each of these, the appropriate applicator is inserted percutaneously through a few millimeter hole into the patient body and positioned in the tumor location. The accurate knowledge of the applicator location is of great importance and to this aim different imaging techniques (echography, tomography, etc…) can be employed by the surgeon during the treatment.

The usage of laser radiation routed through optical fibers presents several advantages with respect to MWA and RFA, such as the probe small dimension and its intrinsic safety. Probes for LA have been developed and employed in medical field targeting different applications, such as the treatment of the cardiac atrial fibrillation, but their usage in oncology is still limited because of the lack of sensors suitable to monitor in real-time the tissue temperature during the medical procedure. Applicators for RF or MWA ablation are often equipped with temperature sensors and also non-invasive approaches based on imaging thermometry have been investigated. Unfortunately, the former involve expensive facilities which are not always available and return results still not so accurate [3-4]; the latter, on the contrary, provide the required accuracy but they cannot be employed during a Laser Ablation because the sensor embodiment and its metallic wires perturb the laser radiation pattern.

Fiber sensors, such as Fiber Bragg Grating (FBG) sensors in bare fibers, are thus the best candidates in this particular application because of the fiber dimension and of their all-dielectric nature. Other possibilities that have already been employed in this field involves OFDR and LCFBG but none of these resulted appropriate for the application [5]. Accurate temperature sensors based on FBGs are commercially available, but their embodiment is bulky and, thus, they cannot be embedded in LA probes. For this reason, bare fiber sensors are embedded in custom probes to meet the dimension requirements. The calibration of bare FBG sensors is routinely performed against standard temperature sensors using climatic chambers or other controlled environments where the temperature is maintained constant at predefined values. Unfortunately, this procedure does no guarantee the sensor qualification for LA since the calibration conditions significantly differ from the operative conditions. Actually, LA is usually performed in organs where the occurrence of a metastasis is higher, as in the case of pancreas and liver. Considering the last one, its low thermal conductivity results in a very high temperature gradient, as large as 10 °C/cm or higher. Standard FBGs present a length that spans from a millimeter to a couple of centimeters; therefore, the presence of non-uniform temperature distributions can introduce unacceptable errors. Moreover, bare FBGs can be affected by influence quantities, such as the bending introduced by respiration.

In this paper, the most relevant aspects related to the qualifications of FBGs as temperature sensors for LA are investigated and the main uncertainty contributions are highlighted. The effects of the embodiment on sensor performances, such as the cross-sensitivity to strain, time constant variation, laser radiation absorption and the non-negligible sensor length are here analyzed. Two characterization setups are eventually proposed in order to evaluate the aforementioned contributions

II. FBG WORKING PRINCIPLE AND INTERROGATION SYSTEM

FBG are fabricated by realizing a periodical modulation in the core refractive index of a single mode optical fiber in order to reflect a specific wavelength that depends on the grating period and the core effective index. As these two quantities are related to the temperature and to the axial strain applied to the grating, the spectrum of the reflected light presents thus a narrow peak whose position λ_B is related to the temperature and to the axial strain applied to the grating. The relation among these quantities can be described using a linear model:

$$\Delta\lambda_B = k_\varepsilon \varepsilon + k_T \Delta T$$

where the sensitivity coefficients between wavelength shift and temperature and strain are approximatively k_ε=1 pm/με and k_T=10 pm/°C, respectively, for gratings having λ_B around 1500 nm. It is evident that the FBG acts as a temperature sensor provided that the strain is known or is maintained constant.

A typical interrogation system for FBGs is shown in Fig. 1 and it is based on a broadband source (SLED) and an optical spectrum analyzer able to detect and to track the Bragg wavelength. The light from the source is routed to the FBG sensor and from this to the spectrometer by an optic circulator. Optical spectra are then processed in order to locate the reflected light peak with the required resolution.

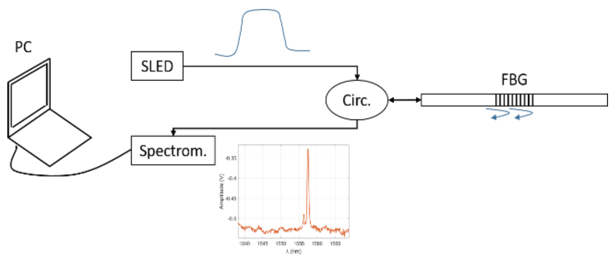

Fig 1. Schematic representation of the basic interrogation system for FBGs.

Since the resolution of the available spectrometer is 100 pm, the corresponding temperature resolution would be about 10 °C, when a sensitivity of 10 pm/°C is considered. In applications such as LA, however, the desired resolution is better than 1 °C. Therefore, signal processing techniques must be applied to achieve such target resolution. For example, through a Gaussian fitting of the acquired optical spectra, the temperature resolution can be improved to 0.1 °C.

III. FBG RELATED ISSUES

A. Strain sensitivity

The cross-sensitivity of FBGs to both temperature and strain, shown in Fig. 1, raises a series of problems in the considered application. The optical fiber containing the grating is inserted into the patient percutaneously through a proper needle, so both axial strain and bending may occur, appearing as a temperature variation. Moreover, even if a perfect insertion is performed, the motion due to the patient respiration could introduce artifacts.

The problem can be avoided by designing a suitable embodiment that encapsulates the FBG and prevents unwanted strain or bending from reaching the fiber. Usually FBG sensors are inscribed in standard telecom single mode fiber (10/125μm), but also in the same fiber used for the delivery of the laser light, which has a larger diameter (in our case it is a double cladding fiber with 20/400/440 μm), as reported in [6]. In both these scenarios, the fiber portion containing the FBG (usually in the end-tip) can be encapsulated in a glass or PTFE capillary. With this solution, the capillary withstands external stress leaving the FBG affected by temperature only.

The usage a capillary solves the strain problem but affects the thermal properties of the probe. Actually, both the thermal capacitance and the sensor time constant increase. Tests have been carried out with bare FBGs and with FBGs protected by a glass capillary having external diameter of 1 mm [7]. The results have shown that the time constant with the capillary is of about 0.2 s, a value that can be considered negligible for the intended application.

Anyway, the capillary not only affects the sensor dynamic response, but it can also absorb part of the laser radiation and modify the temperature distribution inside the tissue.

The problem of the absorption of the laser radiation becomes more evident when the fiber for the laser delivery and the fiber for sensing share the same probe because the laser is in close contact with the FBG [6].

Tests were carried out in free space, placing the laser in front of the sensor and recording the temperature increase. The results have shown a temperature increase of 0.5°C/W and thus non negligible, even when a quartz capillary is employed. Moreover, a glass capillary is very brittle and requires extreme care in handling so alternative polymeric materials, such as PTFE, are under investigation. Errors due to strain and absorption could be higher for these materials than in the case of glass, and thus a characterization is required before the sensor usage.

The presence of the capillary, or of another embodiment, also affects the tissue temperature distribution. As mentioned at the beginning of this section, during LA the tissue temperature presents a large gradient due to the high tissue thermal resistance. The sensor thermal properties, such as its thermal resistance and capacity, can be significantly different from those of the tissue and thus the capillary may affect the temperature of the tissue under measurement. This is a systematic error than can be hardly modeled because of the large variability of the tissue properties. The solution is to minimize the error by reducing the probe dimensions and using a material having thermal propertied close to the tissue, but the actual error can be assessed only though an experimental approach.

B. FBG length

The relation between Bragg wavelength and temperature shown in Eqn. 1 is derived for a uniform temperature along the grating, when is straightforward to model the linear wavelength/temperature dependence. Nevertheless, in operative conditions, this relation is not so trivial to be found.

Off-the-shelf sensors can be as long as 2 cm and thus the sensor surface is exposed to a temperature difference that can be as large as 20 °C. FBG response in the presence of non-uniform temperature has been addressed in a theoretical way, and the results have shown that the sensor returns the average temperature for a linear temperature distribution.

This feature does not represent a problem in several applications, like in structural monitoring where FBGs are frequently employed since the temperature can be considered uniform over their length. On the contrary, during LA the interest is mainly on the maximum temperature reached close to the laser probe since tissue carbonization must be avoided. The sensor provides thus an underestimation of the quantity of interest. This problem can be taken into account when the temperature distribution is known and thus, properly compensated.

When this information is not known it is still possible to reduce the error using shorter FBGs. These sensors are not so widely diffused and also presents a larger spectral response and thus a worse temperature estimation but, when cascaded, allow a quasi-distributed temperature measurement to be obtained.

Fig. 2 shows the sensor optical response of a single FBG having a length of 2 cm and an array of 2 FBGs having a length of 1 mm and spaced by 4 mm. The single FBG provides a narrow peak and thus a better estimation of the temperature. The array presents two large optical peaks, one for each measurement point.

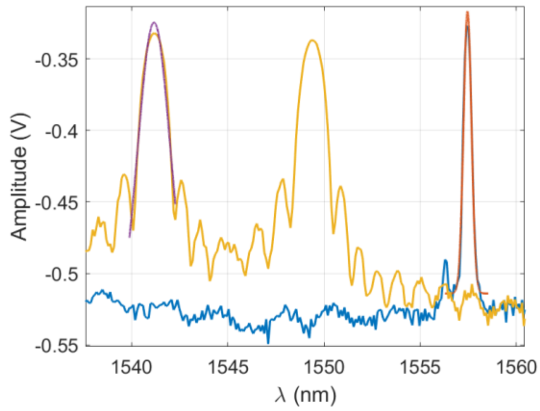

Fig 2. Different FBG responses. The yellow one corresponds to an array of two FBGs of 1 mm length, while the blue one to an FBG of 1.5 cm length. Left and right peaks report the superposition of a Gaussian fitting (red line) employed to better estimate the peak position and thus the temperature.

IV. CHARACTERIZATION SETUP

Given the issues exposed above, a characterization setup where sensors are calibrated in uniform temperature conditions is not adequate to qualify these fiber sensors for LA. Moreover, most of the aforementioned issues can be quantified in an experimental way only, because of the lack of information, such as the thermal properties of the sensor embodiment and the effect of the sensor on the tissue temperature.

In order to overcome these drawbacks, we are developing characterization setups useful to evaluate the sensor performance in the presence of temperature distributions and thermal conditions close to those typically found during LA. In this section, a highly reproducible setup able to generate linear temperature gradients and a setup that mimics real ablation conditions are described.

A. Setup for linear gradient

The first setup here described has been devised to generate non-uniform but known temperature distributions in a highly reproducible way. The setup is based on a rectangular metallic bar, which is heated or cooled at its ends. In the middle the bar hosts the FBG sensor that will be subjected to linear temperature gradient. Fig. 3 reports a prototype of the proposed system. In this case, one side is heated through a resistor fixed on the bar, while the opposite side is cooled with a Peltier cell.

Between the resistor and the Peltier, the bar has been shrunk for a short portion in order to have higher thermal resistance and thus higher gradient. In this particular case, since the shrunken part has two flat edges, the temperature gradient is linear, as confirmed measurements taken with an infrared camera. The setup embeds two electronic sensors fixed on the back side of the bar in the region where the fiber sensor is placed.

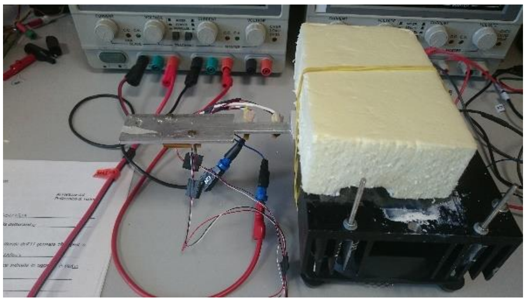

Fig 3. Prototype of the setup for linear temperature gradients.

Test results with bare fibers and fibers embedded in glass capillary showed that, under linear temperature distribution, the FBG returns the average temperature.

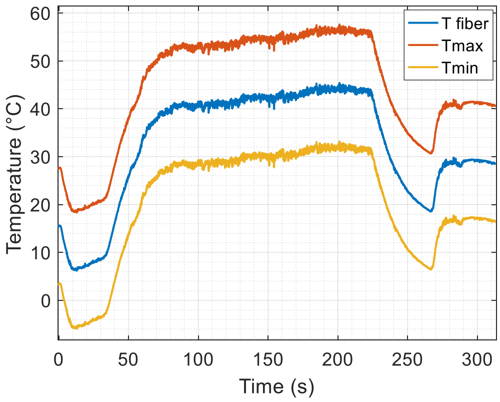

Fig 4. Test of a bare FBG on the bar with linear gradient. Tmax and Tmin are the temperatures from the electronic sensors, while Tfiber is the FBG measurement.

In case of an array of short FBGs, the setup is useful to assess the array capability to resolve small temperature differences over short distances. Tests are currently in progress and results concerning this application will be presented in the full paper.

B. Setup for sensor tests in the presence of thermal gradients generated by a laser

The setup previously described does not provide information about the effect of the laser on the sensor and the sensor behavior in the presence of non-linear temperature changes, which frequently occur during real LA.

In order to reproduce a thermal condition that mimic actual LA cases, a phantom of the liver tissue has been realized. The phantom has both the thermal and the optical properties of an ex-vivo liver, in which is possible to neglect the heat removed by the blood flow. This phantom is made of agar gel in which the optical properties have been modified by loading the gel with dark Indian ink.

This jelly material is soft so it is very easy to insert a probe made of glass capillary containing the fiber sensor. Moreover, the probe can also embed the fiber for the laser delivery and it is thus possible the test the effect due to the laser absorption.

Fig. 5 shows the phantom employed for our tests, where both the sensing fiber and the laser delivery fiber have been inserted. The knowledge of the actual phantom temperature is mandatory to verify the correctness of the fiber sensor response. This information cannot be obtained using auxiliary sensors, as in the previously described set-up, because of the laser radiation effects on electrical sensors. This problem has been solved by using an infrared camera to allow carrying out non-contact measurements. The camera, however, gives the external surface temperature only; therefore the phantom has been subdivided in slices in order to have an easy access the the inner part.

Fig 5. Liver phantom used for the thermal tests.

With this configuration the laser is turned on until the temperature reaches a steady state; then the upper slice is quickly removed thus exposing the sensors and the surrounding hot surface. The camera can thus record the surface temperature (Fig. 6) that will be compared with fiber sensor measurements.

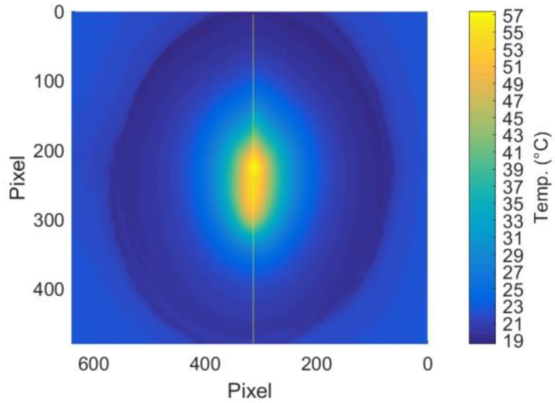

Fig 6. Thermal image of the temperature increase induced in the liver phantom. The FBG inside this region (along the vertical yellow line) measured 55 °C, a value in agreement with the infrared camera.

V. CONCLUSIONS

The main issues related to the qualification of FBGs as temperature sensors for Laser Ablation have been discussed in this paper. This kind of measurement technology has been already devised to provide a distributed or quasi-distributed monitoring of the tissue temperature increase [3-8]. Due to the particular kind of application, large errors can arise the measurements due to the presence of large temperature gradients, but this potential drawback is often not considered or underestimated. This work reviews the main problems related to the usage of FBGs temperature sensors employed to monitor LA and it represents towards the traceability of the measurement results. It turned out that the standard calibration procedure is useful to provide the sensor sensitivity but it is too simple to assess the sensor accuracy in the presence of the severe thermal gradients which frequently occur during LA.

Two setups have been thus devised to test the sensor in the presence of temperature distribution and thermal conditions similar to a typical LA. A setup based on a metallic bar, which is able to produce a highly reproducible and known linear temperature gradients, is employed to test both large FBG and array of short FBGs. The traceability of the generated thermal gradient is entrusted to electronic sensors and to a careful thermal design. This setup is employed to verify the capability of large FBGs to provide the average temperature. In case of FBG arrays, the setup is also useful the test the array spatial resolution, that is, the array capability to distinguish different temperatures at short distances. This setup is currently under calibration.

A second setup based on agar jelly has been thought to mimic real ablation conditions. The thermal gradient inside the jelly is produced with a laser. The FBG measurements are thus compared with an infrared camera outputs, since no electrical sensors can be here employed. The setup is not well reproducible and the expected measurement uncertainty is larger than in the previous setup because of the relevant contribution from the infrared camera and from the test procedure. Anyway, it is useful to assess the laser effects on the sensor measurements and to verify the sensor performance in the presence of non-linear thermal gradients and when the sensor is in close contact with the tissue emulator.

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