A MINIMALLY INVASIVE MICROWAVE HYPERTHERMIC APPLICATOR WITH AN INTEGRATED TEMPERATURE SENSOR

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Abstract: In the field of microwave hyperthermia and thermo-ablation, the use of minimally invasive applicators is recognized as a very promising means for the treatment of small, early stage, cancer lesions because a very thin applicator can be easily introduced inside the body and precisely directed towards a deep seated tumour using the most advanced 3D imaging techniques and surgical stereo-navigation. Minimally invasive applicators have been successfully employed for the treatment of bladder carcinoma and brain tumours but the accurate temperature monitoring of the heated tissue volume still remains an open problem. In this paper we propose a new minimally invasive applicator, integrating a low-cost metallic wired temperature sensor. The miniaturised endocavitary applicator consists of a asymmetric isolated dipole operating at 2.45 GHz. The very slim shape of the applicator allows to easily insert it into the lesion through a soft plastic tube (catheter) while a temperature sensor, properly embedded in the applicator body, measures the tissue temperature at the interface with the catheter surface. An electromagnetic analysis based on the Finite Integration Technique (FIT) and experimental verifications over a tissue sample proved that a coaxial choke, enclosing the temperature sensor wires, allows localize the heating pattern in a restrict volume while drastically reducing measuring artefacts due to the perturbing effects induced by the probe leads.

1 INTRODUCTION

Microwave endocavitary and interstitial hyperthermia has been widely investigated in the past decades as localised thermal therapy for cancer treatment. Many thin applicators have been developed and used in therapeutic applications with valuable results. Most of them are essentially constituted of an insulated monopole, dipole or helix feed through a thin coaxial cables. However several technical solutions (Turner, 1986; Tumeh and Iskander, 1989; Camart *et al.*, 1996; Lin and Wang, 1987; Cerri *et al.*; 1993; Saito *et al.*, 2000) have been proposed in order to localize the heating in a restrict area of tissue around the tumour and to avoid accidental and unwanted hot spots in the healthy tissue.

The effectiveness of a microwave thermal therapy depends not only on the radiative properties of the applicator used but also on the reliable and accurate temperature control during the therapeutic treatment. Impedance tomography, microwave radiometry, magnetic resonance imaging (MRI) and also methods based on ultrasounds are very attractive non-invasive temperature-monitoring techniques. Despite these promising prospects, up till now a very accurate measure of depth tumours temperature can be obtained only by invasive techniques, because only thermocouples, thermistors or optical fibres sensors seem to give the required measuring accuracy, spatial resolution and real-time response. Temperature monitoring by invasive sensors is common practice in the hyperthermic treatments. Usually, very thin probes are separately inserted in the tissue near the antenna and moved ahead and back in order to better estimate the effective heating pattern. In that case only optical fibre sensors could be employed because metallic wired probes located near the antenna strongly interact with the radiating element, producing uncontrollable electromagnetic fields distortion and false temperature readings.

When the tumour is at its early stage a single applicator can be employed to heat the small volume of the lesion by inserting it as sketched in Figure 1.

In this case it is advantageous to integrate the temperature sensor directly in the applicator body in order to avoid additional traumas in the patient and to simplify the hyperthermic treatments itself.
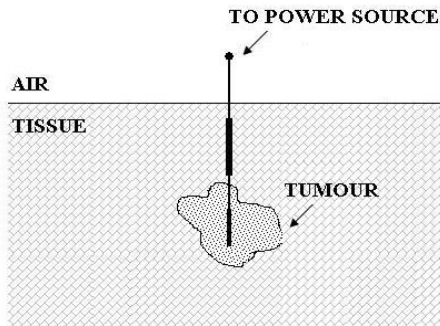

Figure 1: Single applicator heating a small tumour.

When lesions are more extended in volume an array of applicators could be required to uniformly heat the tumor, as sketched in Figure 2.
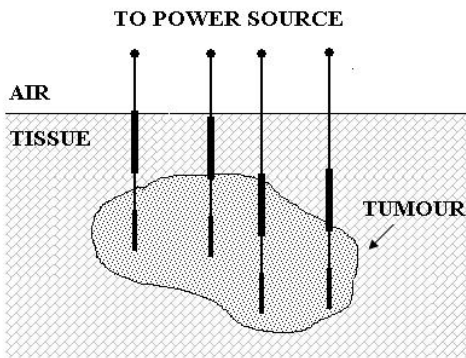

Figure 2: Array of applicators heating a medium-sized tumour.

Also in this case applicators with integrated temperature sensors can be usefully employed because the temperature distribution inside the tumor volume can be numerically estimated starting from the point measurements taken in correspondence of each radiating element, by combining tomography algorithms and bio-heat equation.

From the mechanical and structural point of view it is not difficult to integrate a very thin fiber-optic probe inside the core of the applicator without altering its original diameter. It is worth to note, however, that fiber-optic sensors are very expensive and delicate devices and thus not properly suited for

the production of a rugged mono-use applicators, as required for the routined clinical practice.

Another sensor that is practically unaffected by the strong EM fields existing near the antenna is the Bowman thermistors (Bowman, 1976) because its high resistance leads can not carry significant RF currents. Unfortunately also this sensor type is delicate and expensive to be fabricated.

Ordinary low-cost and sturdy thermometric sensors, as thermocouples and thermistors, use high conductance metallic leads for the connection to the measuring unit. Unfortunately, when they are placed too close to the applicator body, metallic wires cause unwonted electromagnetic (EM) interferences. In these cases dangerous and uncontrolled hot spots in the tissue can occur, as well erroneous (or very noisy) temperature readings.

This work suggests a novel technique for integrating a low-cost wired temperature sensor inside the body of a minimally invasive Microwave Hyperthermic Applicator (MHA) without perturbing the radiated fields.

2 METHODS

2.1 Applicator Design

The proposed interstitial/endocavitary MHA, depicted in Figure 3, essentially consists of a coaxial asymmetric dipole type antenna, radiating in the biological tissue through an insulating tube (catheter).
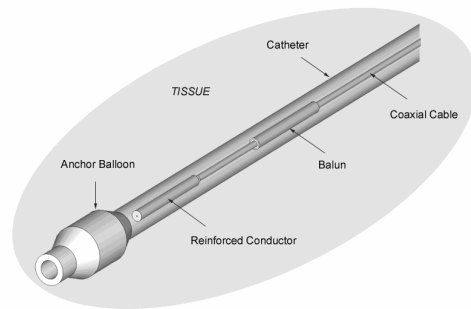

Figure 3: Microwave Hyperthermic Applicator (MHA) (length: 65 mm; thickness: 2 mm) working at 2.45 CHz.

In order to avoid unwanted heating of the healthy tissue, a coaxial balun has been introduced to block the back currents flowing on the surface of the coaxial feeding line (Longo *et al*., 2003). The diameter of the radiating upper arm of the dipole has been properly increased to improve the matching of

the MHA input impedance to the tissue (Jones *et al.*, 1988; Biffi Gentili *et al.*, 1995). Thank to its very thin shape (65 mm in length and 2 mm in thickness) the coaxial applicator can be easily introduced inside a small catheter (3-4 mm in diameter) and subsequently inserted into the lesion. The position of the applicator can be fixed *in loco* using an anchor balloon.

Starting from this basic applicator configuration, a wired temperature probe (thermocouple or thermistor) has been successively introduced into the catheter inside the balun as depicted in Figure 4.
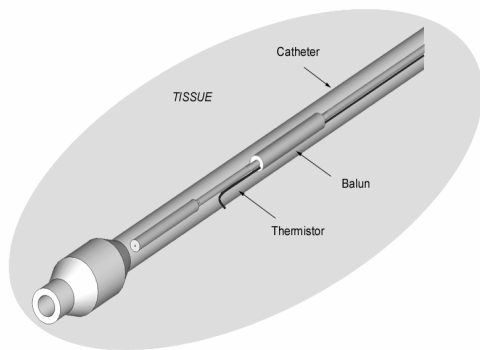

Figure 4: MHA with a temperature probe embedded inside the insulator of the coaxial choke.

The temperature sensitive tip of the probe is embedded in the catheter body in order to measure the temperature of the tissue at the interface with the catheter itself, where higher temperatures are expected. Miniature (SMT) chip inductors can be also inserted close to the sensing tip of the probe to isolate it from the radiating conductors of the coaxial applicator.

2.2 Numerical Analysis

Electromagnetic full-wave analysis has been employed to investigate the Specific Absorption Rate (SAR) distribution produced by the applicator and the perturbations due to the closeness of the metallic wires of the temperature probe. EM fields have been calculated with a time domain Finite Integration Technique (FIT) (CST Studio Suite, 2006) in a three-dimensional spatial domain constituted by a muscular tissue volume including the MHA model.
Perfect Matched Layers (PML) boundaries condition (Berenger, P. 1994) have been used in order to limit the computational domain to $140 \times 60 \times 60$ mm^3 volume. However the thin profile of the applicator and the high permittivity of the human muscle tissue ($\varepsilon_r = 52 - j13$ @ 2.45 GHz) required a very little mesh size ($< \lambda/50$ at 3

GHz) which made heavy the computation load. EM simulations, performed in the 2 to 3 GHz frequency range, run for 2 hour with an Intel Pentium III @1GHz with 1.5 GB RAM memory

2.3 Experimental Set-up

Different types of soft material phantoms with the same dielectric properties of human tissue have been experimented upon in the recent years to investigate SAR and temperature distribution produced by microwave applicators. An accurate phantom should closely represent the electromagnetic properties of the human body in the frequency range of interest and it should be easy to prepare and to handle.

When small volume of tissue are involved in the heating process, true biological tissue sample as a suine liver or a chicken breast appear more suitable to investigate power deposition near the applicator.

In our case the realized MHA prototype has been tested by using chicken breast. The prototype, ending with a SMA male connector, is connected to a microwave source working at 2.45 GHz, capable of a maximum power of 300 W continuous (CW) or pulsed. The input and reflected power of the MHA is monitored by a power meter connected to a bidirectional coupler while the temperature, measured by a thermocouple or a thermistor, is acquired by an A/D converter, recorded in a data file and simultaneously displayed on a PC monitor for the real-time direct control of the heating process.

3 RESULTS AND DISCUSSION

In order to accurately define the heating pattern volume and avoid unpredictable field distortion or undesirable tissue overheating, back currents flowing on the surface of the coaxial feeding line and currents induced on the metallic wires of the thermocouple was blocked by a common coaxial balun. As a result, at the 2.45 GHz operative frequency, the power deposition in the tissue is confined within a small well defined ellipsoidal volume wrapping the radiating section of the applicator body as shown in Figure 5.

As expected, numerical simulations evidence a focusing point of the EM fields near the sensing tip of the sensor that could be responsible of a localised hot spot and temperature overestimation. In order to reduce this unwonted spot, miniaturised chip inductors have been properly inserted in series to the leads of the temperature sensor near the tip to block any RF current flowing in the sensing element. In

our EM model we used lumped inductive elements with an inductance of 1 nH and we compared numerically this solution (Figure 6c) with the reference cases where the tip was kept floating (Figure 6a) or short circuited to the external coaxial conductor of the applicator (Figure 6b) that schematizes the applicator shown in Figure 5.
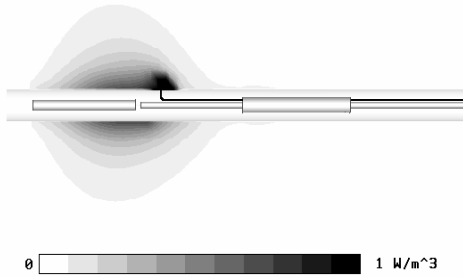

Figure 5: Normalized SAR distribution of a MHA with a temperature probe embedded inside the insulator of the coaxial choke.

If a thermistor is used as temperature sensing element instead of a thermocouple, the sensor tip is constituted by a semiconductor with conductivity ranging between 10^{-2} and 10^{-4} S/m. In both cases the simulation result of Figure 6c shows that the insertion of the micro-chokes practically eliminates the arising of hot spots near the sensing termination and therefore drastically reduces temperature measuring errors.
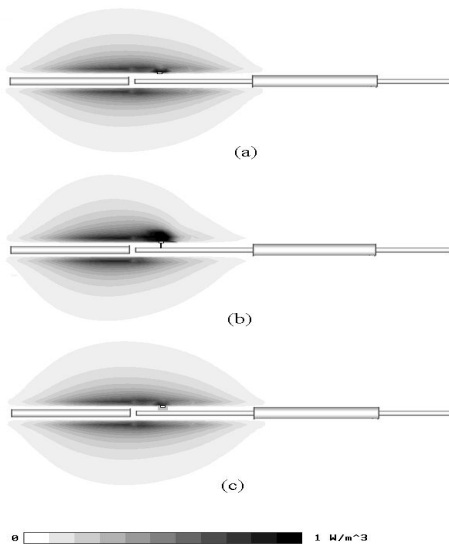

Figure 6: Normalized SAR distribution of a MHA with the integrated temperature probe: (a) floating tip (ideal case), (b) tip short circuited to the coaxial conductor (worst case), (c) tip RF isolated with inductors (actual case).

At a radial distance of 5 mm from the applicator, in the direction of maximum radiation, SAR is less than 50 % of the maximum value calculated at the surface of the catheter and reduces to 90 % at a distance of 10 mm. This confirms that the applicator can be used both for hyperthermic treatments of small tumours ad also for thermo-ablation surgery, depending on the maximum applied power and time. It is worth to note that the device can tolerate 30 W CW or average power and up 150 W pulsed power without any damage or excessive self-heating.

The MHA input matching is also numerically calculated in the frequency range from 2 to 3 GHz, in absence of the thermocouple and in presence of a temperature sensor with the probe close to the applicator and the metallic leads embedded into the coaxial balun. In both cases a reflection coefficient less than -20 dB is assured at the 2.45 GHz working frequency as shown in Figure 7.
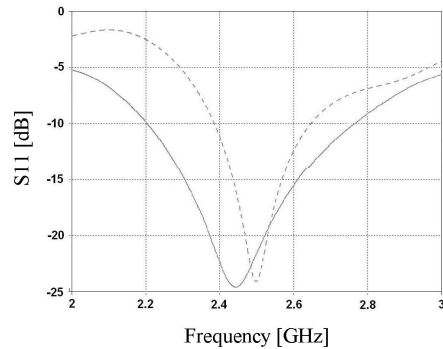

Figure 7: Input reflection parameter vs. frequency of the MHA with (continuous line) and without the integrated temperature probe (dotted line).

Power distribution in the tissue was experimentally evaluated by applying 20 W continuous microwave power to the MHA over a period of about 15 minutes (Figure 8) and a pulse power of 60 W for 10 seconds. The results are in good agreement with the numerical predictions and proved that the induced heating pattern in the biological medium assumes the typical ellipsoidal shape around the applicator radiating end. The overall dimension of the ellipsoidal heating pattern clearly depends on the so called *thermal-dose*, i.e. on the quantity of the EM energy delivered to the medium. It is also evidenced that the metallic sensor tip (thermocouple or thermistor) do not produces a local hot spot, authorizing us to state that an accurate temperature monitoring can be obtained. This was afterward confirmed through a measure made with an auxiliary fiber-optic temperature sensor. Using the integrated

sensor, the temperature at the interface between the catheter and the tissue can be carefully monitored during the microwave heating process because not significant self-heating of the sensor has been observed. Figure 9a shows the evolution of the temperature vs. time when 20 W CW power is applied for 15 minutes at the input of the applicator, while Figure 9b depicts the thermal response of the medium to a 10 second pulse of 60 W peak power.

It is worth note that the increase of the temperature is very fast in both cases because the chicken breast used as phantom for our heating experiments is not perfused by the blood and therefore the thermal response of the tissue is determined only by its thermal conductivity.
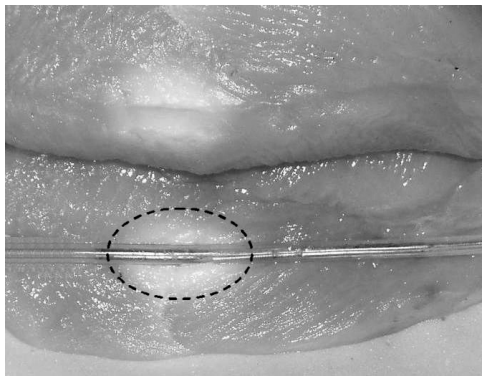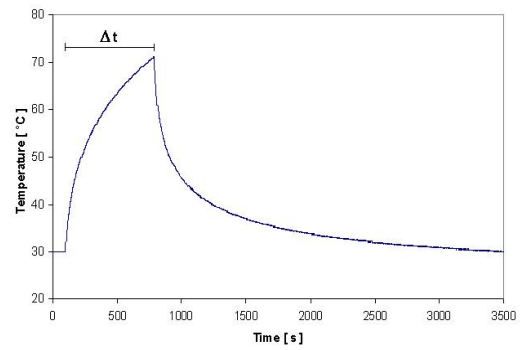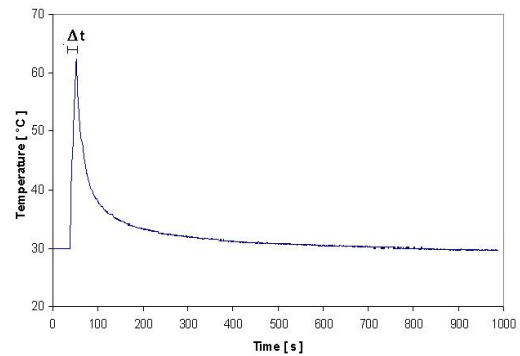

(a)
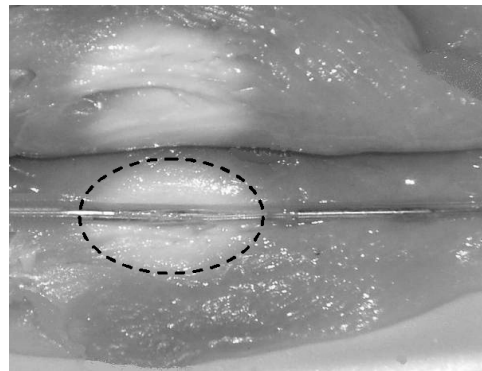
(b)

Figure 8: Thermal pattern in a splitted chicken tissue obtained by applying 20 W CW power to the MHA for 15 minutes (a) and 20 minutes (b) respectively.

(a)

(b)

Figure 9: Temperature evolution at the tissue/applicator interface obtained by applying 20W CW input power to the MHA for 15 minutes (a) and a 10 second pulse of 60W peak power (b).

4 CONCLUSIONS

The proposed minimally invasive MHA integrates a very cheap temperature sensor and therefore it is very suited for the mass-production of mono-use devices. The integration of the radiating element and the temperature sensor inside the same applicator case allows to heat a small tissue volume (target) and to measure the temperature accurately at the same time. Due to the use of a simple coaxial balun the microwave energy is confined around the applicator body reducing the risk of accidental overheating of healthy tissue close to the tumour. The thermistor (or the thermocouple junction) peeps out from the catheter surface in the point where the EM field, and hence the temperature, reaches the maximum. Temperature in deeper zone of the tissue surrounding the applicator can be extrapolated by mathematical models based on the bio-heat equation (Pennes, 1948) if the EM and thermal parameters of the tissue are known.

Multiple applicators (arrays) with integrated temperature sensors could be used in order to treat larger tissue volumes and more accurately estimate the temperature distribution through the combined application of bio-heat equation and tomography algorithms.

The pliability of the miniaturized coaxial cable and of the associated silicon catheter make easier the insertion of the applicator using the natural way of the body or minimally invasive surgical procedures.

Many coaxial applicators for hyperthermic treatments show a heating pattern characterized by a typical tear drop shape. The implementation of a coaxial choke in the MHA investigated in this paper reduces appreciably the drop tail and allows to precisely localize the tissue volume involved during the microwave treatment. Moreover the high degree of miniaturization due to the availability of miniaturized coaxial cable to use in medium-high power applications (dimensions about 1-2 mm in diameter are easily available), permits to extend the clinical applications of this minimally invasive applicator. Hyperthermic treatments of bile-ducts in cancer therapy, impracticable in the past for the restrict dimensions of the ducts, could be possible nowadays as well other delicate surgical interventions that require high precision and reduced invasivity.

The originality of this MHA, from an engineering point of view, lies in the peculiar integration of the metallic wires of a low cost temperature sensor inside of the choke body without perturbing the SAR distribution.

The invasivity of the clinical hyperthermic or thermo-ablation treatment is highly reduced using this type of applicator because in fact no separated insertions for temperature probes, no additional external electrodes as for RF treatments or any other kind of devices are required. Therefore many deep-seated tumors (e.g. certain brain, liver, gastrointestinal or gynaecological tumours), could be effectively and easily treated with the proposed MHA.

The integrated temperature sensing element permits to accurately monitor the maximum temperature reached into the tissue and it can be used to close the control loop of a specific microwave hyperthermic or ablative process by defining the appropriate thermal-dose to be administered to the lesion.

By monitoring the temperature in time domain, very useful data on blood perfusion rate, thermal conductivity and specific heat could be directly extrapolated by the bio-heat transfer equation and used to construct more complex and realistic biological tissue models. As well thermal properties difference between healthy and pathological tissue could be relieved in order to extrapolate diagnostic information.

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