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Temperature increase dependence on ultrasound attenuation coefficient in innovative tissue-mimicking materials

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Abstract

Although high intensity focused ultrasound beams (HIFU) have found rapid agreement in clinical environment as a tool for non invasive surgical ablation and controlled destruction of cancer cells, some aspects related to the interaction of ultrasonic waves with tissues, such as the conversion of acoustic energy into heat, are not thoroughly understood. In this work, innovative tissue-mimicking materials (TMMs), based on Agar and zinc acetate, have been used to conduct investigations in order to determine a relation between the sample attenuation coefficient and its temperature increase measured in the focus region when exposed to an HIFU beam. An empirical relation has been deduced establishing useful basis for further processes of validations of numerical models to be adopted for customizing therapeutic treatments.

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Keywords: Temperature increase; attenuation coefficient; tissue-mimicking materials

1. Introduction

High-intensity focused ultrasound (HIFU) is one of the most investigated technologies in the new landscape of cancer therapy and tumor ablation (ter Haar, 2001). In HIFU therapy, ultrasonic waves emitted by the transducer are focused in a small volume in order to necrose cancerous tissue selectively, leaving intact the surrounding tissue. Despite the multitude of medical applications, the effects of the ultrasonic waves interaction with biological tissues have not been fully described yet. In order to find a relation between the attenuation coefficient, α, of a sample and its temperature increase, ΔT, which verifies in the focus region when subjected to an HIFU beam, TMMs based on

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Agar and zinc acetate have been realized. The absence of scattering agents and, in the same time, acoustic properties suitable to simulate soft tissues make these TMMs ideal samples to be used for the investigation of absorption phenomena of acoustic energy in tissues. Measurements of TMMs temperature increase as a function of HIFU power have been conducted, while their ultrasound attenuation has been determined both as a function of frequency and temperature. Thereafter, an empirical relation between α and ΔT has been deduced.

2. Material and methods

2.1. Tissue- mimicking materials preparation

Measurements have been conducted on two TMMs, hereafter referred to as TMM1 and TMM2, prepared dissolving Agar (2% in weight), a well-known polysaccharide extensively used for the realization of tissue-mimicking materials, in a solution of zinc acetate [$Zn(CH_3COOH)_2$] with a concentration of 0.2 M and 0.4 M respectively. The zinc acetate makes the samples attenuation tuning possible, as observed in previous investigations (Troia et al., 2015). Studied samples have a cylindrical shape with a diameter of about 0.05 m and a thickness of 0.03 m. They have been obtained casting carefully into a mould the solution previously heated under stirring on a hot plate at 100 °C for 5 min.

2.2. Ultrasonic attenuation

The attenuation coefficient, α, of investigated TMMs has been determined using an insertion technique as described in Cuccaro et al. (2015), where details on applied corrections and signal analysis method are reported too. However, in this work, an improved version of the experimental apparatus described in Cuccaro et al. (2015) has been used (see Fig. 1.a) to conduct measurement of α at temperatures different from the ambient one. A temperature control, made of two PT100 thermometers and a heater driven by a PID control, has been added to the previous apparatus in order to vary the temperature, T, between 20 °C and 45 °C with an uncertainty of about 0.07 °C. Moreover, the apparatus has been placed inside an oven with a resolution of 1 °C to speed up the achievement of the desired water temperature.

2.3. Temperature increase induced by HIFU

When a TMM is subjected to an ultrasound beam, a change in temperature and pressure in the focus region occurs. In this work, these quantities have been measured using a fiber optic probe hydrophone (FOPH-2000, RP Acoustics) inserted in the investigated sample. Its functioning is based on the measurement of the refractive index, n, which depends on the acoustic pressure, p, and on the temperature, T, of the sample ($n(p,T)$). A laser (λ= 808 nm) is coupled to the glass fiber and its light is partly transmitted through the sample and partly reflected at the fiber tip. Reflected component is converted into an electrical signal and acquired by a digital oscilloscope (LeCroy Wave Runner 6030A). The HIFU transducer (H-106-MRA, f = 2 MHz), powered by a wave function generator (Agilent 33250A), and the TMM are immersed in de-ionized water (see Fig. 1.b).

3. Results and discussion

The attenuation coefficient of TMM 1 and TMM 2 has been determined as a function of the frequency $(2\ \text{MHz} \le f \le 8\ \text{MHz})$ at six different values of temperature, with T included between 20 °C and 45 °C. In figure 2.a, α values obtained at $T = (20.3 \pm 0.07)$ °C are shown. Fitting experimental results with the curve $\alpha = \alpha_0 \cdot f^n$, generally used to describe the attenuation dependence on frequency in biological tissues, it is possible to highlight the variation of the attenuation coefficient α_0 with the temperature (see Fig. 2.b).

The dependence of α_0 on T can be described by means of a second degree polynomial ($\alpha_0 = a + bT + cT^2$), with the following parameters values: $a = (0.13 \pm 0.01)$ Np·cm^{-1}·Mz^{-n}, $b = (-0.0042 \pm 0.0005)$ Np·cm^{-1}·MHz^{-n}·°C^{-1}, $c = (0.40 \pm 0.06)\,10^{-4}$ Np·cm^{-1}·MHz^{-n}·°C^{-2} for TMM 1; $a = (0.17 \pm 0.02)$ Np·cm^{-1}·Mz^{-n}, $b = (-0.006 \pm 0.001)$ Np·cm^{-1}·MHz^{-n}·°C^{-1}, $c = (0.57 \pm 0.13)\,10^{-4}$ Np·cm^{-1}·MHz^{-n}·°C^{-2} for TMM 2.

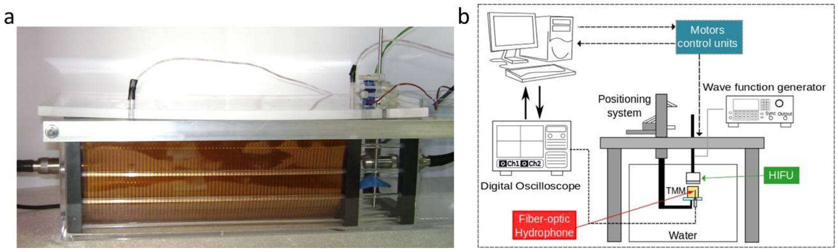

Fig. 1. (a) Experimental apparatus for the measurement of TMMs attenuation coefficient; (b) Experimental set up for the measurement of the temperature increase by means of a fiber optic probe hydrophone.

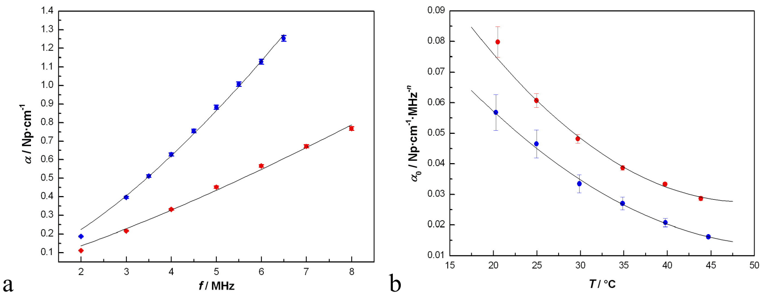

Fig. 2. (a) Attenuation, α, values and respective uncertainty as a function of the frequency, f, at T = (20.3±0.7) °C: ● TMM 1, ● TMM 2, - fitting curve; (b) Attenuation coefficient, α_0, values and respective uncertainty as a function of the temperature, T: ●TMM 1, ●TMM 2, - fitting curve.

After their characterization, TMMs have been exposed to an HIFU beam varying the transducer output power, P, between 10 W and 50 W. For both TMMs, the temperature increase, ΔT, and the acoustic pressure, p, as a function of P are shown in figure 3a and 3b .

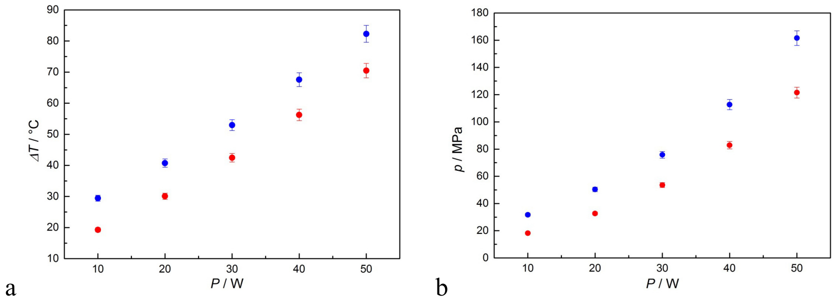

Fig. 3. (a) Temperature rise, ΔT, as a function of ultrasound beam power, P: ● TMM 1, ● TMM 2; (b) Acoustic pressure, p, as a function of ultrasound beam power, P: ● TMM 1, ● TMM 2.

Observing figure 3a and 3b, it is clear that ΔT and p values achieved in the sample during HIFU insonation don't depend only on the beam power but also on the sample properties, such as the attenuation coefficient. Indeed, in an absorbing medium a short burst of focused ultrasounds generates a temperature increase which, for a given configuration and for short impulse durations, Δt , can be described as (Clark et al., 1996):

$$\Delta T(t) = \frac{2\alpha_{\mathrm{abs}} I}{\rho C_p} \Delta t \qquad (1)$$

where ΔT is the temperature rise at the beam focus due to bulk heating, α_{abs} is the absorption coefficient, I is the spatial peak intensity at the focus, ρ and C_p are the density and the heat capacity of the medium respectively. However, equation 1 doesn't succeed in describing our ΔT experimental results. For a better correspondence between experimental and theoretical values, it is necessary to introduce the dependence of the absorption coefficient, that in this work it is assumed to be close to the measured attenuation coefficient because of the absence of scattering agents in investigated samples, on the temperature. Considering $\alpha_{\text{abs}} = a + bT$, equation 1 can be rewritten as:

$$\Delta T = \frac{2(a+bT)I}{\rho C_p}\Delta t \ . \qquad (2)$$

Solving Eq. 2, a theoretical expression for T to be used for fitting experimental values is obtained. In figure 4a and b, experimental values of T as a function of I and the fitting curve are shown for TMM 1 and TMM 2 respectively. Comparing a and b values obtained from the fitting procedure with those ones obtained experimentally, the quality of the theoretical relation between ΔT and α emerges.

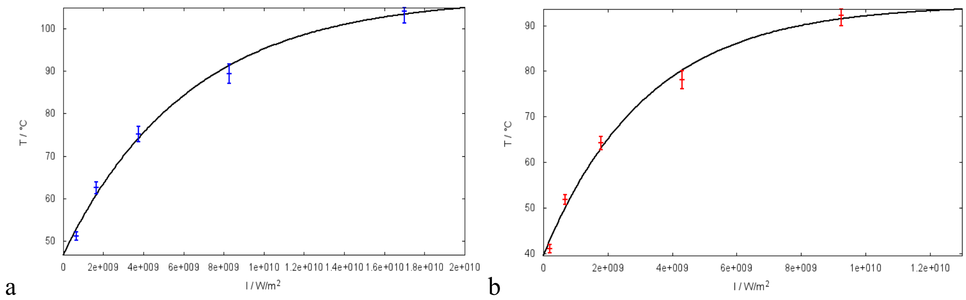

Fig. 4. Temperature as a function of ultrasound intensity: (a) TMM 1 - experimental data (blu) and theoretical curve (black); (b) TMM 2 - experimental data (red) and theoretical curve (black);

4. Conclusions

In this work, the existence of a relation between the ultrasonic attenuation coefficient of a TMM and the temperature increase which occurs in the focus region when it is subjected to an HIFU beam has been highlighted and an equation to predict temperature rise knowing the attenuation dependence on temperature and the focus intensity has been provided, even if it needs still improvements. Nevertheless, the experimental evidences provided by the present work may be of clinical interest for the definition of the HIFU treatment planning to be applied to the patient.

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