Dual Mode Transducer for Ultrasound Monitored Thermal Therapy

Bouchoux Guillaume[*,**], Berriet Rémi [*], Lafon Cyril [**], Fleury Gérard [*], Cathignol Dominique [**], Chapelon Jean-Yves [**]

** Imasonic, 15 rue A. Savary, 25000 Besançon, France*
*** Inserm U556, 151 cours A. Thomas, 69424 Lyon Cedex 03, France*

Abstract. A flat single element transducer able to perform and to monitor interstitial therapy is studied. This transducer must generate the acoustic intensity necessary to induce thermal lesions. It must also meet real-time monitoring requirements. A 3x8 mm² 7.5 MHz composite transducer was built. Acoustic intensities up to 30 W/cm² emitted during more than 20 s with efficiency around 80% were measured. The length of the impulse response at -30 dB was 3 periods (0.3 mm). Insertion losses were found to be around 6 dB. A system interrupting high intensity emission periodically in order to acquire RF echo lines was set up. In-vitro tests on porcine liver were done. M-mode images showing the evolution of the lesion depth during the high intensity emission were obtained. The lesion depth estimated on M-mode images was well correlated with the depth measured on the liver samples after the experiments. In this study, we demonstrated that thermal lesions can be both generated and monitored by a specially-designed flat ultrasound transducer.

Keywords: Ultrasound; Interstitial; Therapy; Monitoring; Transducer.
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INTRODUCTION

Interstitial ultrasound thermal therapy has been shown to be effective for treating certain kinds of tumors in deep-seated organs [1,2]. A small ultrasound transducer is brought in contact with the target. Then a continuous high intensity ultrasound wave is emitted by the transducer so that thermal lesions can be induced in the adjacent tissues. The transducer can be displaced to scan a larger area.

MRI temperature images have been proven to be an effective way to monitor thermal ablations [2, 3]. However, the temporal resolution of this method can be low, and the cost of the treatment is considerably increased. Using ultrasound could accelerate and simplify the monitoring procedure. By nature, interstitial ultrasound therapy can hardly be monitored by an external ultrasound scanner. Combining therapy and monitoring functions on the same transducer could respond to the space limitation. The ability to generate continuously an acoustic intensity greater than 15 W/cm² is required for therapeutic purpose. Good resolution and sensitivity are needed for the monitoring. In this study a flat mono-element transducer able to generate thermal lesions and to monitor the treatment has been designed, built, and tested.

MATERIAL AND METHODS

Design of a Dual Mode Transducer

The monitoring mode requires a fine spatial resolution and a good signal to noise ratio. Therefore, the transducer must have a short impulse response and low insertion losses. The transducer must also be able to generate high intensity acoustic waves during a long duration in order to induce thermal lesions. The maximum acoustic intensity that can be emitted by a transducer is limited by heating due to losses. Losses must be minimized and self-generated heat has to be evacuated in an effective way. A trade off has to be found between high sensitivity and good electro-acoustic efficiency on one side, and the impulse response length on the other side.

The center frequency of the monitoring mode and the therapeutic frequency may be identical in interstitial applications. This frequency is chosen according to the desired lesion and imaging depth, since the attenuation in tissues depends on frequency.

In this study, a 3x8 mm^2 flat single element transducer is considered. Both imaging and monitoring are done at 7.5 MHz.

A basic structure of transducer was chosen. The active part of the transducer is a 1-3 piezocomposite material for low acoustic impedance and high electro-mechanical coupling coefficient compared to pure PZT. Thus a large bandwidth and short impulse response can be obtained. The two components of the composite were chosen with low losses in order to obtain a high transmit. The shape of the impulse response of the transducer is also improved using an acoustic matching layer and an electric matching circuit. A cooling system is also considered in order to evacuate self-generated heat from the transducer during therapy.

The impulse response, insertion loss, and efficiency of the transducer were simulated using a KLM model. This electro-acoustic model is not able to predict the ability of the transducer to generate high intensity acoustic waves. For this purpose, the temperature of the transducer during therapy was modeled by a 1-D thermal model.

The basic structure was optimized in order to approach desired monitoring and therapy characteristics. The designed transducer was fabricated and mounted on a holder (figure 1).

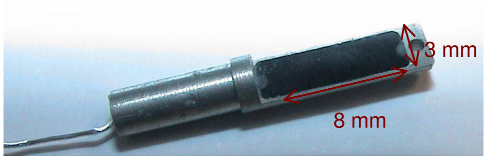

FIGURE 1. A dual mode transducer

Transducer Acoustical Characterization

The impulse response of the transducer was measured. The transducer was excited by short spikes and the echo returning from a flat brass target was amplified and digitized. The insertion losses at the center frequency were measured when exciting the transducer by a 10 periods burst. Insertion losses are the ratio between the amplitude of the received echo and the amplitude of the excitation in a load matched to the generator.

The transmit efficiency of the transducer is the ratio between the electric power delivered to the transducer and the generated acoustic power. Electric power was measured with a wattmeter, and the acoustic power was measured on an absorbing target using the acoustic radiation force balance. A small thermocouple was glued in contact on transducer surface.

In-vitro Experiments

A driving system that interrupts periodically high intensity emission in order to acquire a RF echo lines was set up. The commutation was done by a computer-controlled mechanical relay. A function generator and an amplifier provided the therapeutic excitation. During monitoring cycles, the transducer was excited by a pulse and the response was amplified, digitized and transferred to the memory of the computer. The digitized RF echo lines were then treated by the computer and displayed as an M-mode image. The cycles consist in 30 ms long therapeutic emissions followed by 7 ms long periods for echo line acquisition were chosen. The high intensity emission duty cycle was thus 80 % and 20 echo lines per second were acquired.

Experiments were conducted in-vitro on degassed porcine liver samples at ambient temperature. The sample was placed at a distance of 6 mm from the transducer surface in a degassed water tank. Several acoustic intensities, frequencies and treatment durations were tested. During these experiments the surface of the transducer was cooled by water circulation.

RESULTS

Transducer Acoustical Characterization

The measured impulse response duration was 350 ns at -20 dB and 400 ns at -30 dB, this is equivalent to 0.3 mm (figure 3). Center frequency of impulse response is 7.65 MHz. At this frequency, the insertion losses are 6 dB. The electro-acoustic efficiency of the transducer is above 80% on a frequency band from 6.5 MHz to 8 MHz (figure 4). During a 20 s, 7.5 MHz, 24 W/cm² emission the temperature of the transducer cooled by water circulation increased by 40 °C (figure 5).

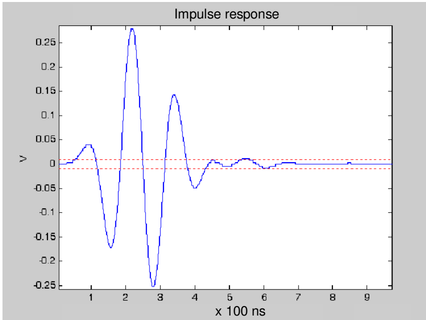

FIGURE 3. Measured impulse response of the transducer.

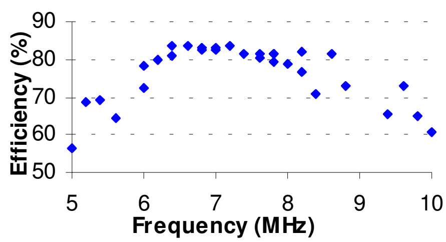

FIGURE 4. Electro-acoustic effiency

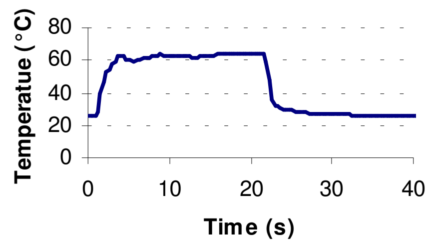

FIGURE 4. Temperature during emission

In-vitro Experiments

Several experiments were done on porcine liver in-vitro, using different exposure conditions. Figure 7 shows typical results. In the first case, no lesion was found on the liver sample after the treatment, and no change occurred on M-mode image. In the second case a small lesion developed. A hyperechoic zone settled after 13 s of exposure on the M-mode image. In the third case, a bigger lesion was formed. The liver sample was highly damaged and torn apart during treatment. On M-mode image, a hyperechoic zone appears after 10 s, and its depth grew slowly until the end of treatment. Table 1 summarizes depths of the thermal lesions measured after the treatments and depths of hyperechoic zones at the end of corresponding M-mode images for several exposure parameters.

	10 W/cm² 7.5 MHz 30 s	16 W/cm² 7.5 MHz 20 s	20 W/cm² 7.5 MHz 30 s	20 W/cm² 7.5 MHz 60 s	30 W/cm² 6.5 MHz 60 s
Hyperechoic depth	0	5	7	11	13
Lesion depth	0	3	6	9	12

TABLE 1. Depth of thermal lesions estimated on M-mode images compared to depth measured on liver samples after treatment for several exposure conditions.

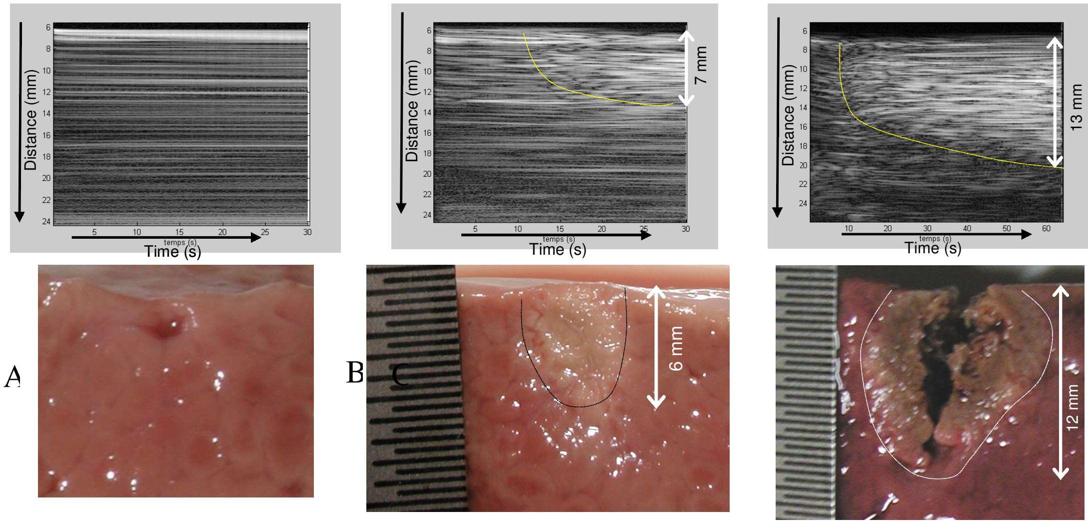

FIGURE 7. M-mode images acquired during treatments and corresponding pictures of the liver samples after treatments for different exposition parameters. A: 10 W/cm², 30 s, 7.5 MHz ; B: 20 W/cm², 30 s, 7.5 MHz ; C: 30 W/cm², 6.5 MHz

CONCLUSION

A flat 3x8 mm² 7.5 MHz dual-mode transducer was built. The monitoring mode offered a good axial resolution and a very good sensitivity. Efficiency was above 80 % over a wide range of frequency and high intensity emissions were done with reasonable heating of the transducer. Monitored thermal therapy was performed in-vitro with this transducer. Lesions up to 12 mm deep were induced. M-mode echo images of the treatment showed hyperechoic zones corresponding in size to actual thermal lesions depth.

REFERENCES

1. Prat F. et al., *Gastrointest Endosc,* **53**(7), 797-800 (2001).
2. Diederich CJ et al., *Med Phys* **31**(2), 405-13 (2004).
3. Hynynen K. et al., *Radiographics* **16**(1), 185-95 (1996).