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Thermoacoustic resonance effect and circuit modelling of biological tissue

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In this letter, thermoacoustic resonance effect is predicted from theoretical analysis with series resistor-inductor-capacitor resonance circuit model and then observed experimentally using muscle tissue illuminated by multi-pulse microwave source. Through model fitting, the circuit parameters are extracted to characterize quantitatively the resonant response of the tissue. Coherent demodulation is applied to obtain the enhanced signal-to-noise ratio and spatial information by treating tissue as a communication channel. This physical phenomenon shows significantly higher sensitivity than conventional single microwave pulse induced thermoacoustic effect, enabling the potential design of low-power thermoacoustic imaging device for portable and on-site diagnosis.

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Microwave-induced thermoacoustic (TA) effect refers to the thermal expansion of tissue leading to the generation of an acoustic transient following the microwave illumination.^{1–7} Based on the TA effect, microwave-induced thermoacoustic tomography (TAT) has been proposed and applied to *ex vivo* and *in vivo* medical imaging, such as breast cancer detection, providing high microwave absorption contrast and high acoustic resolution simultaneously.^{8–11} Existing prototypes of TAT experiments are using single microwave pulse with a high peak power of up to 10 kW and above to illuminate biological tissue.^{12,13} Thereby, a wide-band thermoacoustic signal is induced inside the tissue sample and detected by ultrasound transducers for image reconstruction. A major obstacle for current prototypes is that high peak power microwave sources are bulky and expensive for an envisioned portable medical imaging device for on-site diagnosis.

In this letter we report the resonance response of the tissue induced by multiple pulses of the microwave source. The response of the tissue can be modelled with a damped harmonic oscillator, i.e., a series RLC circuit. Once the multi-pulse microwave source matches the resonance frequency of the biological tissue, the signal-to-noise ratio (SNR) of thermoacoustic resonance (TAR) signal is enhanced significantly, and narrowband characteristic is also achieved. Additionally, we propose a coherent demodulation scheme to obtain depth-resolved information for a future imaging implementation of the technique.

The basics of microwave-induced thermoacoustic effect is well established in the literature.^{1,2} Very similar to the photoacoustic effect induced from pulsed lasers,^{14–20} biological tissue illuminated by pulsed microwave will absorb some of the electromagnetic (EM) energy. This results in heating and thermal expansion of the tissue, as sketched in Fig. 1(a). Under the constrain that the tissue absorber (here about 5 mm in diameter) is small compared to the microwave wavelength, it acts as a point source, and the acoustic pressure emitted, $p(t)$, can be described through a forced oscillator^{21,22}

$$\frac{\partial^2 p(t)}{\partial t^2} + a^2 \frac{\xi + \frac{4}{3}\eta}{\rho} \frac{\partial p(t)}{\partial t} + a^2 c^2 p(t) = \Gamma \frac{\partial H(t)}{\partial t}, \quad (1)$$

where a is the propagation phase constant, ρ is the tissue density, η is the shear viscosity, ξ is the bulk viscosity, and c is the acoustic velocity in tissue. Γ is the Gruneisen constant expressed as $\Gamma = \beta c^2 / c_p$, where β is the thermal expansion coefficient and c_p is the constant pressure heat capacity per unit mass. $H(t) = \sigma \bar{E}^2(t)$ is the heating function of microwave illumination, where σ is the conductivity of tissue, and $\bar{E}(t)$ is the short-time averaged electrical field strength.

In Eq. (1), the acoustic pressure follows a second order differential equation with source term $\Gamma \frac{\partial H(t)}{\partial t}$, which is electrically analogue to a series RLC circuit,²³ where the

FIG. 1. (a) Diagram of thermoacoustic effect induced by microwave illumination, and (b) its equivalent series RLC resonance circuit model driven by voltage source $v(t)$.

pressure, $p(t)$, is replaced with a time-dependent current $i(t)$, and the heat source $H(t)$ with voltage source $v(t)$ [see Fig. 1(b)].

$$\frac{\partial^2}{\partial t^2} i(t) + \frac{R}{L} \frac{\partial}{\partial t} i(t) + \frac{1}{LC} i(t) = \frac{1}{L} \frac{\partial}{\partial t} v(t), \quad (2)$$

where the resistance R , inductance L , and capacitance C represent the tissue properties of Eq. (1). Next, we discuss the electrical analogue and extraction from the measurement for the parameters. Assuming that the voltage source is a cosine tone burst driven at a frequency ω_R , the response of $i(t)$ at steady state can be expressed as

$$i(t) = |I_p| \cos(\omega_R t + \theta_p), \quad (3)$$

where θ_p is the phase shift and I_p is the complex transient current expressed as

$$I_p = \frac{V}{R + j\omega L + \frac{1}{j\omega C}} = \frac{V}{R + j\left(\omega L - \frac{1}{\omega C}\right)}. \quad (4)$$

The magnitude of I_p will be maximized at the resonance frequency ω_0

$$\omega_R = \omega_0 = \frac{1}{\sqrt{LC}}. \quad (5)$$

We can relate the bandwidth (BW) and the quality factor (Q) with L , R , and C as

$$BW = \frac{R}{L}, \quad (6)$$

$$Q = \frac{\omega_0}{BW} = \frac{\omega_0 L}{R} = \frac{1}{R} \sqrt{\frac{L}{C}}. \quad (7)$$

Thus we expect that the thermoacoustic response of the tissue at steady state will follow the envelope frequency of the microwave source, and once these envelope hits the resonance frequency a resonant response should be observed. This resonance frequency of the tissue is related to the electrical analogue: $\omega_{\text{tissue}} = \omega_0 = 1/\sqrt{LC} = ac$. We now demonstrate that this resonance is sufficiently strong to be observed experimentally.

FIG. 2. Diagram of the experimental setup for the observation of both TA and TAR effect induced by single pulse and multi-pulse microwave source.

The experimental setup which allows to compare the TA and TAR effect is shown in Fig. 2. A microwave generator (SMBV100A, Rohde & Schwarz) generates single and multi-pulse microwave source for the observation of TA and TAR effects, respectively. For the latter, the amplitude shift keying configuration is used together with a serial digital input. The microwave source is amplified up to 100 W with a microwave power amplifier (ZHL-100 W-GAN+, Mini-Circuits). Considering the tissue absorption characteristics with respect to microwave frequencies,²⁴ we choose 440 MHz as the carrier frequency. The pulsed microwave signal is then fed into a custom-designed helical antenna operating at 440 MHz. The thermoacoustic response of the tissue is detected with an ultrasound transducer (V323-SU, Olympus), fed into a preamplifier (54 dB gain, Model 5662, Olympus), and recorded with a digital oscilloscope (WaveMaster 8000A, Lecroy) at 500 MHz sampling rate. Later, they are transferred into a computer for further signal processing. In the tank, both the helical antenna and the ultrasound transducer are immersed in mineral oil ($\epsilon_r = 2.1$, $\sigma \approx 0$). Muscle tissue samples ($\epsilon_r = 56.8$, $\sigma \approx 0.8$) are shaped into small balls of 5 mm diameter and placed close to the end of the helical antenna.

Applying the same peak power source for every pulse in the experiment, the thermoacoustic signals are detected by both the single pulse and the multi-pulse microwave

(a)

(b)

FIG. 3. (a) Incident multi-pulse and single pulse microwave waveform in time domain; (b) induced acoustic signals (dashed box) after band-pass filtering by TAR and TA effects with regards to the incident microwave source.

illumination. Six continuous pulses (TAR) are transmitted for multi-pulse illumination, followed by a single pulse illumination (TA) afterwards, as shown in Fig. 3(a). All pulses have a pulse widths of $3.67 \mu\text{s}$. Fig. 3(b) depicts the acoustic signal after suitable band-pass filters: the two dashed boxes indicate TAR signal and TA signal, respectively. It is seen that the TAR signal is clearly discernible from the background due to its higher SNR and narrowband characteristic. In contrast, the TA signal from a single pulse microwave is almost immersed in the noisy background. Here, even time-of-flight estimation ($\sim 30 \mu\text{s}$ delay between transmit and receive) cannot allow to detect the signal. The SNR for both TAR and TA signals are calculated within two dashed boxes ($30\text{--}60 \mu\text{s}$, $105\text{--}115 \mu\text{s}$) in Fig. 3(b) by

$$\text{SNR} = 10 \log \left(\frac{P_{\text{signal}}}{P_{\text{noise}}} \right). \quad (8)$$

The SNR of the TAR signal is about 11.28 dB, and that of the TA signal is only 2.51 dB, resulting in almost 9 dB SNR improvement. Thus, a sufficient high SNR does not require a high peak power microwave source but can be obtained from the TAR effect. Increasing the number of cycles does not necessarily increase the SNR as energy dissipation in the biological tissue finally limits the TAR signal. This is equivalent to the resistance R in the series RLC circuit.

We obtain the constants R , L , C , ω_0 , and Q_{circuit} from fitting the measured response to a simulation program with integrated circuit emphasis (SPICE) model. These values are

FIG. 4. Model fitting of (a) microwave source in experiment (blue) and voltage source in circuit simulation (red); (b) induced acoustic signal in experiment (green) and current in circuit simulation (red).

$R = 0.5 \Omega$, $L = 0.8 \mu\text{H}$, $C = 0.45 \mu\text{F}$, $\omega_0 = \sqrt{1/LC} = 1.67 \times 10^6 \text{ rad/s}$, and $Q_{\text{circuit}} = (1/R)\sqrt{L/C} = 2.67$. The waveform to drive the resonant system is shown in Fig. 4(a) together with the response in Fig. 4(b). We find a good overall agreement between simulation and experiment. Interestingly, the TAR effect may allow to characterize tissue by the parameters ω_0 and Q_{circuit} .

The spectrum of the TAR signal, TA signal, and noise are obtained by Fourier transformation within the dashed boxes (see Fig. 5). The central frequency of the TAR and TA signals are 0.27 and 0.21 MHz, respectively. As predicted, the TAR signal shows narrowband characteristic determined by multi-pulse illumination rather than the wideband TA signal induced by single pulse microwave illumination, allowing narrowband filtering and detection, which could sufficiently suppress the background noise because the noise amplitude is proportional to the square root of the signal bandwidth. Narrowband TAR signal also relaxes the requirements of choosing ultrasound transducer and pre-amplifier to achieve higher sensitivity even without using high peak power illumination source.

To further enhance the sensitivity for TAR signal detection and extract spatial information of tissue, the RLC resonance model of biological tissue is treated as a communication channel and analyzed with scheme of coherent demodulation. Fig. 6 depicts that the reference signal obtained after an envelope detector and differentiator, which is then cross-correlated with the detected TAR signal to obtain the delay time of the acoustic source. The analysis here indicates a distance of about 50 mm between biological tissue and the ultrasound transducer.

In summary, we have demonstrated thermoacoustic resonance (TAR) and evaluate its parameters through an electrical analogue (RLC circuit). The response from multi-pulse microwave illumination of muscle tissue allows to determine ω_0 and Q of circuit model. The TAR response has higher SNR and narrower bandwidth than the conventional TA signal. The model parameters allow to characterize the tissue, and imaging may be feasible with the presented cross correlation scheme. The TAR effect provides a unique prospect to develop a microwave-induced TAT prototype for low-cost portable devices operating at considerably lower peak power than current implementations.

FIG. 5. Spectrum of TAR signal (solid blue line), TA signal (dotted green line), and noise (solid dotted blue line).

FIG. 6. Proposed coherent demodulation scheme by treating biological tissue as a RLC communication channel to extract the depth-resolved information.

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