Simulation of a Piezoelectric Catheter-based Acoustic Ablation Device

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Introduction

Thermal ablation driven by an acoustic heat source has tremendous potential for the treatment of diseased tissue. In many applications, such as treating tumors in the liver, bulk removal of tissue is required, and thermal ablation can reliably cause necrosis for a relatively large tissue volume by inducing coagulative necrosis [1]. In current practice, the most common thermal ablation technique is radiofrequency (RF) ablation, which uses electromagnetic radiation in the 400-700 kHz range [2]. Compared to RF ablation, acoustic ablation can be used to target tissue more selectively while also being minimally invasive [3].

Currently, high-intensity focused ultrasound (HIFU) is a widely used therapeutic technique used to ablate tissue using an external transducer that focuses acoustic energy into a small volume of tissue [3]. By using hundreds of overlapping focal zones, the tumor volume can be ablated as shown in Figure 1 below [4]. However, in areas where there is substantial motion (such as from breathing) or regions with bones or gaseous pockets, extra-corporeal HIFU is not suitable due to decreases in pressure gain caused by attenuation and phase aberration. Interstitial applicators allow for more control over the ablation and minimize the effects of attenuation and phase aberration [5].

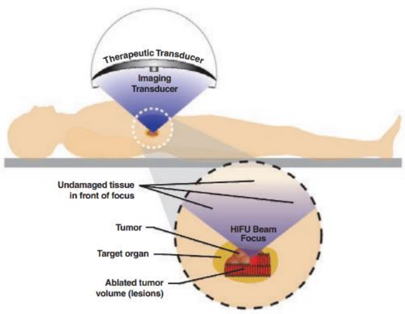

Figure 1. High-intensity focused ultrasound (HIFU) [3]

Catheter-based ablation devices consisting of linear arrays of independently powered tubular transducers offer precise control of the heating pattern, allowing for the spatial power deposition to be altered both along the length of the applicator and in the axial direction for more accurate targeting. In comparison with RF ablation devices, catheter-based applicators using arrays of acoustic transducers have been shown to offer more accurate three-dimensional spatial control of heating and energy deposition and enhanced penetration of energy [1]. Catheter-based acoustic ablation devices also have the added advantages of applying a thermal dose quickly, limiting the diffusion of heat in healthy tissue, and imaging of the tissue can be done using the ultrasound transducers on the applicator [5].

Catheter-based acoustic ablation devices have great potential to be used for the treatment of cancer in regions that are surgically inoperable. In addition to clinical data showing the efficacy of thermal ablation using a catheter-based acoustic ablation device, simulations modeling acoustic ablation have been performed by using numerical solutions of the Pennes bioheat-transfer equation [1].

Finite Element Model

To understand the interaction between acoustics and heat transfer necessary for an acoustic ablation device, Veryst used COMSOL Multiphysics® to simulate the propagation of acoustic waves through liver tissue. Veryst simulated a catheter ablation system with an array of piezoelectric acoustic transducers based on a Boston Scientific patent [6]. We modeled the ablation device and tissue in COMSOL using a 2D-axisymmetric geometry.

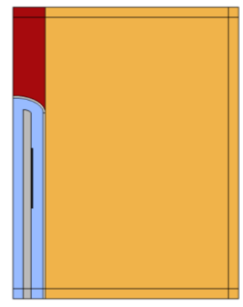

Figure 2. 2D geometry of ablation device, surrounding tissue (yellow), and blood (red)

The ablation device consists of a rigid stainless-steel shaft with an array of twenty-four annular PZT (lead zirconate titanate) transducers held to the shaft by an acoustic insulating epoxy, as seen in Figure 3. The ablation device was encased in a polycarbonate bag filled with aqueous cooling fluid.

Figure 3. 3D geometry of ablation device.

Laminar Flow

We designed the catheter system supporting the piezoelectric transducers with a hollow central shaft to allow for a cooling fluid (25˚C) to circulate through the device, keeping the temperatures of the ablation device from reaching the ablation temperate desired in the tissue. We used a laminar flow model to simulate the circulation of the cooling fluid through the catheter system.

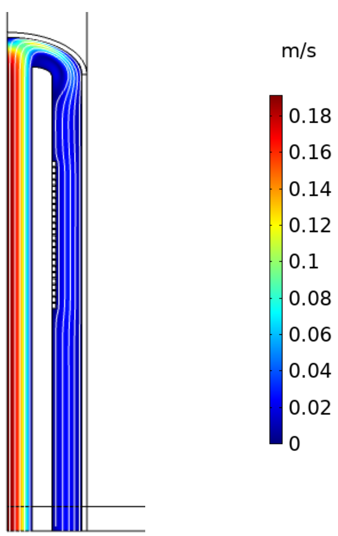

Figure 4. Fluid velocity and streamlines in a cross section of the ablation device

Acoustics

We modeled the acoustic pressure field from the ablation device using the power generated by a voltage difference of 2.75 V on the piezoelectric transducers. By first running an eigenfrequency study to identify the frequencies with the desired mode shape of the transducers (displacement in the radial direction), we were able to identify that a frequency of 5.27 MHz would generate the most effective pressure waveform for ablation with our geometry. We used a perfectly matched layer (PML) to resolve the waves that were not completely attenuated by the tissue and blood. The acoustic attenuation coefficient for each material was evaluated based on the frequency using parameters given in Ref. [7].

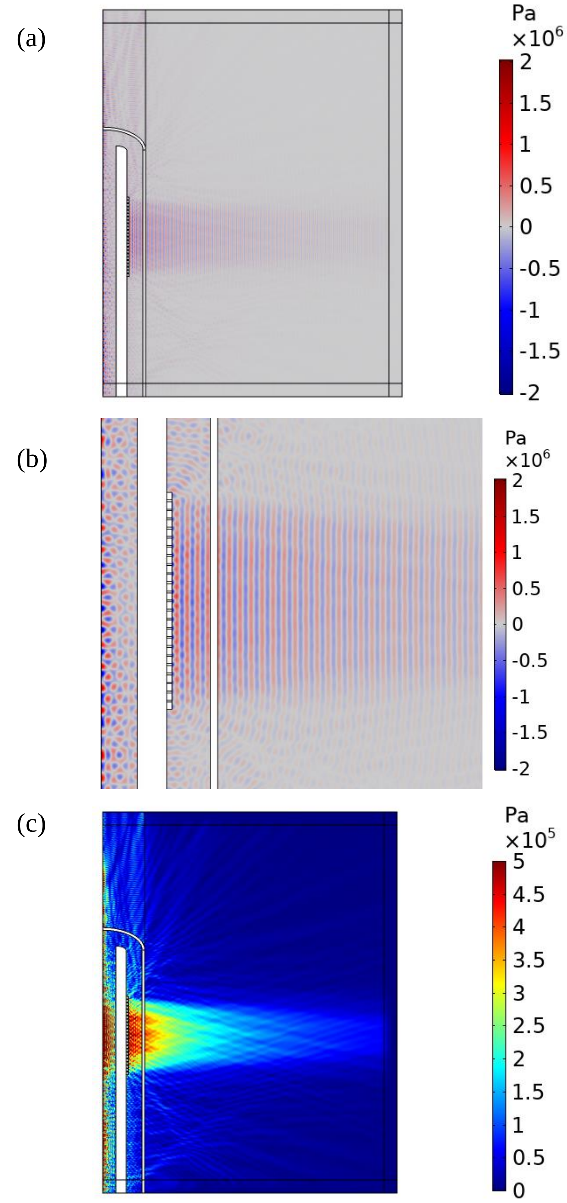

Figure 5. Plots showing (a) acoustic pressure (b) closer view of acoustic pressure near transducer array (c) absolute acoustic pressure

Bioheat Transfer Theoretical Model

The heat transfer physics in the model was solved using the Pennes bioheat equation, with an added term for perfusion losses as shown below. In Equation 1, ρ is the tissue density (1.050 g/cm^3) [8], c_{pt} is the specific heat capacity for tissue (3540 J/kg/K) [9], T' is the temperature rise over blood temperature (°C), k_t is the thermal conductivity of tissue (0.469 W/m/K) [9], ω_b is the blood perfusion rate (6.4e-3 1/s) [10], c_{pb} is the specific heat capacity for blood (3594 J/kg/K) [11], and Q_{ac} is the total generated heat from the acoustic energy deposition of the transducers.

The total heat generated from the transducers can be approximated as the sum of the contributions from each individual transducer [10]. We used frequency-dependent acoustic absorption coefficients from Ref. [7] (0.694 1/m for water at 5.27 MHz, 54.5 1/m for tissue at 5.27 MHz) for the acoustic study in COMSOL.

$$\rho c_{pt} \frac{\partial T'}{\partial t} = \nabla[k_t \nabla T'] - \omega_b c_{pb} T' + Q_{ac} \quad (1)$$

Bioheat Transfer Simulation

In order to simulate heat transfer in the tissue, we used the plane wave power deposition density generated by the acoustic study as a heat source in the bioheat transfer study.

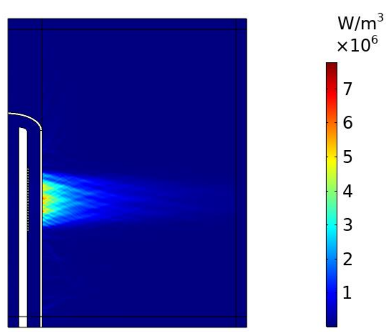

Figure 6. Heat source from acoustic pressure

We used a step function to turn off the heat source after 6 minutes of heating, simulating allowing the tissue to cool after the ablation device is turned off.

We modeled the boundaries of the tissue and blood using a heat flux boundary condition where the external temperature was body temperature (37˚C) and where we estimated the heat transfer coefficient h to be 50 W/m^2/K.

We set the initial temperature in the ablation device to 25°C and the initial temperature in the cooling fluid to 25°C. The temperature profile of the tissue after the bioheat transfer study is shown below in Figure 10.

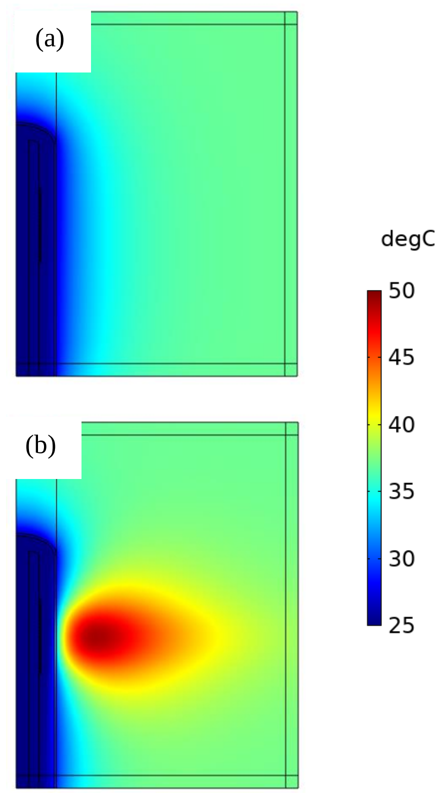

Figure 10. Temperature distribution in ablation device, tissue, and blood (a) at initial state before the start of ablation (b) after 5 minutes of heating

The acoustic heat source raised the temperature in the tissue to a maximum value of 49°C, taking into account blood perfusion.

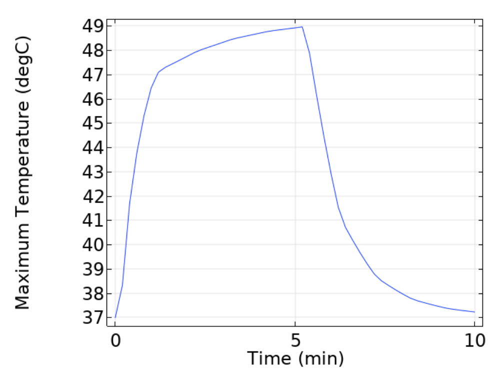

Figure 11. Maximum temperature in the tissue over the ablation period

Resulting Damage

We evaluated the fraction of damaged tissue (with a value of 1 indicating necrosis) using an Arrhenius kinetics damage integral, with values of activation energy and frequency factor for liver tissue from Ref. [9]. Compared to the original design (Figure 12a), the damage varied based on factors such as the distribution of power in the transducers (Figure 12b and 12d) and the temperature of the cooling fluid (Figure 12c).

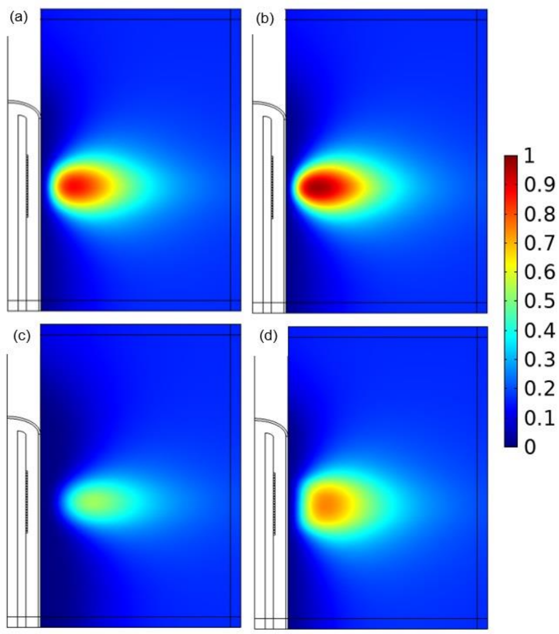

Figure 12. Fraction of damaged tissue from Arrhenius kinetics damage integral after 10 minutes (6 minutes of heating and 4 with the power source turned off) in (a) original model with uniform power distribution along transducer array (b) model with transducer power maximized in central transducers of the array (c) model with cooling fluid at a temperature of 10°C (d) model with transducer power maximized in upper and lower regions of the transducer array

Comparing the model with transducer power maximized in central transducers of the array (b), we can see that the ablation zone is narrower along the length of the applicator and that the damage is more focused into this narrower region.

The model with cooler fluid at 10˚C (c) shows lower damage due to lower temperatures being sustained, but the ablation zone is also shifted radially further from the ablation device.

Changing the power distribution to have the transducer power maximized in upper and lower regions of the transducer array (d) creates a more even temperature distribution along the length of the applicator widens the ablation region. The maximum temperature is lower than in the original model due to the widened ablation region, but the total power is still the same.

Summary

Understanding and modeling the thermal damage that can be achieved by heating a tissue using an acoustic heat source allows us to simulate the design of an interstitial ablation device to be used for the treatment of tumors which cannot otherwise be treated. Using COMSOL Multiphysics®, we simulated a catheter-based acoustic ablation applicator with an array of piezoelectric acoustic transducers and calculated the bioheat transfer in liver tissue from acoustic heat source due to pressure losses in the tissue. Determining the temperature as well as the shape of the ablation zone are important considerations for designing an ablation device, and changing the power distribution of the linear array of transducers can produce accurately targeted ablation zones.

Overall, this model showed how simulation tools can allow for the rapid iteration of various geometries and parameters and provide a great starting point for the development of complex medical devices.

References

[1] Prakash P, Diederich, CJ. Considerations for theoretical modeling of thermal ablation with catheter-based ultrasonic sources: implications for treatment planning, monitoring and control. Int J Hyperthermia. 2012; 28(1): 69-86. [Pubmed: 22235787]

[2] Mast TD, Makin IR, Faidi W, Runk MM, Barthe PG, Slayton MH. Bulk ablation of soft tissue with intense ultrasound: Modeling and experiments. J Acoust Soc Am. 2005 Oct; 118(4): 2715-24. [Pubmed: 16266191]

[3] Dubinsky TJ, Cuevas C, Dighe MK, Kolokythas O, Hwang JH. High-Intensity Focused Ultrasound: Current Potential and Oncologic Applications. AJR Am J Roentgenol. 2008 Jan; 190(1): 191-9. [Pubmed: 18094311]

[4] Brace C. Thermal Tumor Ablation in Clinical Use. IEEE Pulse. 2011 Sept-Oct; 2(5): 28-38. DOI: 10.1109/MPUL.2011.942603

[5] Lafon C, Melodelima D, Salomir R, Chapelon YJ. Interstitial devices for minimally invasive thermal ablation by high-intensity ultrasound. Int J Hyperthermia. 2007; 23(2): 153-163, DOI: 10.1080/02656730601173029

[6] Crowley RJ. 1995. *U.S. Patent No. 5,630,837*

[7] Duck FA, Propagation of sound through tissue, in *The Safe Use of Ultrasound in Medical Diagnosis*, ter Haar G and Duck FA, editors, British Institute of Radiology, London, 2000, pp. 4 – 15.

[8] International commission on radiological protection, Adult reference computational phantoms, ICRP Publication 110, Ann. ICRP 39 (2), 2009 pp.48-51 Table A.1

[9] Hasgall PA, Di Gennaro F, Baumgartner C, Neufeld E, Lloyd B, Gosselin MC, Payne D, Klingenböck A, Kuster N. “IT’IS Database for thermal and electromagnetic parameters of biological tissues,” Version 4.0, May 15, 2018, DOI: 10.13099/VIP21000-04-0.

[10] Valvano, JW, JR Cochran and KR Diller. Thermal conductivity and diffusivity of biomaterials measured with self-heated thermistors. Int. J. Thermophys. 1985: 6: 301-311. DOI: 10.1007/BF00522151

[11] Blake AS, Petley GW, Deakin CD. Effects of changes in packed cell volume on the specific heat capacity of blood: implications for studies measuring heat exchange in extracorporeal circuits. Br J Anaesth. 2000 Jan; 84(1): 28-32. DOI: 10.1093/oxfordjournals.bja.a013376

[12] Salgaonkar VA, Scott S; Kurhanewicz J, Diederich CJ. High-intensity interstitial ultrasound for thermal ablation of focal cancer targets in prostate. AIP Conference Proceedings 1821, 160002. 2017. DOI: 10.1063/1.4977655