



Simulation study on acoustic streaming and convective cooling in blood vessels during a high-intensity focused ultrasound thermal ablation

Maxim A. Solovchuk^a, Tony W.H. Sheu^{a,b,*}, Win-Li Lin^c, Ihyuan Kuo^d, Marc Thiriet^e

^a Department of Engineering Science and Ocean Engineering, National Taiwan University, No. 1, Section 4, Roosevelt Road, Taipei 10617, Taiwan, ROC

^b Taida Institute of Mathematical Science (TIMS), National Taiwan University, Taiwan, ROC

^c Institute of Biomedical Engineering, National Taiwan University, Taiwan, ROC

^d Medical Engineering Research Division, National Health Research Institute, Miaoli, Taiwan, ROC

^e LJLL, University of Paris # 6, Paris, France

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ABSTRACT

This study investigates the influence of blood vessels on temperature distribution during high-intensity focused ultrasound (HIFU) ablation of liver tumors. A three-dimensional acoustics-thermal-fluid coupling model is simulated to compute the temperature field in the hepatic cancerous region. The model is based on the linear Westervelt and bioheat equations as well as the non-linear Navier–Stokes equations for the liver parenchyma and blood vessels. The effect of acoustic streaming is also taken into account in the present HIFU simulation study. We found from this three-dimensional coupling study that in large blood vessel both the convective cooling and acoustic streaming can significantly change the temperature field and thermal lesion near blood vessels.

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1. Introduction

Human liver is a highly perfused organ, which has functions to secrete bile, store glycogen, distribute nutrients from the blood and gastrointestinal tract [1]. In addition, it eliminates endogenous/exogenous substrates and toxins. This physiologically complex and important organ is unfortunately susceptible to the primary and metastatic malignant proliferations. Hepatocellular carcinoma (HCC) is rapidly becoming the most common malignancy worldwide. The survival rate of surgical resection, while being considered as a real hope for the treatment, is only 25–30% in 5 years [2]. A high risk of postoperative recurrence is often reported for multi-focal malignancies. After repeated resections a poor success rate is expected.

HIFU therapy has been applied to ablate solid tumors in different areas of our body, including the pancreas, liver, prostate, breast, uterine fibroids, and soft-tissue sarcomas [2,3]. In comparison with the conventional tumor/cancer treatment modalities, such as the open surgery, radio- and chemo-therapy, HIFU outperforms them because of its non-invasion, non-ionization, and

comparatively fewer complications after treatment. However, lethal complications may develop if the vital blood vessels adjacent to the tumors are severely damaged. Large blood vessels are probably less vulnerable to HIFU damage than the solid tissues (such as tumor tissues) due to energy dissipation in the blood flow. Therefore, HIFU is a relatively safe alternative to ablate the tumors in close proximity to major blood vessels, where surgical resection may be hazardous.

Clinical trials have been recently conducted to evaluate the safety and effectiveness of applying the high-intensity focused ultrasound treatment of the hepatocarcinoma (HCC) [4,5]. When the liver tumor is very close to large blood vessels, surgical treatment becomes complicated. Quite recently [6] it was first shown that HIFU can safely achieve a virtually complete necrosis of tumors close to major blood vessels. After a single session of HIFU treatment, the rate of complete necrosis was about 50%. While substantial, this rate of necrosis following the HIFU ablation is not completely satisfactory. Lack of a complete response can be explained as a consequence of large tumor size and cooling effects in large vessels.

Cooling from a large blood vessel may alter the treatment efficiency. A basic understanding of the factors that can affect the tissue necrosis volume is necessary to improve the thermoablative therapy. In the past, temperature elevation in soft tissues was mostly modeled by the diffusion-type Pennes bioheat transfer equation, which has the heat source produced by the incident

* Corresponding author at: Department of Engineering Science and Ocean Engineering, National Taiwan University, No. 1, Section 4, Roosevelt Road, Taipei 10617, Taiwan, ROC. Fax: +886 2 23929885.

E-mail addresses: solovchuk@gmail.com (M.A. Solovchuk), twshsheu@ntu.edu.tw (T.W.H. Sheu).

Nomenclature

c_0	speed of ultrasound in tissue, m/s
c	specific heat, J/kg °C
\mathbf{F}	force vector per unit volume, N/m ³
k	wave number
k_t	thermal conductivity of tissue, W/m °C
I	sound intensity, W/m ²
p	acoustic pressure, N/m ²
\mathbf{P}	fluid static pressure, N/m ²
q	ultrasound power deposition, W/m ³
t	time, s
t_0	initial time, s
t_{final}	final time, s
T	temperature, °C
u	blood flow velocity, m/s
w	velocity in z direction, m/s
w_b	blood perfusion rate, kg/m ³ s
x	coordinate in the x direction

y	coordinate in the y direction
z	coordinate in the z direction

Greek symbols

α	absorption coefficient, Np/MHz m
β	non-linearity coefficient
δ	acoustic diffusivity
λ	wavelength, m
μ	shear viscosity of blood flow, kg/m s
ρ	density, kg/m ³
Ψ	acoustic velocity potential
ω	angular frequency, MHz

Subscripts

t	tissue
b	blood

acoustic wave and heat sink owing to the perfusion in capillaries [7]. The amount of removed heat can be estimated by averaging the effect of blood perfusion over all tissues. Homogenization assumption is probably no longer valid in modeling the temperature elevation in the regions containing some sufficiently large vessels, inside which the blood is flowing.

Both biologically relevant convective cooling in large blood vessels and perfusion cooling in microvasculatures need to be taken into account altogether. Inclusion of these two possible cooling means will greatly increase the modeling complexity since the equations of motion for the blood flow need to be solved together with the divergence-free velocity constraint equation. Curra et al. [8] developed a model to determine the influence of blood flow on the temperature distribution in the tissue during the focused ultrasound surgery. Their model was constructed on the basis of acoustic and bioheat equations. The blood vessel was on the acoustic axis. Kolios et al. [9] and Hariharan et al. [10] studied the influence of large blood vessels on the lesion size. They considered also only the acoustic equation and bioheat equation. However for a real blood vessel it is necessary to take into account the blood flow motion governed by the non-linear hemodynamic equations. The 3D mathematical acoustics-thermal-fluid model presented in [11] takes the additional acoustic streaming effect into account with an aim to show that the incident finite-amplitude ultrasound wave can affect the blood flow motion in large hepatic vessels and consequently the temperature distribution in tumor. In [11] the numerical simulation was carried out in a patient specific liver geometry. The distance between the focal point and blood vessel was about 1 cm. It was shown that acoustic streaming can affect the blood flow distribution in hepatic arterial branches and lead to a mass flux redistribution. In the present work the simple geometry is considered to understand the effect of acoustic streaming on the temperature distribution. The blood vessel is parallel to the acoustic axis. So far no numerical result, which is computed from the heat transfer equation for large blood vessel and its surrounding tissue coupled with the non-linear hemodynamic equations with an acoustic streaming effect being considered, is ever reported for HIFU tumor ablation. The whole physics remains nowadays poorly understood.

2. Three-field coupling model

Intense ultrasound energy can be delivered to a small region of the targeted tissue. Absorption of energy of this sort in such area

can, as a result, elevate tissue temperature to a certain high magnitude. Such a short-duration temperature increase can destroy tumoral cells with a high dividing rate, thus becoming more sensitive. For the sake of quantifying tissue response to an applied ultrasound, the biological structure in the region under current investigation needs to be taken into account.

2.1. Acoustic equation for ultrasound propagation

The linear Westervelt wave equation [12] given below for ultrasound pressure p will be employed to model the finite-amplitude wave propagation in a soft tissue:

$$\nabla^2 p - \frac{1}{c_0^2} \frac{\partial^2 p}{\partial t^2} + \frac{\delta}{c_0^4} \frac{\partial^3 p}{\partial t^3} + \frac{\beta}{\rho_0 c_0^4} \frac{\partial^2 p^2}{\partial t^2} = 0 \quad (1)$$

The first two terms govern the linear lossless wave propagating at a small-signal sound speed. The third term needs to be included in the current analysis due to thermal conductivity and fluid viscosity. In soft tissues, which are assumed to be thermoviscous, the acoustic diffusivity δ accounts for the thermal and viscous losses in a fluid and is modeled by

$$\delta = \frac{2c_0^3 \alpha}{\omega^2} \quad (2)$$

In the above, α denotes the acoustic absorption coefficient and $\omega (\equiv 2\pi f)$ is the angular frequency. In the current HIFU numerical study, we neglect the non-linear effect, which will be the focus of our future study, for simplifying the analysis.

In this study each small element dS of the transducer surface is assumed to vibrate continuously at the same velocity $u = u_0 \exp(i\omega t)$ along the direction normal to the surface. The resulting linear wave equation can be transformed to the diffraction integral equation for the velocity potential as follows [13,14]

$$\psi_{\bar{p}} = \int \int_S \frac{u}{2\pi r} \exp^{-(\alpha+ik)r} dS \quad (3)$$

In the above, r is the distance from the source point on the transducer surface dS to a field point \bar{p} , and k is the wave number. The pressure amplitude at point \bar{p} can be calculated from the following expression

$$p_{\bar{p}} = ikc\rho_t\psi \quad (4)$$

where ρ_t is the density of tissue, and c is the speed of ultrasound in tissues.

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