A Change on Backscattered Energy (CBE) simulation model to analyze the influence of heating in ultrasound data

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Abstract: Thermal variations influence media properties and consequently induce changes on the propagation of ultrasound (US) waves. Two main questions may arise: Is it possible to estimate temperature based on the variations from the measured US signals? Is it possible to obtain new information (images) from the analysis of the US signals variations? We created two simulation models that enable the assessment of how temperature variations influence the changes on backscattered energy (CBE). Two heating scenarios were considered: slow temperature variations by water-bath heating and heating induced by therapeutic US. The water-bath simulation mimics a real experiment. The therapeutic ultrasound model simulates the heating originated by a piston-like transducer.

CBE simulations were performed in 2D and presented the same patterns as gray-level variations observed on images collected from the real experiment. This fact enables us to conclude that CBE variations observed at the raw-data level propagate along the scanners pipeline and are observed at the final image level. In the same way, the simulations with therapeutic ultrasound allow us to validate past studies were temperature estimation strategies were performed based on the assumption that changes B-Mode gray-level was due to changes on CBE.

Keywords: Change in Backscattered Energy, Ultrasound, COMSOL simulations.

Introduction

Ultrasound (US) is widely used in clinical diagnostic due to its portability and low-cost instrumentation. In the last decades two important concerns of US research were: the improvement of the quality of the image, and the development of trustworthy methods for non-invasive temperature estimation [1,2]. Concerning improvement of image quality, modalities such as harmonic imaging by administration of contrast agents, and elastography were the most important ones. On the other hand, different features from US were extracted aiming to estimate temperature non-invasively. Among the extracted features, changes on backscattered energy (CBE) was one of the more promising ones. CBE is influenced by temperature because they are sensitive to changes in speed-of-sound (SOS) and medium expansion/compression. In literature, it is reported that CBE vary monotonically positive in lipid medium scatterers and monotonically negative in aqueous medium scatters [3].

Recent studies observed variations similar to CBE, computed from raw ultrasound data, in conventional B-mode ultrasound images [4]. In pixel-by-pixel analyses, intensity increases, or decreases with temperature were observed, leading to raising the hypothesis that CBE variations at the raw data level propagate along the processing pipeline of the ultrasound scanners, and are observed in the final images.

In this paper, and aiming to validate the previous hypothesis, we present two simulation models that enable to simulate CBE in the presence of two different heating sources: water-bath heating, and heating by therapeutic ultrasound. Simulations were performed in the finite element software COMSOL Multiphysics® (COMSOL, Inc).

Materials and Methods

I. Water-bath heating simulation

The water-bath heating simulation reproduces the experimental setup presented in Fig. 1.

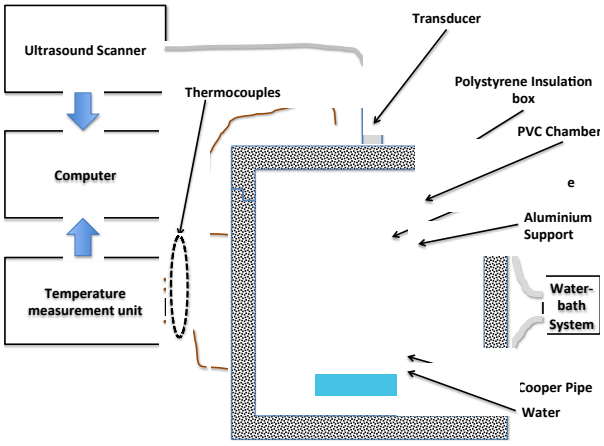

Fig. 1: Experimental setup for image acquisition and temperature change. The experiment is composed by six main parts: a PVC chamber (green) that contains water (blue), a cooper pipe (reddish brown) and an aluminum support (gray); a water temperature control system; an ultrasound scanner; a temperature measurement system; a computer; and a polystyrene isolation box.

In this simulation, the heating process is achieved by simulating the flow of hot water, at 46ºC, in a cooper pipe, which is considered as the heating source. The water inside the chamber is heated, and consequently the solid structures, including the tissue sample under analysis. The simulation model is, in fact, a series of coupled sub-models as presented in Fig. 2.

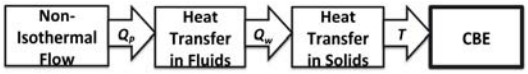

Fig. 2: Coupling between different COMSOL modules. Where Q_P is the heat source from the pipe. Q_W is the transferred heat from the embedded water to the solid structure and T is the system temperature.

Fluids heating convection are ruled by the following equation [5,6]:

$$\rho C_p \frac{\partial T}{\partial t} + \rho C_p u.\nabla_t T = \nabla_t k \nabla_t T + Q_P \qquad (1)$$

Where ρ (kg/m^3) is the density, T(K) is the water temperature inside the chamber, C_P(J/(kg.K)) and k(W/(m.K)) are the water specific heat at constant pressure, and thermal conductivity, respectively. The term Q_P from equation 1 is the heat source originated by the cooper pipe and is defined by: $Q_P = hz(T - T_P)$, where z(m) is the pipe perimeter, and h(W(m^2.K)) is the heat transfer coefficient.

At this point, the water inside the chamber is considered as the heat source for the solid structures (aluminum support and sample). The solids heating conduction process is defined by [6]:

$$\rho C_p \frac{\partial T}{\partial t} = \nabla_t k \nabla_t T + Q_W \qquad (2)$$

Where Q_W(W/m), is the transferred heat from the embedded water to the solid structure.

II. Therapeutic US heating

The other heating approach was by therapeutic ultrasound, which is used in clinics to promote thermal therapies, such as hyperthermia. US heating is simulated by a piston-like transducer. Another aspect in this model is that we consider the bio-heat transfer equation, approaching the simulation to an in-vivo scenario. Fig.3 presents the coupled sub-models used in this simulation.

Fig. 3: Coupling between the different modules for therapeutic US heating simulation, where I is the sound wave intensity and T is the system temperature.

The compression and rarefaction epochs imposed by the US induces friction between the media particles, which promotes temperature increases. The US absorption process decreases signal intensity through the sample. The heating process is defined by the bio-heat transfer equation [6]:

$$\rho C_t \frac{\partial T}{\partial t} = \nabla.(k_t \nabla T) + \omega_b \rho_b C_b (T_a - T) + Q + q_m \qquad (3)$$

Where, ρ, C_t and k_t are the density, specific heat and thermal conductivity of the sample tissue, ω_b, ρ_b, C_b are the perfusion rate, density and specific heat of the blood, T_a is the arterial temperature, T is the tissue temperature at instant t, Q is the external heating source and q_m the metabolic heating.

III. CBE simulation

The backscattered energy depends on the attenuation and on the backscattered coefficients of each medium. The backscattered coefficient, also, depends on SOS and density. CBE is modeled in literature by [2]:

$$CBE(T) = \frac{\alpha(T_R)}{\alpha(T)} \frac{\eta(T)}{\eta(T_R)} \frac{1 - e^{-2\alpha(T)z}}{1 - e^{-2\alpha(T_R)z}} \qquad (4)$$

where T is the temperature in ºC, T_R is a reference (basal) temperature, $\alpha(T)$ is the temperature-dependent attenuation, $\eta(T)$ is the backscattered coefficient, and z is the path traveled by a given US wave within the sample volume.

The ratio between the pretended backscattered coefficient (at a given temperature) and the basal backscattered coefficient (at 37ºCelsius) is defined by [2]:

$$\frac{\eta(T)}{\eta(T_R)} = \frac{\left(\frac{\rho_m c(T)_m^2 - \rho_s c(T)_s^2}{\rho_s c(T)_m^2}\right)^2 + \frac{1}{3}\left(\frac{3\rho_s - 3\rho_m}{2\rho_s - \rho_m}\right)^2}{\left(\frac{\rho_m c(T_R)_m^2 - \rho_s c(T_R)_s^2}{\rho_s c(T_R)_m^2}\right)^2 + \frac{1}{3}\left(\frac{3\rho_s - 3\rho_m}{2\rho_s - \rho_m}\right)^2} \qquad (5)$$

where ρ_m is the medium density, $c(T)_s$ is the temperature-dependent SOS of the scatterers and $c(T)_m$, the SOS of the medium.

In literature [7], two types of scatterers are assumed: lipid-based and aqueous-based scatterers, both being part of water-based medium. Table 1 presents the attenuation, SOS and density variation for each type of scatterer. Where $\alpha(T)$, $c(T)_m$, $c(T)_{ls}$, $c(T)_{as}$, ρ_{ls} and ρ_{as} are the temperature dependent attenuation coefficient, SOS in soft-tissue, SOS in lipid scatterers, SOS in aqueous scatterers, density of lipid scatterers and range density of aqueous scatterers, respectively. We implemented equations 4 and 5 in COMSOL and generated solutions over time and temperature. In both simulation models regions with different types of scatterers were defined, thus enabling the observation of CBE with temperature. For the water bath heating experiment we defined regions based on a porcine kidney sample. Images from the sample were acquired using the water-bath experimental setup (Fig. 1).

Table 1 – Properties for CBE simulation equations [7]

Prop	Equation/Value
$\alpha(T)$	$0.98 - 0.003T + 0.8e^{-3}T^2 - 0.68e^{-5}T^3$
$c(T)_m$	$1533.1 + 3.06T - 0.03T^2$
$c(T)_{ls}$	$1810.7 - 13.89T + 0.10T^2$
$c(T)_{as}$	$1471.4 + 4.00T - 0.04T^2$
ρ_{ls}	0.95
ρ_{as}	[1.05;1.2]

As a result from this image acquisition, we observed that different intensity changes occurred in the presence of temperature changes. In a pixel-by-pixel analysis, we observed that positive, negative and undefined correlations exist, being possible to compute an image as presented in Fig. 5a (more details in [4]). Green pixels represent a positive correlation (above 0.5) between the gray-level intensity variation and temperature, red pixels represent a negative correlation (under -0.5) between gray-level intensity variation and temperature, and the blue pixels represent the undefined correlations (with correlation in range of [-0.5 ; 0.5]). We hypothesize that these variations are due to CBE changes. Thus, aiming to prove this hypothesis we defined regions with different scatter types based on the image in Fig. 5a. Due to computational limitations the number of regions were simplified, resulting in the simulations in Fig. 5b and 5c. For the therapeutic ultrasound heating simulation we simplified the definition of the regions, i.e. we created a kidney sample by defining randomly regions with aqueous or with lipid scatterers in the geometry. This is a valid approach since we just wanted to observe the influence of US heating on the medium, and confirm past assumptions that it is possible to compute temperature maps from B-mode images, because average intensity changes are due to CBE changes [8].

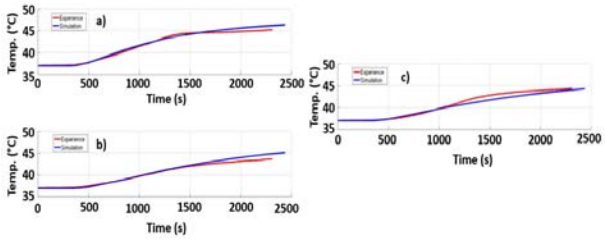

Fig. 4: Simulated (Blue line) vs measured (Red line) temperatures. a) In a specific position inside the medium and chosen to be approximately the same position as used in the real laboratory experiment. b) Water temperature at the top of the chamber. c) Water temperature at the bottom of the chamber.

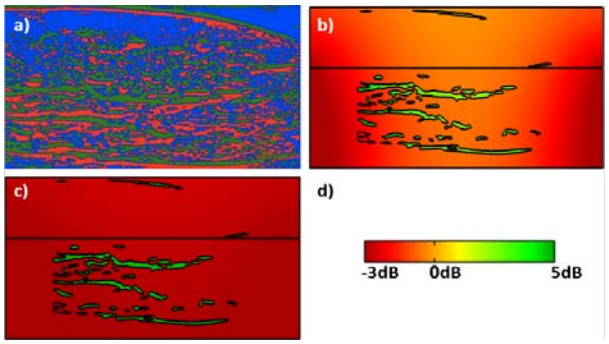

Fig. 5: Simulated CBE variation with temperature at 900s (b) and 2280s (c). a) Correlation between temperature and pixel intensity change, obtained from conventional B-mode images (More details on [4]). Green pixels indicate regions with positive correlations with temperature, red regions negative correlations, and blue regions undefined correlations (i.e., correlations coefficients less than 0.5). d) Color scale related with the CBE simulations in b) and c).

Results & Discussion

Fig. 4 represents the simulated temperature evolution as compared to the measured temperatures for the water-bath heating. The temperatures from simulation are closely related with the temperatures in the experiment, once the root mean square difference is less than 0.52 ºC for all the three-measured points.

Looking at Fig. 5b and Fig. 5c, the temperature increase results in an increase or decrease of CBE, depending on the scatterer types defined for a given region. In Fig. 5b a gradient can be observed due to the inhomogeneous heat diffusion at time instants close to the beginning of the simulation. While time is running, temperature stabilizes at a high level (46ºC) and a homogenous CBE is attained (Fig. 5c). An important observation is that Fig. 5b and 5c are simplified versions of Fig. 5a. This validates the assumption that the intensity variations observed in US images due to temperature are caused by CBE observed at the raw level.

Fig. 6 represents the CBE variation with temperature. The variation tends to be monotonically positive or negative, as theoretically expected. The therapeutic US heating simulations are presented in Fig.7. The simulation considered a frequency of 1MHz and a diameter of 32mm for the therapeutic ultrasound transducer. It was observed that the sample tissue with random media achieves the final temperature (46ºC) in half of the time (1020s) of the water-bath heating; therefore, a homogenous temperature is achieved in the central region close to the therapeutic transducer. CBE vary in the same way as in the previous simulation. With the temperature increase, in the lipid-medium tissues zones (represented as green), CBE increase and

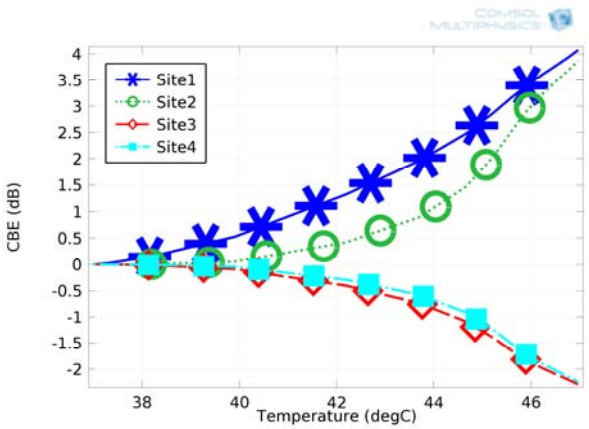

Fig. 6: CBE variation with temperature measured at random sites in kidney sample tissue.

in the aqueous-medium tissues (represented as red), CBE decrease. In [8] we developed non-invasive temperature estimations based on conventional B-mode images, when an ex-vivo tissue sample was heated by therapeutic ultrasound. Again, we hypothesized that these estimations were possible because CBE at the raw US data level propagated along the processing pipeline of conventional scanners. With this simulation we concluded that therapeutic US heating induces CBE patterns similar to the temperature maps obtained in [8].

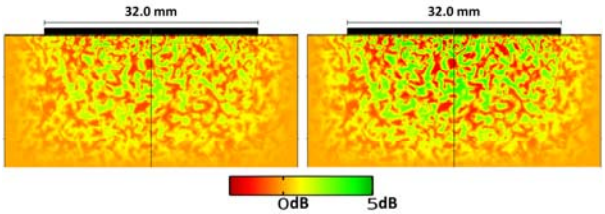

Fig. 7: Simulated CBE variation in therapeutic US heating: a) 390s and b) 1020s.

Fig. 8 present CBE variations from arbitrary places in sample tissue. More linear CBE curves were observed than in the water-bath experiment. This can be explained by the fast heating, resulting in less pronounced steady-state temperature gradients in the area under study.

Conclusion

This paper presents two simulation models that led to the conclusion that CBE changes at the raw data level propagate along the processing pipeline of conventional US scanners and have an expression in the final images. Thus, one can extrapolate that by processing the final images it is possible to extract similar information as by processing raw data, and consequently to develop trustworthy methodologies based on CBE theoretical assumptions. Future developments will consider models that will be able to explain undefined correlations observed in regions with mixed scatterer types.

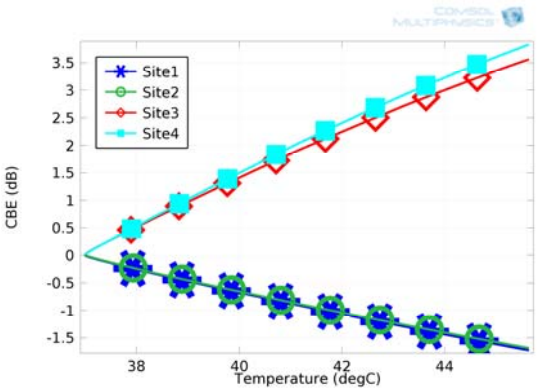

Fig. 8: CBE variation with temperature measured at random sites in the sample with random mediums.

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References

[1] Simon C., VanBaren P., Ebbini E.S.. Two dimensional temperature estimation using diagnostic ultrasound Ultrasonics, Ferroelectrics and Frequency Control, IEEE Transactions on. 1998;45:1088 -1099.

[2] Arthur R M, Straube W L, Trobaugh J W, Moros E G. Non-invasive estimation of hyperthermia temperatures with ultrasound. International journal of hyperthermia. 2005;21:589–600.

[3] Arthur R. Martin, Straube William L., Starman Jared D., Moros Eduardo G.. Noninvasive temperature estimation based on the energy of backscattered ultrasound Medical Physics. 2003;30:1021.

[4] Teixeira C. A., Fontes A., Kruger M. A. Von, Alvarenga A. V., Pereira W. C. A.. Expressiveness of temperature-induced changes in backscattered energy in conventional B-mode images in Proceedings of Meetings on Acousticsl;19:075006 2013.

[5] Munson Bruce Roy, Young Donald F, Okiishi Theodore Hisao. Fundamentals of fluid mechanics. New York 1990.

[6] COMSOL . COMSOL Multiphysics User’s Guide 2012.

[7] Straube W L, Arthur R M. Theoretical estimation of the temperature dependence of backscattered ultrasonic power for non-invasive thermometry. Ultrasound in Medicine and Biology. 1994;20:915922.

[8] Teixeira C. A., Alvarenga A. V., Cortela G., Kruger M. A. Von, , Pereira W. C. A..Feasibility of non-invasive temperature estimation by the assessment of the average gray-level content of B-Mode images. Ultrasonics, 2014;54:1692-1702.