Development of Ultrasound Thermometry Technique Using Tissue-Mimicking Phantom

Aleksander E. Berkovich, Evgeni M. Smirnov, Andrey D. Yukhnev, Yakov A. Gataulin, Daria E. Sinitsyna, Dmitriy A. Tarakhov
Peter the Great St. Petersburg Polytechnic University, 29 Politechnicheskaya St., St. Petersburg, Russia
E-mail: a.yukhnev@mail.ru

Abstract
A phantom with heating of tissue-equivalent material has been designed to elaborate the method of ultrasound thermometry. Algorithms have been developed to evaluate backscatter ultrasound signals shift caused by temperature-sensitivity of the speed of sound. Time-dependent material temperature field needed for estimation of the method accuracy was measured with thermistor probes.

1. Objectives
High Intensity Focused Ultrasound is widely used in various fields of medicine [1]. Ultrasonic devices for blood coagulation in therapy of vascular injury and internal organs, for veins obliteration, as well as for ablation of breast and thyroid tumors are currently under extensive development worldwide. Our experience in this area was partially presented in [2], where special equipment for testing of focused ultrasound devices are described.

For carrying out of hyperthermia process and for its efficiency control a real-time temperature monitoring is needed. Currently, the magnetic resonance method is the only one available for non-invasive thermometry, but it requires rather expensive equipment and has a low temporal resolution. Ultrasound thermometry considered as an alternative, cheap and fast, non-invasive technique is under development for about 20 years [3]. It has already shown good results during tests using various ultrasonic phantoms [4], as well as biological tissues and living organisms, but clinical applications of the technique is yet to come.

The present contribution covers description of equipment used for development and optimization of the ultrasound thermometry, as well as algorithms for backscatter ultrasound signal data-processing. It is a step to get an effective and low-cost high-precision medical tool via combination of therapeutic and diagnostic ultrasound devices.

2. Methods
Ultrasound thermometry technique is based on phenomenon of the speed of sound change due to a temperature increase. This phenomenon causes a time shift of the backscatter ultrasound signals from a heated material relative to the reference signal of the unheated material. The temperature increase, $\Delta T$, at a current point can be calculated as $\Delta T = K \varepsilon$, where $\varepsilon$ is the so-called ultrasound signal thermal shift strain defined as $\varepsilon = \frac{d}{dz} (\Delta d)$, where $z$ is a coordinate along the ultrasound beam direction and $K$ is a material dependent coefficient [3].

Phantom with a tissue-equivalent material (agar-agar filled with graphite) is used in the present work. The material simulates density $(1050 \text{ kg/m}^3)$, speed of sound $(1550 \text{ m/s})$, attenuation coefficient $(0.7 \text{ dB/(cm-MHz)})$, heat capacity $(3700 \text{ J/kg-°C})$ and heat conductivity $(0.59 \text{ W/m-°C})$ of a biological tissue. Nearly two-dimensional temperature field is created by heating a nichrome wire, $0.3 \text{ mm}$ diameter and $150 \text{ mm}$ length, located inside the material at the depth of $20 \text{ mm}$ (Figure 1).

During heating the wire the material temperature at chosen distances from the wire is measured with six thermistor probes. The heating time, $t$, is counted from switching-on the DC power supply. The temperature measurement uncertainty is estimated as $0.1 \text{ °C}$. Simultaneously with measurements of time-dependent material temperature (results are illustrated in Figure 2), ultrasound signals are...
registered near the heated wire in an ultrasound-linear-transducer scanning plane, 20 mm width and 10 mm depth, perpendicular to the wire. An array of data for one record includes 161x512 values of ultrasound signal. In order to eliminate the temperature probes influence on the ultrasound signal and taking into account the temperature field uniformity along the wire, zones for temperature measurement and for ultrasound scanning are separated in space, as shown in Figure 3.

Figure 4 shows the temperature increase profile $\Delta T(z)$ measured at $t=20$ s with six probes positioned in the $y=0$ plane (Figure 3). The depth coordinate, $z$, as well as the transversal coordinate, $y$, are counted from straight lines passing through the heated wire center. The profile covers experimental points obtained from measurement data averaged over 5 experiments. The approximation/interpolation curve shown was obtained with the least-squares smoothing procedure.

Figure 1. Experimental setup: 1 – tissue-mimicking phantom; 2 - heated nichrome wire, 3 - temperature probes, 4 - ultrasound probe, 5 - acoustic window, 6 - DC power supply, 7 - ultrasound scanner, 8 – ADC, 9 – PC

Figure 2. Temperature temporal variations measured at different distances from the heated wire center. Time-points of backscatter ultrasound signal record are put on the $t$-axis

Figure 3. Location of zone 1 for temperature measurements with six probes and zone 2 for backscatter ultrasound signal record (two sections of the phantom are shown)

Figure 4. The measured temperature increase profile in the vicinity of the heated wire at instant of 20 s data of 6 temperature probes and an approximation curve
Figure 5 illustrates the time shift of the ultrasound signal of the heated material relative to the reference signal. In [3,5] the time-shift is evaluated by a cross-correlation procedure. We suggest an alternative procedure, which demands less calculation and is accurate enough. The main idea is to evaluate the time-shift $\Delta d$ as a difference between the average values of the direct and reflected signal for each half-cycle of oscillations.

![Image of ultrasound signal shift](image_url)

**Figure 5. Illustration of the ultrasound signal shift: line 1 - before heating, line 2 – at a heating instant**

Generally, difficulties in calculating the signal shift are caused by variations of the signal amplitude and by a rather limited number of recorded points for each half-wave of the curve (Figure 5). For more accurate and stable evaluation of the shift of the $i$-oscillation, it was determined as a difference, $\Delta d_i$, between centers of the corresponding groups of points (taken from the data obtained before and after heating), with the same sign of deviation from the mean value. The group center depth coordinate was calculated as $z_i = \sum_{k=1}^{N_i} A_k z_k / \sum_{k=1}^{N_i} A_k$, where $A_k$ is the recorded ultrasound signal value, index $k$ runs over all the points (1 to $N_i$) corresponding to a current half-oscillation.

Unfortunately, experimental signals can contain various kinds of defects: splitting an oscillation into two oscillations, occurrence of an unformed oscillation etc (Figure 5). The presence of these defects can lead to considerably errors in evaluations of the half-oscillation coordinates, and hence in the evaluation of the signal-shifts. As a rule, considerable defects arise near the heated wire. In order to reduce the influence of these defects, a special procedure is introduced for data conditioning. The procedure is based, first, on introducing a condition on the rate of the shift change: $|f(z_{i+1}) - f(z_i)| < c(z_{i+1} - z_i)$, where $f(z)$ is the magnitude of the shift $\Delta d_i$ at point $z_i$, and, second, on a limitation of the difference of the shifts for two neighbor points: $|f(z_{i+1}) - f(z_i)| < g$.

For evaluation of the ultrasound signal thermal shift strain $\varepsilon$ demanding application of a differencing procedure, the conditioned data for $\Delta d_i$ should be approximated with an appropriate function. Some researchers [5] use the probability integral assuming that the temperature field satisfies the constant-
properties heat conductivity equation. We use the arctangent function \( \varphi(z) = a(\pi / 2 + \arctg(bz_j)) \) for this purpose, with two coefficients, \( a \) and \( b \), that are defined using the least-squares method involving an iterative procedure to remove points with excessive deviations. The condition for removing such points is as follows: \( |f_j(z) - \varphi_j(z)| > \gamma\sigma_j \), where \( \sigma_j = \sqrt{\frac{1}{N}\sum_{i=1}^{N}(f^2 - \varphi^2)} \) is the root-mean-square deviation, \( \varphi_j(z) = a_j(\pi / 2 + \arctg(b_jz_j)) \) is the approximation curve for the \( j \)-iteration. The experience of using the conditioning/approximating procedures developed has showed that the best values of the parameters introduced are as follows: \( c = 0.1 \), \( g = 4x(10/512) \) mm, \( \gamma = 1.5 \), \( j = 1 \div 4 \).

3. Results

Figure 6 presents an example of results from thermal shift data processing by the above-described procedures. The ultrasound data were taken for the beam passing 0.8 mm away from the center of the heated wire. To relate these data with values of temperature increase, the assumption of axial symmetry of the temperature field was adopted, so that \( \Delta T(y=0, z=0) \) is assumed to be equal to \( \Delta T(y=0, z=0.8 \text{ mm}) \) measured by a temperature probe positioned in zone 1 (Figure 3). Symbols in Figure 6 show experimental values of the signal shift that have remained after application of the data conditioning procedure and after removing points with excessive deviations from the approximating curve shown by solid line.

The same data processing procedure was applied to the data recorded for all the beams (of 161) emitted by the ultrasound transducer. The ultrasound signal thermal shift strain is evaluated as \( \varepsilon = \frac{d}{dz}(\Delta d) = \frac{ab}{1+(bz)^2} \), according to the adopted approximating (arctangent) function. The thermal strain map obtained for the instant of 20 s after the wire heating start is shown in Figure 7 (as previously, the co-ordinate origin is located at the center of the wire). Generally, the map corresponds to axisymmetric temperature field arising during the unsteady heating of the tissue-mimicking material. However, a notable deviation from the axial symmetry is observed in the central part of the map that might be attributed to interaction of ultrasound beams with the wire.

As mentioned in Section 2, the temperature increase \( \Delta T \) at a current point of ultrasound measurements can be calculated as \( \Delta T = K\varepsilon \). Since the factor \( K \) is unknown for the used tissue-mimicking material, a part of the obtained experimental data (the smaller one) can be used to estimate this coefficient, and the remaining part (the bigger one) can be used to check an overall consistency of the measurement results obtained with thermometer probes and with the ultrasound technique.

In the present work, the data corresponding to the points positioned along the \( z=0 \) line at 0.8 mm < \(|y| < 5.0 \text{ mm} \) was used for the purpose of the \( K \)-factor estimation. Figure 8 presents correlation of the thermal shift strain evaluated at these points and the temperature increase measured with temperature probes (here again the assumption of the temperature-field axial symmetry was used). Assuming a linear relation between \( \Delta T \) and \( \varepsilon \), we have evaluated the value of the \( K \)-factor for the tissue-equivalent material used as 540 °C.

The \( K \)-factor obtained was used to convert the thermal shift strain map given in Figure 7 into a temperature increase map shown in Figure 9. The central stripe-like zone in the temperature increase map, with \(|z| < 0.8 \text{ mm} \) has been omitted due to unreliability of the data for this zone where the wire produces strong distortions of the ultrasound field. Comparing the temperature increase map obtained with the ultrasound thermometry with the map reconstructed from the temperature probes measurements (Figure 10), one can conclude that the ultrasound thermometry technique developed produces reasonable results. Additionally, Figure 11 presents difference between the temperature increase values obtained with the two methods. In the main, the difference does not exceed 2 °C.
Figure 6. Ultrasound signal shift versus depth for the ultrasound beam positioned at $y=0.8\text{mm}$: (symbols) experimental data for $t=20\text{ s}$ and (line) approximation by arctangent function.

Figure 7. Map of ultrasound signal shift strain for $t = 20\text{ s}$.

Figure 8. Correlation of the temperature increase and the ultrasound signal thermal shift strain: experimental data ($z=0\text{ mm}$) are approximated with a linear function: $K = 540$, $R^2=0.88$.

Figure 9. Temperature increase map obtained with the ultrasound thermometry for the instant of 20 s.

Figure 10. Axisymmetric temperature increase map reconstructed from the temperature probes measurements for the instant of 20 s.

Figure 11. Map of difference between temperature increase obtained with the ultrasound thermometry and measured with the temperature probes.
4. Conclusions

A phantom of tissue-mimicking material has been designed to elaborate in detail the method of ultrasound thermometry based on phenomenon of temperature-sensitivity of the speed of sound. A long heated wire, 0.2 mm diameter, was used to get time-dependent axisymmetric temperature increase relative to the reference stationary state of the phantom material.

Analysis of the primary backscatter ultrasound data used for evaluation ultrasound signals shift due to temperature increase has shown that typically these data contain various kinds of defects, mostly pronounced in the vicinity of the wire and in the region of relatively small temperature increase. Original algorithms have been developed for the primary data conditioning and their approximation of a smooth function needed for quantitative evaluation of the thermal shift strain.

Synchronized test data for time-dependent material temperature field was obtained with a set of thermistor probes. These data were used both for the evaluation of the $K$-factor in the adopted linear relation between values of local thermal shift strain and temperature increase and for estimation of an overall accuracy of the developed ultrasound thermometry technique. For the tissue-mimicking material used, a value of the $K$-factor obtained by correlation of the mostly reliable part of the thermal shift strain data and the test temperature data is equal to 540 °C.

Overall comparison of the temperature maps obtained with the ultrasound thermometry with temperature probes measurements has shown a reasonably accuracy of the developed method of ultrasound diagnostics of temperature field. In the main, the difference does not exceed 2 °C.

Further work will be devoted to additional improvement of accuracy and robustness of the technique presented, as well as to its application to the case of anisotropic tissue-mimicking materials that approximate biological objects more properly.

Acknowledgments. The study has been carried out with the financial support of the Ministry of Education of the Russian Federation (grant No.2015-2018-07-044 “Multifunctional complex for the diagnosis and therapy of breast and thyroid tumors”).

References
[1] Khokhlova V.A., Crum L.A., ter Haar G., Aubry J.-F. (Eds.) 2017 High Intensity Focused Ultrasound Therapy (Berlin: Springer)
[2] Berkovich A.E., Bursian A.A., Senchik K.U. et al. Biomed Eng (2016) 50: 96
[3] Maass-Moreno R., Damianou C.A., Sanghvi N.T. J. Acoust. Soc. Am. (1996) 100: 2522
[4] Fuhrmann T.A., Georg O., Haller J. et al. J. Therapeutic Ultrasound (2016) 4: 28
[5] Anand A., Kaczkowski P. J. Ultrasound in Med. & Biol., (2008) 34, 1449