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Effect of Point Spread Function in Ultrasound Imaging on Estimated Heat Source of High-Intensity Focused Ultrasound in Thermal Strain Imaging

熱歪みイメージングを用いた HIFU 加熱推定における超音波点拡がり関数の影響

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1. Introduction

High-intensity focused ultrasound (HIFU) is a minimally invasive therapeutic modality. To enhamce the treatment accuracy, *in situ* estimation of the focal region and the temperature rise in prior to the HIFU coagulation treatment should be useful for the treatment planning.

Among the temperature estimate techniques, thermal strain (TS) imaging based on the backscttered ultrasound RF-echo is superior in the real-time capabitity. The temperature rise of several degrees Celsius is estimated from a linear relationship between thermal strain and temperature rise via tissue dependent coefficient[1)]. However, the beam width of therapetuic and imaging ultrasound are often in a similar size, which may lead to underestimation of the strain.

The objective of this study is to investigate the effect of elevational point spread function (PSF) on the HIFU heating estimation by TS imaging.

2. Material and Method

2.1 Thermal strain imaging

When a tissue is exposed to HIFU, temperature rise causes temporal shifts of RF-echo signals. The temporal shifts result from two physical phenomena: the thermal expansion and the change in the sound speed of the tissue. The temperature rise induced by HIFU irradiation, $\delta\theta(z)$, is expressed as

$$\delta\theta(z) = \frac{c_0(z)}{2}\left(\frac{1}{\alpha(z)-\beta(z)}\right)\cdot\frac{\partial}{\partial z}\left(\delta t(z)\right) \qquad (1)$$

where $c_0(z)$ is the sound speed at the initial temperature, $\alpha(z)$ is the coefficient of thermal expansion, $\beta(z)$ is the relative temperature coefficient of sound speed, and $\delta t(z)$ is the time-shift on the echo at a depth *z*.

The coefficient $\beta(z)$ in an agar-based tissue-mimicking material (TMM) phantom was assumed to be 1.3×10^{-3} /K over a temperature range of 23 to 37℃ calculated from the temperature coefficient of 2.1 m/s/K measured in the previous research[2)]. The coefficient $\alpha(z)$ was assumed to be the same that in water, which is at least an order of magnitude smaller than $\beta(z)$ in water bearing tissue at around 37℃[3)].

2.2 Experimental setup and Sequence

The experimental setup is shown in **Fig. 1**. Experiments were performed in an acrylic tank filled with degassed water at room temperature. A 256-channel HIFU transducer with both diameter and focal length of 120 mm was equipped on the side of the tank and driven at 1 MHz by a staircase voltage driving system (Microsonic). A sector imaging probe (Hitachi Aloka Medical UST-52105) with a center frequency of 3.5 MHz was set in the central hole of the HIFU transducer, connected to the ultrasound imaging system (Verasonics Vantage 256) for RF-data acquisition. In order to measure the temperature induced by HIFU exposure, a 0.3 mm diameter thermocouple (CHINO) was placed at the point where the temperature rise was maximum. As a sample tissue, an agar-based TMM phantom (OST) was located so that it contains the focal region.

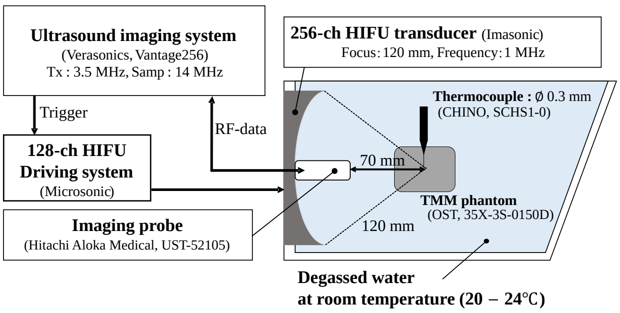

Fig. 1 Schematic of experimental setup.

Fig. 2 shows the overall sequence of ultrasonic exposure. Prior to inducing temperature rise, RF-data were acquired with 5 diverging waves steered at angles of -6, -3, 0, 3, and 6°. After that, HIFU was irradiated to induce a temperature rise of several

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degrees Celsius, with a constant exposure duration of 0.9 s at an intensity of 330 W/cm^2. The RF-data were collected from 2 to 28 s after the end of HIFU irradiation. The axial displacement was calculated from the frames before and after the HIFU exposure by applying the autocorrelation method[4]. The strain was obtained by differentiating the axial displacement along the axial direction and smoothed using 1D Savitzky-Golay filter and 2D Gaussian filter.

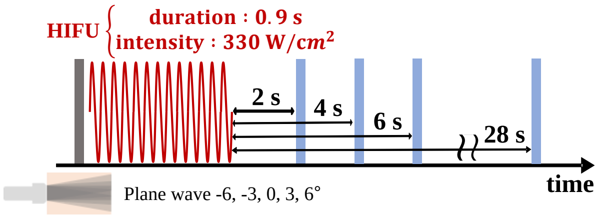

Fig. 2 Sequence of HIFU and RF acquisition.

3. Result and Discussion

Fig. 3 shows the set of strain distributions acquired at 2, 8, 16, and 28 s after the end of HIFU exposure. Each distribution is the average of 6 results.

From the results of thermocouple measurement, the absorption coefficient of TMM phantom was estimated to be 45 dB/m/MHz. For examining the quantitative relation between measured thermal strain and temperature rise, axisymmetric temperature maps were calculated using HITU simulator[5]. The temperature maps were convolved with the elevational PSF and averaged in the elevational direction based on the pressure distribution to consider the effect of spatial averaging in elevational direction (**Fig. 4**). The estimated temperature rise was calculated using $\beta = 1.3 \times 10^{-3}$ /K and $\alpha = 0.1 \times 10^{-3}$ /K (those of water)[6]. Both estimated and simulated temperature rise distributions were averaged in ROI (4.5 mm×5 mm) centered on the geometric focus. In **Fig. 5**, when the convolution was performed with an elevation width of 8.5 mm, the estimated temperature and simulation results show a good agreement. It is observed that the spatial averaging effect of the temperature can be almost ignored at not less than 28 s after the end of HIFU duration due to thermal conduction.

4. Conclusion

In this study, we demonstrated that the thermal strain was underestimated due to elevational spatial averaging effect. The effect was reduced as the time lapses from the end of HIFU exposure. Furthermore, the results support that the convolution of temperature distribution compensated the underestimation due to the limited spatial resolution of ultrasound imaging.

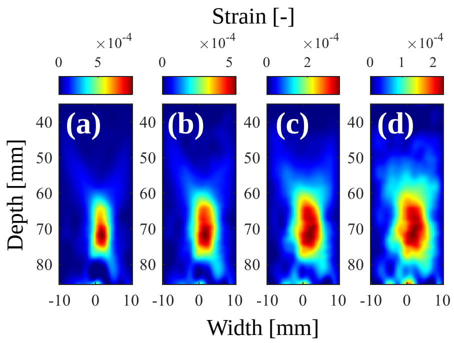

Fig. 3 Set of strain distributions acquired at (a) 2, (b) 8, (c) 16, and (d) 28 s after the end of the HIFU exposure.

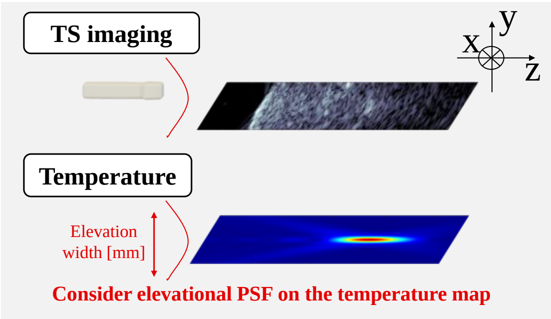

Fig. 4 Interpretation of analysis method on the spatial averaging effect in the elevational direction.

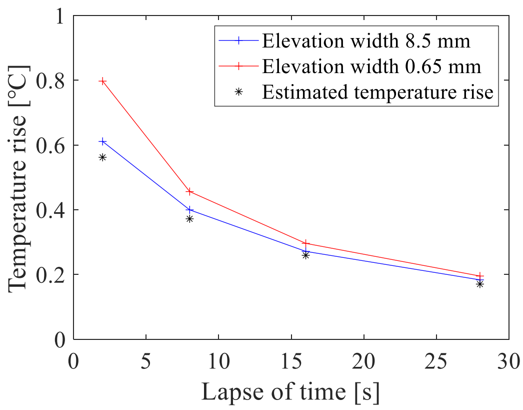

Fig. 5 The effect of spatial averaging in the elevational direction on the estimated temperature.

References

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