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ORIGINAL ARTICLE

Ultrasonic measurement of sound velocity fluctuations in biological tissue due to ultrasonic heating and estimation of thermo-physical properties

Yukako Tsujimoto[1] · Mai Morimoto[1] · Naotaka Nitta[2] · Iwaki Akiyama[1]

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Abstract

Purpose Tissue characterization in terms of the differences in thermo-physical properties of biological tissues was investigated in this study. The objective was to measure the ratio of variation in sound velocity due to ultrasonic heating and to derive the relational expression between the ratio and thermo-physical properties.

Methods The ratio of sound velocity variation before and after the temperature rise of tissue samples exposed to ultrasound was measured by ultrasonic pulse echo method. The thermo-physical properties were estimated for a tissue-mimicking material and porcine muscle and fat tissues due to theoretical expression. The transducer for heating had a resonance frequency of 3.2 MHz, and the transducer for measurement of sound velocity variation had a resonance frequency of 5.2 MHz.

Results In the phantom study, the measured values of the temperature rise agreed with the values calculated by a finite element method (FEM). The estimated values of the temperature rise from the sound velocity variations of muscle and fat tissues were 0.36 °C and 1.1 °C, respectively. Also, the estimated values of thermo-physical properties agreed with the reference values within an error of 10%.

Conclusions The thermo-physical properties of the porcine tissues were measured by sound velocity variation due to ultrasonic heating within the safety regulations.

Keywords Tissue characterization · Thermo-physical properties · Sound velocity variation · Temperature rise

Introduction

Many studies on ultrasonic tissue characterization have previously been reported, which attempted to acquire, in vivo, biological information equivalent to a pathology diagnosis by carrying out ultrasonic measurement of the physical characteristics of biological tissue [1, 2]. In this ultrasonic tissue characterization, researchers have investigated techniques primarily relating to the viscoelasticity of tissue, e.g., sound velocity and attenuation [3, 4]. Regarding elastography, in which the elasticity characteristic distribution of biological tissue is visualized, studies have examined its clinical utility for ultrasonic tissue characterization [5–7]. As an addition to the above, this research focuses on physical characteristics relating to the thermo-physical properties of biological tissue. When biological tissue is exposed to ultrasound, heat is produced and the tissue temperature rises. This rise in temperature depends on physical quantities characteristic of the tissue such as ultrasonic intensity, specific heat, heat conduction coefficient, and attenuation. On the other hand, it is known that the tissue's characteristic value of sound velocity varies depending on the temperature, and the temperature coefficient exhibits a distinctive value due to the tissue; for example, fat tissue exhibits a negative temperature dependence coefficient, while non-fat tissue exhibits a positive temperature dependence coefficient [8]. Furthermore, it has been reported regarding non-fat tissue that when tissue coagulates due to a rise in temperature, the temperature dependence coefficient becomes negative [9]. For these reasons, the rate of change over time in sound velocity due to ultrasonic heating has the potential to serve as a parameter for tissue characterization focused on fat content.

✉ Iwaki Akiyama
iakiyama@mail.doshisha.ac.jp

1 Medical Ultrasound Research Center, Doshisha University, 1-3 Tataramiyakodani, Kyotanabe, Kyoto, Japan

2 National Institute of Advanced Industrial Science and Technology, Tsukuba, Japan

In the field of ultrasonic therapy, on the other hand, treatment of malignant tumors is carried out through heating coagulation of biological tissue using high-intensity focused ultrasound (HIFU) [10]. When using this HIFU-based heat coagulation therapy, it is crucial to monitor the rise in temperature of tissue, and ultrasound has been proposed as a method for doing that [11–14]. This technique uses the pulse echo method to estimate changes in sound velocity accompanying the temperature rise of biological tissue. When estimating the change in sound velocity during heating, the phase change of the echo signal is extracted, and thus, it is necessary to assume that the tissue displacement during measurement does not affect the change in phase attributable to the change in sound velocity. That is, it is important to acquire the echo signal in a short time interval, in which tissue displacement in the living body can be ignored. Mano et al. have imaged the 2-dimensional distribution of the temperature rise by heating a phantom through exposure to ultrasound, and sending/receiving ultrasonic pulses; so, they are orthogonal with the heating beam [15]. Thus, in the present study, the authors measured the rate of change over time in sound velocity due to ultrasonic heating, examined techniques for estimating the state of tissue from that rate of change, measured the rate of change over time in sound velocity in porcine fat and non-fat tissue, and thereby examined the effectiveness of the technique. Also, excessive temperature rise causes protein denaturation, and thus according to WFUMB Guidelines, it was decided to keep the temperature rise of biological tissue to 4.0 °C or lower and less than 5 min, except in the case of fetuses and embryos [16]. Thus, the aim of this study was to clarify the relationship between physical quantities, particularly thermo-physical quantities, and measured changes in sound velocity due to the temperature rise produced by ultrasonic heating, and to clarify the relationship between the fat percentage of tissue and the rate of change in sound velocity in biological tissue measured when the temperature rise was 4.0 °C or less. Higher measurement precision and speed were realized in this study by a focused probe in which the transducer for heating and the transducer for sound velocity measurement were coaxially integrated. When sound velocity changes were measured in porcine fat tissue and non-fat tissue, the rate of change over time in sound velocity in fat tissue was negative, and the rate of change over time in sound velocity in non-fat tissue was positive. This suggests that it is possible to measure the thermo-physical properties of fat tissue and non-fat tissue using this technique.

Materials and methods

Bioheat transfer equation

The rate of temperature rise of biological tissue due to exposure to ultrasound is given by the following bioheat transfer equation [17].

$$\frac{\mathrm{d}T}{\mathrm{d}t} = \kappa \nabla^2 T - \frac{\Delta T}{\varepsilon} + \frac{q_\mathrm{v}}{C_\mathrm{v}} \tag{1}$$

Here, κ is the thermal diffusivity of tissue, ε is the time constant of cooling due to perfusion, ΔT is the change in temperature, q_v is the heat generation rate, and C_v is the heat capacity per unit volume. In soft tissues of the living body, the thermal diffusivity κ is in the range 0.10–0.15 mm^2/s, and the time constant ε is in the range 10–2000 s. If the initial temperature is uniform and the temperature rise is very small, then the first term on the right side (the thermal diffusion term) can be ignored, and if heating is done for a short time, the second term (the perfusion term) can be ignored. In addition, the heat generation rate q_v can be expressed as $q_\mathrm{v} = 2\alpha I$ by the ultrasonic attenuation constant α of biological tissue and ultrasonic intensity I. Based on the above, the following equation is obtained.

$$\frac{\mathrm{d}T}{\mathrm{d}t} \cong \frac{q_\mathrm{v}}{C_\mathrm{v}} = \frac{2\alpha I}{C_\mathrm{v}} \tag{2}$$

That is, the rate of temperature rise is determined by the ratio between the product of the ultrasonic intensity I and the attenuation constant α, and the heat capacity per unit volume C_v. It is assumed in this paper that ultrasonic attenuation in tissue is all due to the ultrasonic absorbtion.

Changes in sound velocity due to exposure to ultrasound and thermo-physical quantities

The rate of temperature rise due to exposure to ultrasound is given by the following equation, using the rate of change in sound velocity and the temperature coefficient of sound velocity.

$$\frac{\mathrm{d}T}{\mathrm{d}t} = \frac{\mathrm{d}c}{\mathrm{d}t} \cdot \left(\frac{\mathrm{d}c}{\mathrm{d}T}\right)^{-1} \tag{3}$$

If Eq. (2) is substituted in Eq. (3), the following equation is obtained.

$$\frac{\mathrm{d}c}{\mathrm{d}t} \cong \frac{\mathrm{d}c}{\mathrm{d}T} \cdot \frac{2\alpha I}{C_\mathrm{v}} \tag{4}$$

It is evident from the above equations that the rate of change in sound velocity due to exposure to ultrasound dc/dt is approximated by the product of the rate of change of sound velocity with respect to temperature dc/dT, the attenuation constant α, the ultrasonic intensity I, and the reciprocal of the heat capacity per unit volume. Sound velocity varies due to the temperature rise caused by exposure to ultrasound, and thus, if an ultrasonic pulse is transmitted separate from the ultrasound for heating, a time shift proportional to the change in sound velocity will appear in the echo signal. With this technique, the change in sound velocity is found from the time shift τ in the echo from the position x. If the temperature rise is very small and the relation $\Delta c(x) \ll c(x)$ is satisfied, then the change in sound velocity $\Delta c(x)$ can be approximated with the following equation [18].

$$\Delta c(x) = \frac{\mathrm{d}\tau}{\mathrm{d}x} \cong c(x)\frac{\mathrm{d}\tau}{\mathrm{d}t} \tag{5}$$

Here, the rate of change with respect to time of the time shift τ, dτ/dt, can be estimated by providing a time gate at the delay time corresponding to the echo from the measurement region in the echo signal, calculating the time shift τ of the extracted signal, while shifting the time delay t of the gate, and then finding the slope. The echo shift time τ produced by ultrasonic heating is very small, and thus the echo shift time τ is calculated to high precision by combining two methods, the cross correlation method and the autocorrelation method [19].

Also, if the heating time is taken to be Δt_{h}, then the change in sound velocity Δc is expressed by the following equation using the sound velocity temperature coefficient and Eq. (4).

$$\Delta c = \frac{\mathrm{d}c}{\mathrm{d}t} \cdot \Delta t_h = \frac{\mathrm{d}c}{\mathrm{d}T} \cdot \frac{2\alpha I}{C_{\mathrm{v}}} \cdot \Delta t_h \tag{6}$$

The following equation is obtained from Eqs. (5) and (6).

$$\frac{\mathrm{d}\tau}{\mathrm{d}t} = \frac{\Delta c(x)}{c(x)} = \frac{\mathrm{d}c}{\mathrm{d}t} \cdot \Delta t_h \cdot \frac{1}{c(x)} = \frac{\mathrm{d}c}{\mathrm{d}T} \cdot \frac{2\alpha I \Delta t_h}{c(x) C_{\mathrm{v}}} \tag{7}$$

That is, the rate of change in sound velocity $\Delta c(x)/c(x)$ becomes equal to the rate of change in echo shift dτ/dt. Also, it is a function of the sound velocity temperature coefficient dc/dT, attenuation α, sound velocity $c(x)$, heat capacity per unit volume C_{v}, ultrasonic intensity I, and heating time Δt_{h}. Thus, it is possible to measure the rate of change in sound velocity due to exposure to ultrasound, and estimate the state of the tissue from that rate of change.

In the proposed method, it is assumed that the attenuation coefficient and specific heat do not change with temperature.

The attenuation coefficient α of biological tissues is a function of temperature. dα/dT of biological tissue such as bovine liver is less than 0.1 dB/cm/°C at 3 MHz in the temperature range up to 50 °C [8, 9, 20]. Since the temperature rise during the ultrasonic exposure is less than 2–3 °C in the proposed method, the change in attenuation coefficient is less than 0.2–0.3 dB/cm/°C. The coefficient of specific heat capacity to temperature is about 0.02 J/cm^3/°C around the body temperature [21]. The change in specific heat capacity is estimated as less than 0.06 J/cm^3/°C as the temperature rise is less than 3 °C. The change of attenuation coefficient and specific heat capacity are considered to be negligible.

Measurement of temperature coefficient of sound velocity

As a measurement sample, a tissue-mimicking material (TMM) phantom, simulating soft tissue, was prepared based on the IEC 60601-2-37 standard [22]. Figure 1 shows the experiment system. The measurement sample and a transducer were placed in an incubator. The sound velocity and temperature were measured, while varying the temperature inside the incubator. The tissue sample was cut by a scalpel to be a specimen with a thickness in the range of 5–7 mm. The specimen was placed in a 20 × 22 × 5-mm depression

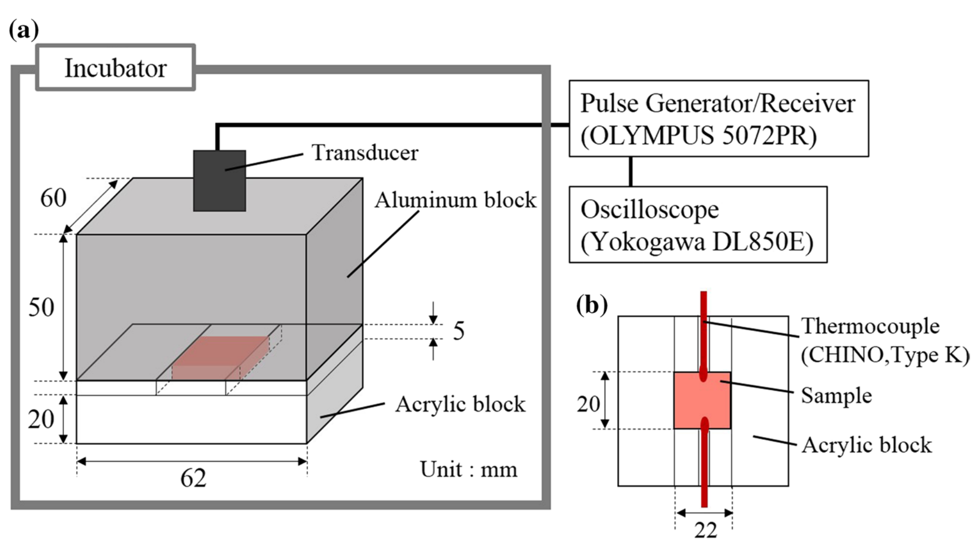

Fig. 1 Experiment system for measurement of temperature coefficient of sound velocity. A tissue sample laid between two blocks and a transducer was placed in an incubator. The temperature in the tissue was measured by two K-type thermocouples

in the center of an acrylic block, and an aluminum block was placed on top of that. The depth of the depression was 5.00 ± 0.05 mm. The thickness and the flatness of the specimen were ensured by the depth of the depression and the surface of the aluminum block and an acrylic block, respectively, since the specimen was vertically pressed and horizontally spread by the weight of the aluminum block. A planar transducer (B5K10I, JAPAN PROBE, diameter 10 mm, resonance frequency 5 MHz) was driven by a pulse generator (OLYMPUS, 5072PR), and echo signals were recorded with an oscilloscope (YOKOGAWA ELECTRIC, DL850E). The sampling frequency was 100 MHz, and the number of quantization bits was 12 bits. Propagation time within the sample was calculated from the echo signal. Two thermocouples (CHINO Type K, $\phi = 0.5$ mm) were inserted into the sample, and the temperature inside the sample was recorded with a similar oscilloscope. Sound velocity c in the sample is given by the following equation.

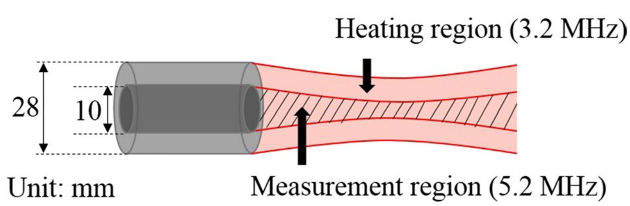

Fig. 2 The transducer for heating is used with a 3.2-MHz resonance frequency, concave ring shape, inner diameter of 10 mm, and outer diameter of 28 mm. It is arranged coaxially on the outer perimeter of the transducer for sound velocity measurement. It is used with a 5.2-MHz resonance frequency, concave circular shape, focal distance of 60 mm, and diameter of 10 mm

$$c = \frac{2d}{t_{sd}}, \tag{8}$$

where d and t_{sd} are the thickness of the sample and the propagation time inside the sample, respectively. The temperature coefficient of sound velocity in the measurement sample was found using this technique.

Two-wavelength integrated coaxial transducer (3.2K28/12I 5.2K10I R60)

In this research, a transducer in which the transducers for heating and sound velocity measurement are coaxially integrated was used to achieve a match between the heating region and measurement region. The transducer used is shown in Fig. 2. The transducer for heating (with a 3.2-MHz resonance frequency, concave ring shape, inner diameter of 10 mm, and outer diameter of 28 mm) is arranged coaxially on the outer perimeter of the transducer for sound velocity measurement (with a 5.2-MHz resonance frequency, concave circular shape, focal distance of 60 mm, and diameter of 10 mm). The normalized magnitude of peak pressure distribution of heating ultrasound formed by the ring-shaped transducer is shown in Fig. 3. It was measured by two-dimensional scanning of a hydrophone (Toray Engineering, HY05N).

Measurement of temperature rise due to exposure to ultrasound by experiments and simulation

Using a TMM phantom ($100 \times 100 \times 20$ mm) as the measurement sample, the value of the temperature rise in the ROI

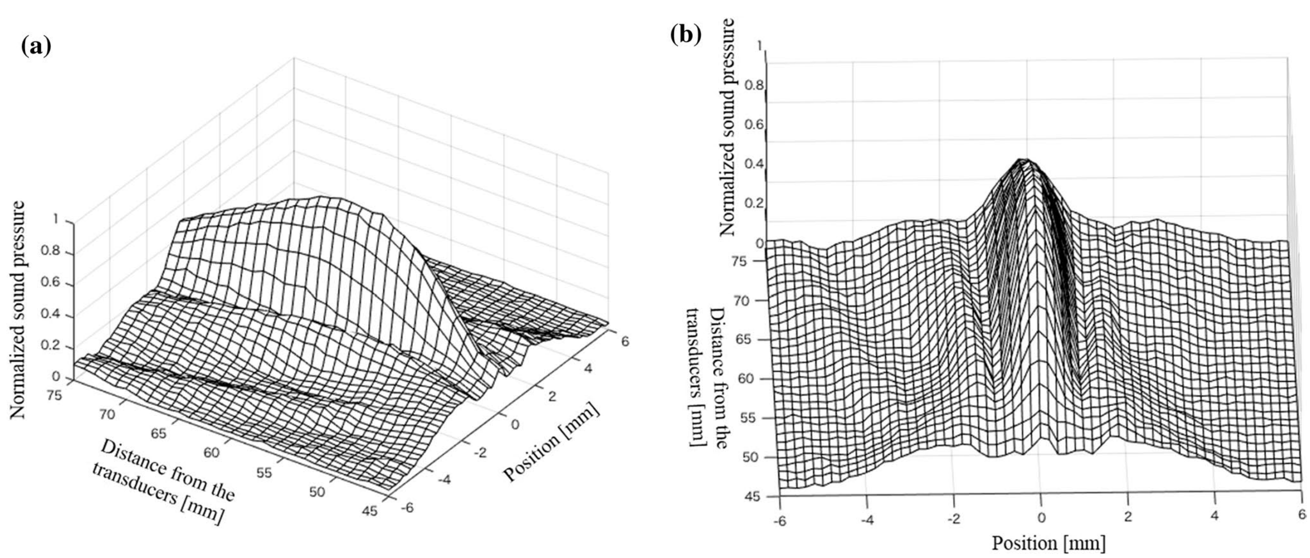

Fig. 3 The distribution of normalized magnitude of peak pressure of heating ultrasound formed by the concave ring-shaped transducer. A needle-type PVDF hydrophone (Toray Engineering, HY05N) was used for the measurement

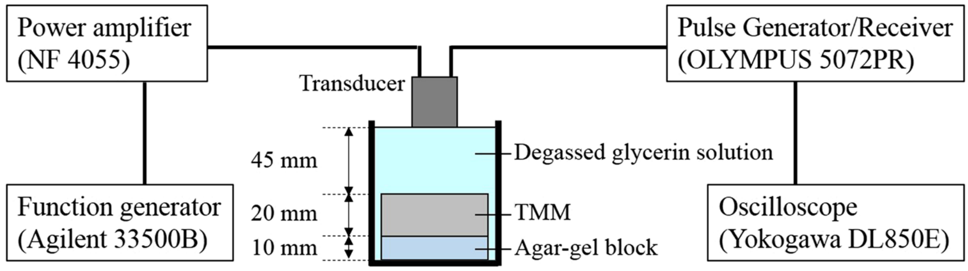

Fig. 4 Experiment system for measurement of temperature rise of TMM phantom due to exposure to ultrasound. Sample heating and sound velocity measurement were carried out using the transducer shown in Fig. 2

Table 1 The initial conditions used in the simulation

Sound velocity (m/s)	Density (kg/m^3)	Acoustic impedance (MRayl)	Attenuation coefficient (dB/cm/MHZ)	Specific heat (kJ/kg/K)	Thermal conductivity (W/m/K)
1540	1050	1.6	0.5	3.8	0.58

near the focus inside the sample was calculated. The experiment system is shown in Fig. 4. Sample heating and sound velocity measurement were carried out using the transducer shown in Fig. 2. The inside of the $100 \times 100 \times 100$-mm acrylic case was filled with an aqueous solution of glycerin of 11% concentration. An agar-gel block and the TMM phantom were placed in the acrylic case. The output signal (5 $V_{P\text{-}P}$, burst period 100 ms, 160,000 cycles) of an oscillator (AGILENT, 33500B) was amplified 20 times using an amplifier (NF Electric Instruments, 4055), and applied to the transducer for heating. Heating and measurement were performed by repeating a cycle of 50 ms of heating and 50 ms of rest 10 times, for a total of 1.00 s. The transducer for measuring sound velocity was driven with a pulse generator, and echo signals from the sample before and after heating were recorded with an oscilloscope. The number of quantization bits was set to 12 bits, and the sampling frequency to 100 MHz.

The simulation of temperature rise in the TMM phantom was carried out by the finite element method to solve a bio-heat transfer equation. This method was described in detail elsewhere [23]. The simulation was carried out in the region of the TMM phantom only. The initial conditions used in the calculation are listed in Table 1 [22]. The distribution of heat generation was calculated by $2\alpha I$, where I is the acoustic intensity in the phantom. The distribution of acoustic intensity was calculated by taking account of attenuation in the phantom from the distribution of normalized magnitude of peak pressure as shown in Fig. 3. The uniform distribution of attenuation coefficient was assumed to be 0.5 dB/cm/MHz [22] in the simulation. The temperature rise was calculated by averaging the values in the sample volume correspondent to the ROI set in the experiments.

Heating by exposure to ultrasound and estimation of thermo-physical properties

First, the temperature coefficient of sound velocity was measured for porcine muscle and fat tissue.

Next, heating was performed by exposure to ultrasound, and the rate of change in sound velocity was measured in the ROI before and after the rise in temperature. The experiment system is shown in Fig. 5. To suppress attenuation, an agar-gel block was placed between the measurement sample and transducer. The measurement samples were porcine muscle tissue and fat tissue with a thickness of 10 mm. At each of three different points, the echo signal was recorded before and after ultrasonic heating. The sample was heated by amplifying the output signal (5 $V_{P\text{-}P}$) of an oscillator (AGILENT, 33500B) 10 times using an amplifier (NF ELECTRONIC INSTRUMENTS 4055),

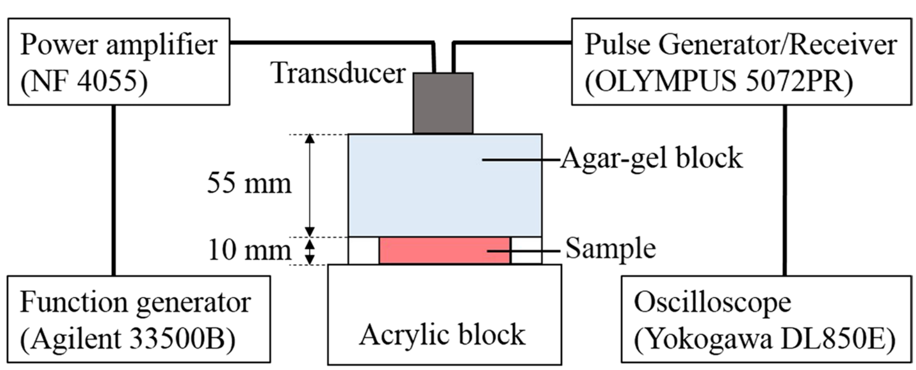

Fig. 5 Experiment system for heating of the tissue sample with exposure to ultrasound, and measurement of the rate of change of sound velocity before and after the exposure

and then applying to the transducer for heating. Impulse voltage was applied to the transducer for sound velocity measurement using a pulse generator, and the echo signals before heating and 100 ms after heating were recorded using an oscilloscope. The number of quantization bits was set to 12 bits, and the sampling frequency to 100 MHz.

Results

Phantom study

The temperature coefficient of sound velocity for the TMM phantom was 1.6 m/s/ °C. The next step was to find the change in sound velocity before and after exposure to ultrasound, as described in the section "Measurement of temperature rise due to exposure to ultrasound by experiments and simulation". The echo signals before and after heating were, respectively, extracted by applying a rectangular window function (time width 1.0 μs), and the echo shift time was calculated using the correlation method. The area around where the echo shift time rose was presumed to be the heat source. Next, the change in the sound velocity Δc was found by taking approximation of $d\tau/dt \cong \Delta\tau/\Delta t$ in Eq. (5) as follows:

$$\Delta c \cong c(x)\frac{\Delta \tau}{\Delta t}, \tag{9}$$

where $\Delta\tau$ is the echo shift time and Δt is the time difference between two gates. Since the peak sound pressure region along the sound propagation direction is the focal region, as shown in Fig. 3, $\Delta\tau/\Delta t$ is calculated by the difference of the echo shift time between the interval of two gates of 7.0 μs corresponding to the 5 mm. The beam width of the focal region was 2 mm, as shown in Fig. 3. Therefore, the ROI was set as the cylindrical region of 2 mm in diameter and 5 mm in length. I_{SPPA} and I_{SPTA} were calculated as 149 W/cm^2 and 74.4 W/cm^2, respectively. These values were calculated from the measured values of sound pressure in water by a hydrophone (ONDA, HNP-0200). The initial temperature was 23.6 °C all in the TMM phantom, the Glycerin solution, and the agar-gel phantom.

The value of the temperature rise in the ROI was estimated from the calculated change in sound velocity and the temperature coefficient of sound velocity (1.6 m/s/ °C). The change over time in the temperature rise value was obtained by repeating the same processing at 100-ms intervals until 1.00 s after heating. Figure 6 shows the plots of temperature rise vs. exposure time for the measurement and simulation results. Sound velocity change was calculated by substituting 1540 m/s to $c(x)$ in Eq. (9) [22].

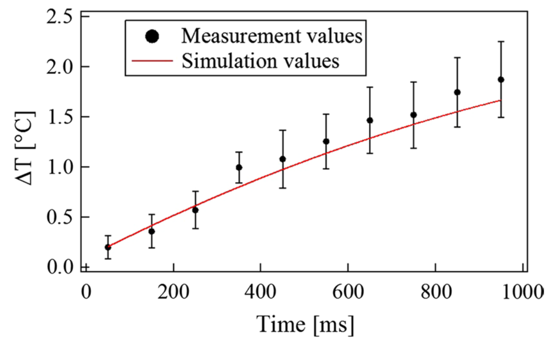

Fig. 6 Relation of temperature rise to elapsed time in TMM phantom due to exposure to ultrasound. Comparison of the measured values with simulated values. Simulation was employed by the finite element method with the bioheat transfer equation

Biological tissue study

First, Fig. 7 shows the results of measuring the temperature coefficient of the sound velocity. Linear approximation was performed for the results, and the slope of the approximation line is shown in Table 2. For muscle tissue, the slope was positive, and for fat tissue it was negative. Also, fat tissue exhibited a stronger temperature dependence than muscle tissue. I_{SPPA} in the muscle tissue and in the fat tissues was calculated as 20.9 W/cm^2 and 28.5 W/cm^2, respectively.

Next, the results for the rate of change in sound velocity before and after ultrasonic heating are shown. Extraction was performed by applying a rectangular window function (time width 1.0 μs) to the measured echo signal before and after heating, and the echo shift time was calculated by the correlation method. The area around where the echo shift time rose was presumed to be the heat source, and the echo shift times for the three measured points and their average were calculated. Figure 8 shows the results of carrying out the same processing for muscle tissue and fat tissue, respectively. Since these tissue samples were not homogeneous like the TMM phantom, the thickness of tissue samples was set at 10 mm less than that of TMM (20 mm). Δt in Eq. (9) was set to 4.0 μs, and $\Delta\tau/\Delta t$ was calculated using the echo shift time obtained by line fitting, as shown in Fig. 8. For the calculation of $\Delta\tau/\Delta t$, strong reflected waves from the front and the back boundaries of the specimen as well as strong echoes due to the tissue non-uniformity need to be avoided. Under these conditions, the sample volume was set at 4.0 μs in time correspondent to 3 mm in length. The $\Delta\tau/\Delta t$ in biological tissue samples was calculated by line fitting of the values of $\Delta\tau$ to the time of gates, since the measured values of $\Delta\tau$ fluctuated more than those of the TMM phantom. The results are shown in Table 3.

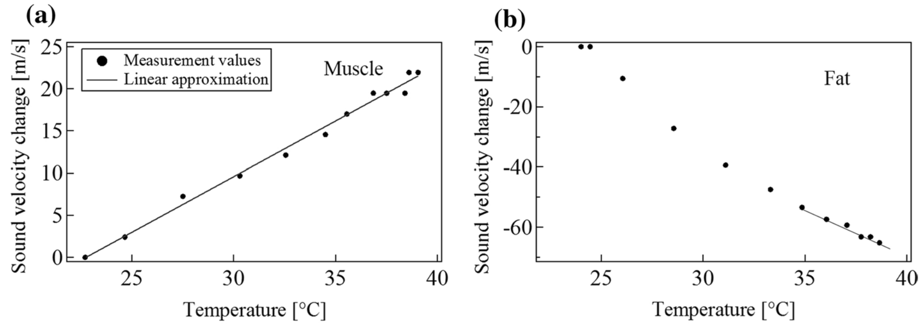

Fig. 7 Relation of sound velocity change to temperature of the three samples. Linear approximation is performed for the results. **a** Porcine muscle, **b** porcine fat

Table 2 Temperature coefficient of sound velocity

Sample	d*c*/d*T* (m/s/ °C)
Muscle	1.2
Fat	− 3.1

Table 4 Temperatusre rises for porcine muscle and fat tissue samples

Sample	$\Delta\tau/\Delta t$	d*c*/d*T* (m/s/ °C)	$(\Delta\tau/\Delta t)\cdot(dc/dT)^{-1}\cdot t_h^{-1}$	Δ*T* (°C)
Muscle	0.27×10^{-3}	1.2	2.3×10^{-3}	0.36
Fat	-2.4×10^{-3}	− 3.1	7.7×10^{-3}	1.1

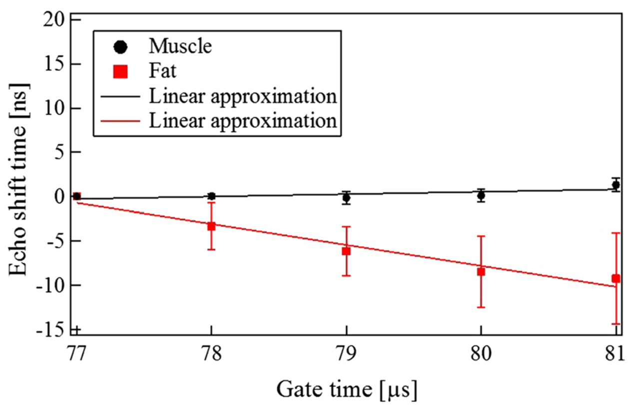

Fig. 8 Relation of the echo shift time to gate time in porcine muscle and fat

Table 3 Experimental results of muscle and fat tissue samples

Sample	Applied voltage (V)	Exposed time (ms)	Echo shift time (ns)	$\Delta\tau/\Delta t$
Muscle	50	100	0.7	0.27×10^{-3}
Fat	50	100	− 9.5	-2.4×10^{-3}

Discussion

In the TMM phantom, the measured value of the temperature rise increased when the heating time elapsed, and was 1.9 °C at 950 ms after the start of heating. In the calculated values of the simulation, there was a temperature rise of 1.7 °C at 950 ms after the start of heating. The measured values of the temperature rise almost matched the simulation results using the finite element method. It is likely that the heating region and measurement region matched due to the use of a transducer with coaxial integration of the transducers for heating and sound velocity. For this reason, it should be possible to use the ultrasonic pulse echo method to measure the temperature rise (1.5 °C or less) of a sample due to ultrasonic heating for a short time (1 s or less).

From Table 3, it was found that the rate of change of sound velocity in porcine muscle tissue was positive, and the rate of change of sound velocity in fat tissue was negative. However, the slope for fat tissue was always far smaller than the slope for muscle tissue. This likely occurred due to the difference in heat capacity. The fact that there is a difference in the rate of change of sound velocity between ultrasonically heated muscle tissue and fat tissue suggests the possibility of tissue identification using the rate of change in sound velocity. Also, the value of the temperature rise of tissue was estimated by the results in Table 2 and Table 3. Sound velocity change for muscle tissue and fat tissue was calculated by substituting 1580 m/s and 1454 m/s with $c(x)$ in Eq. (9), respectively [24].

Those results are shown in Table 4. The value of the temperature rise for muscle tissue was 0.36 °C, and the value of the temperature rise for fat tissue was 1.1 °C. This shows that the temperature rise due to ultrasonic heating was within the range of the safety standard (1.5 °C).

To stabilize the specimen during the exposure time, it was fixed on an acrylic block. However, the reflected waves from the surface of the acrylic block affect the intensity of the ultrasound in the specimen. Therefore, stochastic fluctuation of the measured values was observed, as shown in Fig. 8. To reduce the fluctuation, values measured at three different positions were averaged.

The thermal conductivity of the acrylic block and fat and muscle tissues is 0.21, 0.21, and 0.49 W/km, respectively. The value of thermal conductivity of the acrylic block is the same as that of the fat tissue. For the muscle tissue, the values of temperature rise measured by the experiments on the acrylic block are considered to be the same values as those for the fat tissue. Further studies are required to investigate this issue.

It is difficult to correctly measure the thermo-physical properties of biological tissue containing gas bodies such as microbubbles, because the thermo-physical properties of the gas are quite different from those of the biological tissue. Further research using microbubbles is required to resolve this issue.

Conclusion

The change over time in the temperature rise due to exposure to ultrasound was measured using a transducer in which the transducers for heating and measurement were coaxially integrated. Also, the temperature rise of samples exposed to ultrasound was theoretically analyzed using the bioheat transfer equation. The results showed that the two almost matched. The effectiveness of estimating thermo-physical properties using the rate of change of sound velocity due to ultrasonic heating was examined for porcine tissues. As a result, it was shown to be possible to carry out in vivo measurement of thermo-physical properties within the safe range, and within a time during which movement of the living body can be ignored. Further investigation is required to apply the proposed method to biological tissues including gas bodies such as microbubbles.

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Compliance with ethical standards

Conflict of interest Yukako Tsujimoto, Mai Morimoto, Naotaka Nitta, and Iwaki Akiyama declare that they have no conflicts of interest.

Human and animal rights This article does not contain any studies with human or animal subjects performed by any of the authors.

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