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Magnetic Resonance Imaging (MRI) Compatible Ultrasound Therapeutic System

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Since an ultrasound (US) imaging system can image in real-time and interactively, it can be used as an image guidance assisting magnetic resonance imaging (MRI) for minimally invasive therapy. The effects of MRI-compatible US probes on MRI monitoring were evaluated, and it was found that the MRI-compatible US probes, whose backing material contained 100 ppm ferrite, did not disturb MR monitoring except at a few mm radius from the US probe's position. MRI temperature monitoring of a swine liver irradiated with a high-intensity focused US beam from an MRI-compatible therapeutic transducer with US image guidance was then performed, and the potential usefulness of such a therapeutic system in minimally invasive therapy was demonstrated. [DOI: 10.1143/JJAP.41.3579]

KEYWORDS: minimally invasive therapy, MRI compatible, HIFU, temperature mapping

1. Introduction

Minimally invasive therapy (MIT) is performed through no more than a few tiny incisions. Since it is expected that MIT affords patients the least possible physical trauma, there is a strong demand for such therapy. The surgeon's view of the area to be operated on, through tiny incisions on the surface of body with the naked eye, is so limited during MIT that image guidance is necessary for performing MIT safely. Interactivity with surgical operation is required for such image guidance for minimal surgery time. Optical scopes have been widely used for this purpose, but only the surfaces of organs can be observed in this manner.

Magnetic resonance imaging (MRI) is a modality suitable for such image guidance because it can discriminate between tissues with a multiparametric imaging capability.[1)] The invention of an open-configuration MRI system, which allows direct access to the patient during the treatment procedure, has made MRI-guided MIT possible.[2)] However, the imaging speed of MRI, typically a few seconds per image slice, suffer the interactivity. Ultrasound (US) imaging can performed at a much faster rate of 20–30 ms per image slice, while its field of view is limited to about 10 cm from the US probe. For the reasons mentioned above, US imaging and MRI are complementary in terms of imaging speed and field of view. Accordingly, we developed a US probe that can be fitted in an MRI gantry to produce a therapeutic system with two complementary image-guidance modalities.

Among the several modalities of noninvasive therapy, high-intensity focused ultrasound (HIFU)[3)] is one of the most promising. In a HIFU treatment, tissue temperature monitoring is important for heating the target tissue efficiently while avoiding normal tissue damage. Using the chemical shift of water protons, MRI can map tissue temperature without contact.[4–6)] Hynynen *et al.* demonstrated the usefulness of MRI monitoring for HIFU treatment.[7)] In their experiment,[7)] the target of HIFU was adjusted by MRI, however, US images are more suitable than MRI for such adjustment for the following two reasons. Firstly, the fast imaging rate of US makes it possible to adjust the position of a HIFU transducer including a US imaging probe and its focal zone to the target tissue interactively. Secondly, since the same physical phenomenon, i.e., propagation of an acoustic wave, is used in both therapy and imaging, the spatial discrepancy between the imaged target and the therapeutic focus should be minimally even if the diffraction slightly affects the wave propagation in the body. With these two reasons in mind, we developed an MRI-compatible US probe and used it in an MRI-compatible therapeutic system.

2. Materials and Design

The US probe used in the MRI gantry must satisfy the following requirements: the materials must not disturb the static magnetic field of the MRI, and the structure must not generate eddy currents that disturb the dynamic magnetic field of the MRI. After evaluating several parts and materials of conventional US probes, we found that the backing material disturbed the MR images. The attenuation coefficient and acoustic impedance of the backing material must be chosen such that it can perform its roles: to improve the pulse response of the transducer and to reduce the effect of multiple reflections in the transducer. In order to obtain these two properties, a compound of polymer and metal or metal oxide powder is used as the most popular backing material. Ferrite is popular as such a powder material.

We took two approaches in changing the backing material for the development of our MRI-compatible US probe. First, we replaced ferrite with a compound of similar physical properties: zirconia. The densities of barium ferrite and zirconia are 5.2 g/cm^3 and 5.9 g/cm^3, respectively. Next, we replaced ferrite with heavy particles, in order to reduce the amount of powder used. The overall amount of magnetic impurities in the probe can be reduced by using heavy particles as an acoustic mass, assuming that the molar content of magnetic impurities in a powder does not change depending on the kind of powder. We chose tungsten as the powder material, whose density is 19.3 g/cm^3, since it is the heaviest element with the exception of radioactive elements and noble metals.

3. Methods

3.1 Evaluation of MRI-compatibility

The disturbance from the US probe on MRI images can be evaluated by observing the difference between the images of a uniform phantom with and without the US probe. A water phantom, which contains a small amount of gadolinium, was used as such a phantom. Considering the applicability to MIT, an open MRI system at 0.3 mT (AIRIS® II, Hitachi Medical Corp.) was chosen for the experiment and the effects of US probes were evaluated in its gantry. In order to evaluate the distortion of MR images affected by US probes more quantitatively, the phase difference between the normal spin-echo images and the echo-time-shifted spin-echo images was also measured. A high-field MRI system at 1.5 T (STRATIS® II, Hitachi Medical Corp.) was chosen for higher accuracy of the measurement.

The phase shift, $\Delta\phi_{MD}$, due to magnetic distortion is represented as follows:

$$\Delta\phi_{MD} = \gamma \Delta H \epsilon, \tag{1}$$

where γ is the gyromagnetic ratio of protons, ΔH is the magnetic field increment due to distortion, and ϵ is the echo time shift. In this work, $\epsilon = 10$ ms and $\gamma = 42.58$ MHz/T. In MRI, a gradient magnetic field, Gr, is applied to specify the position where MR signals are generated, based on the principle that the MR frequency is proportional to the static magnetic field. Therefore, ΔH results in the shift of the image position, ΔX, as

$$\Delta X = \frac{\Delta H}{Gr}. \tag{2}$$

According to above equation, the maximum amount of distortion of MR images and the distorted area were measured.

Tissue temperature monitoring by MRI should be carried out under minimally magnetic distortion because the phase shift caused by a change in temperature is small. The change in temperature, ΔT, was calculated from the phase difference, $\Delta\phi_T$, between protons at different temperatures after echo time (T_E) according to the following equation:

$$\Delta T = \frac{\Delta\phi_T}{2\pi T_E \gamma H C}, \tag{3}$$

where T_E is the echo time, and C is the temperature-dependent coefficient of the water proton chemical shift, -0.01 ppm/°C.[8] Substituting $\Delta\phi_{MD}$ for $\Delta\phi_T$ in eq. (3) gives the equivalent temperature from the distortion of the magnetic field.

3.2 Influence on US images

The influence on the US images of replacing the backing material was evaluated by taking US images with a diagnostic US system (EUB-6000, Hitachi Medical Corp.) and also by measuring the round-trip pulsewidth. The experimental arrangement is shown in Fig. 1. The reflector, consisting of an aluminum plate, was placed 5 cm from the US probe. The probe was driven by a pulser-receiver (5900PR, Panametrics) and the data was acquired by a digital oscilloscope (TDS3034, Tektoronics).

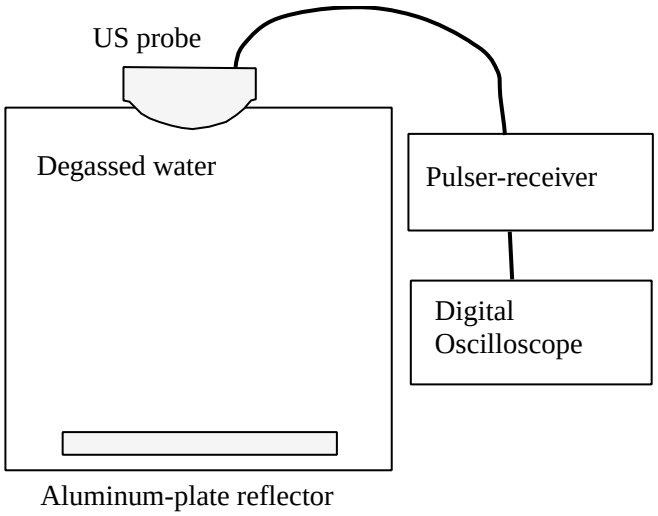

Fig. 1. Setup for measurement of the round-trip pulsewidth.

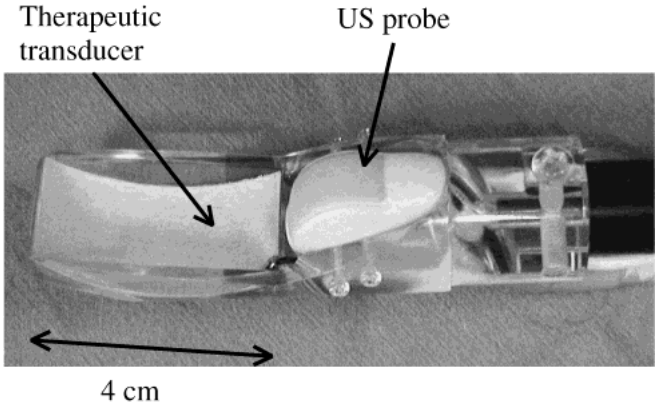

Fig. 2. MRI-compatible US therapeutic applicator.

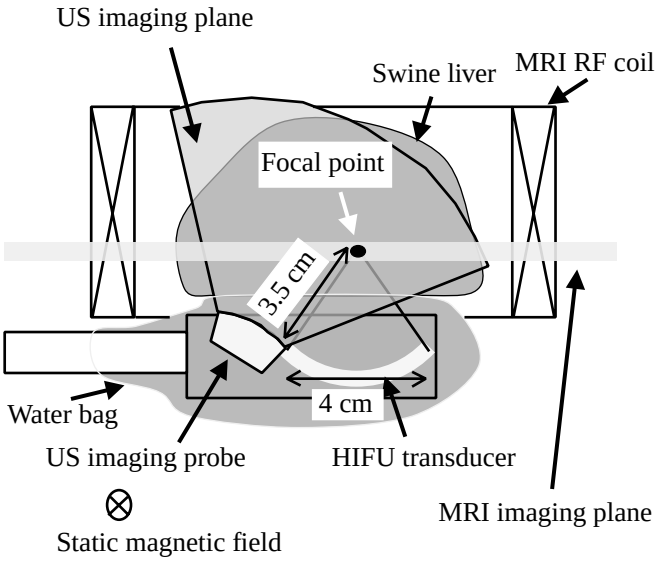

Fig. 3. Setup for MRI temperature measurement of tissue heated with MRI-compatible US applicator.

3.3 MRI temperature measurement of tissue heated with MRI-compatible US applicator

Figure 2 shows an MRI-compatible applicator, consisting

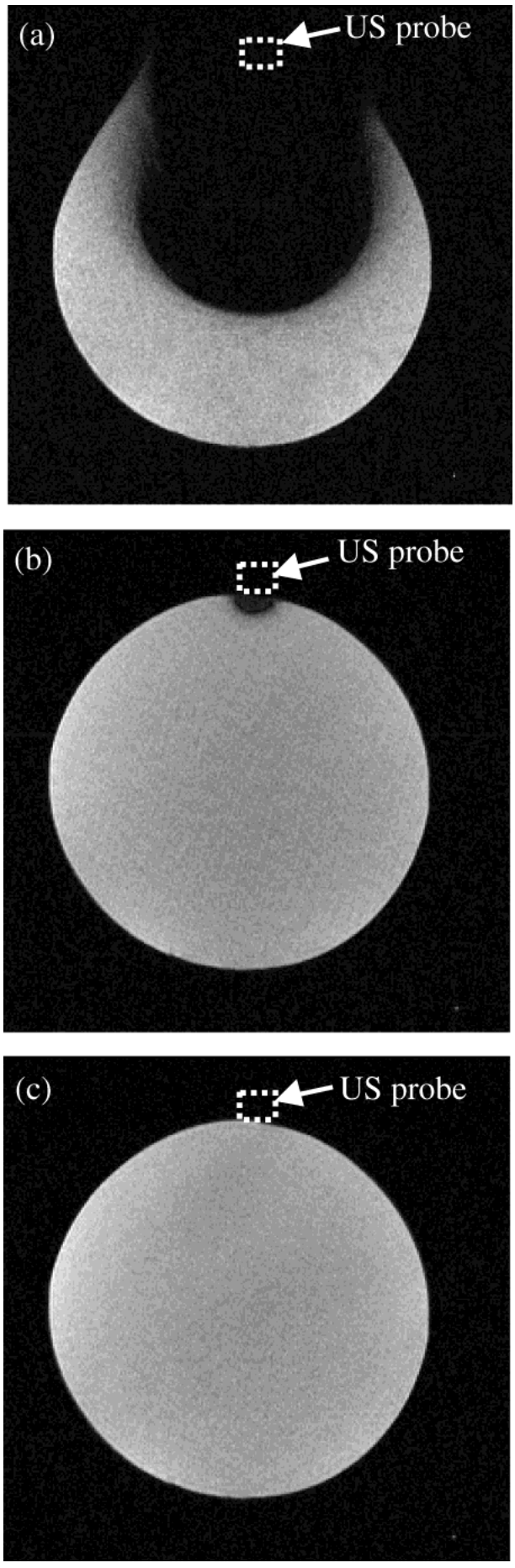

Fig. 4. Effects of MRI-compatible US probes on MR images. (a) Conventional US probe. (b) Zirconia US probe. (c) Tungsten US probe.

of an MRI-compatible US probe and a high-power HIFU transducer in an MRI-compatible housing. The black part of the housing is made of polyphenyleneoxide resin, and the transparent part is made of polymethlymetacrylate resin. The HIFU transducer consists of a lead zirconium titanate(PZT) ceramic (C213, Fuji Ceramics) with a spherical curvature radius of 35 mm. The MRI-compatible applicator was set in a bag filled with degassed water, attached to a block of swine liver, and placed in an RF coil (MR-QWC-61, Hitachi Medical Corp.), as shown in Fig. 3. The HIFU transducer was driven at 3.25 MHz for 4 s, at a voltage corresponding to the peak ultrasonic intensity in water of 2.8 kW/cm^2.

4. Results and Discussion

4.1 Evaluation of MRI-compatibility

Figure 4 shows the effect of the MRI-compatible US probe on MR images taken by AIRIS® II. Figure 4(a) shows that a conventional US probe disturbed the region within 10 cm from the probe, Fig. 4(b) shows that a zirconia US probe disturbed the region within 1 cm from the probe, and Fig. 4(c) shows that a tungsten US probe disturbed the region within only 2 mm from the probe surface. Figure 5 explains the reason for the conventional US probe disturbing the MR image. When there is a disturbing material (the conventional US probe in this case) in the imaging area, the gradient magnetic field is distorted. If this distortion is so large that the frequency of the received MR signal is out of the MRI's re-

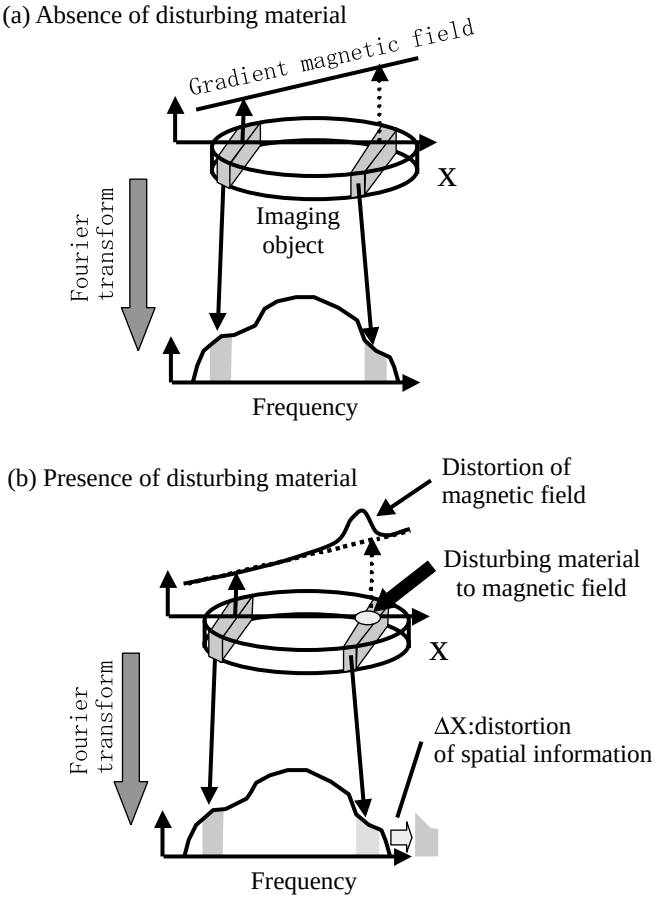

Fig. 5. Explanation for distortion of MR image caused by conventional US probe.

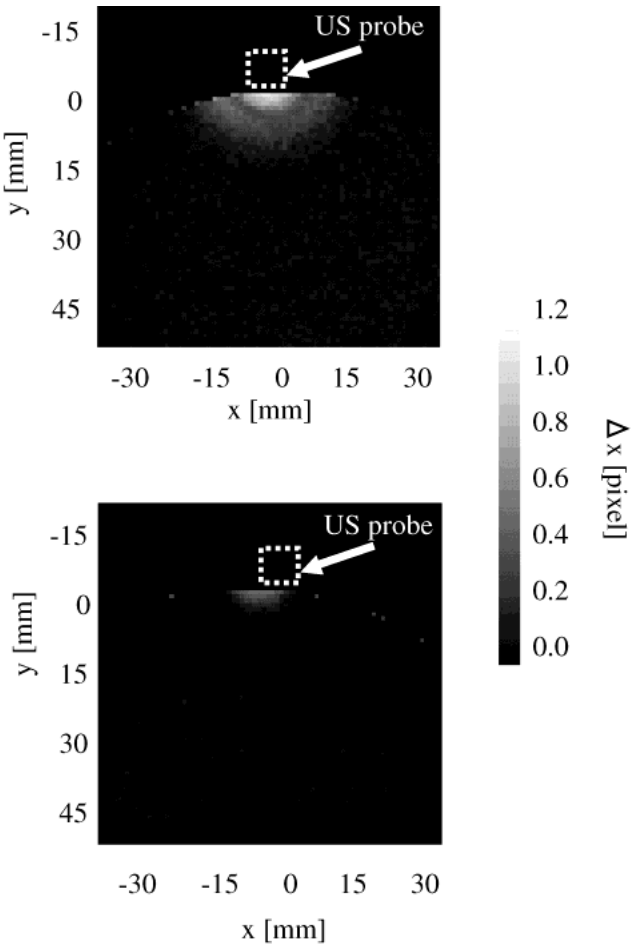

Fig. 6. Space-distortion images caused by (a) zirconia US probe (b) tungsten US probe.

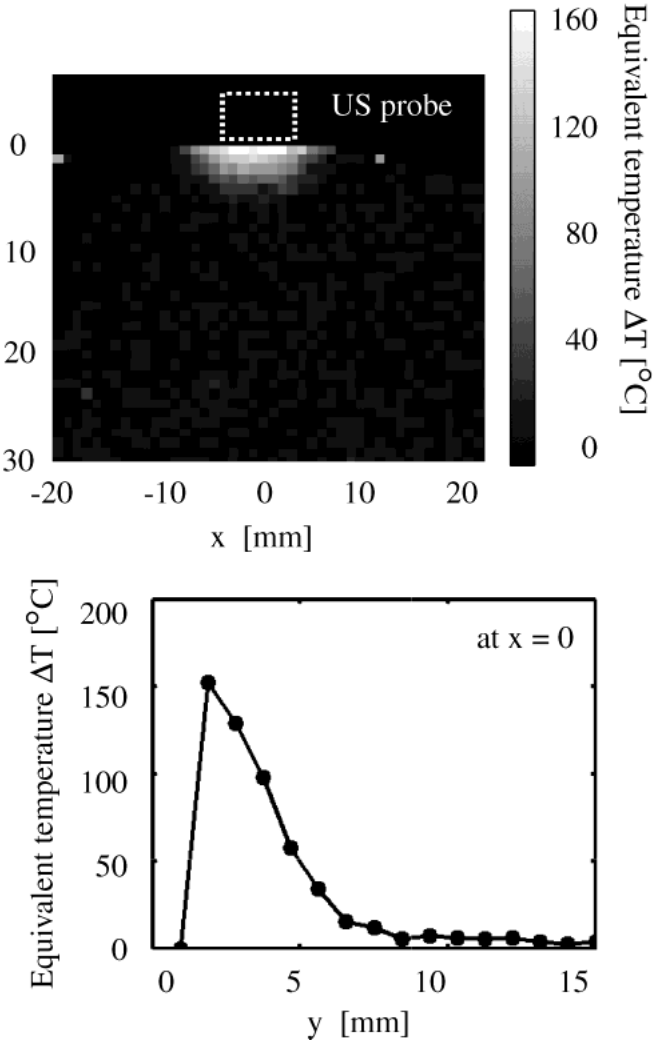

Fig. 7. Equivalent temperature change caused by the tungsten US probe.

ceiving bandwidth, the image signal from that area is lost.

Since MRI is performed in most cases to examine parts of the body much deeper than the surface, the tungsten probe can be clinically used as an MRI-compatible US probe.

Figure 6 shows a map of the distortion of an MR image taken by high-field MRI; STRATIS® II. The color bar represents ΔX per pixel of the MR image. While in the case of zirconia, the maximum distortion of the image is slightly larger than a pixel (0.94 mm in this case), in the case of tungsten, it is less than a pixel. These results are consistent with the MR images in Fig. 4 at 0.3 T.

The graph in Fig. 7 shows equivalent temperature converted from the distortion of the magnetic field in the case of tungsten. In the region close to the US probe, the phase shift caused by the probe is much larger than the phase shift caused by the temperature change from 10 to 50°C. However, the magnitude of such phase shift rapidly decreases with the distance from the probe and becomes negligible at a distance of 10 mm.

This level of phase shift will not disturb temperature monitoring as long as the treatment area is a certain distance from the applicator. Since the typical focal length of HIFU at 3–4 MHz ranges from 25 to 50 mm, the tungsten probe can be used as the MRI-compatible US probe in a US-aiming, MRI-monitored HIFU treatment system.

The difference between the zirconia and tungsten probes in the images taken by the MRI system was caused by the difference in the amount of magnetic impurities in the probes. In spite of the fact that the molar contents of magnetic impurities in the powders were similar, the magnetic impurities per weight were different, due to the difference in their density. The zirconia powder contained 0.04% and the tungsten powder contained 0.015% ferrite per weight. For this reason, in spite of the fact that the difference in mass between the powders in each backing material was small, the total amount of ferrite in the zirconia probe was three times more than that in the tungsten probe, i.e., the zirconia backing material contained 300 ppm and the tungsten backing material contained 100 ppm ferrite per weight.

It may be useful to estimate the amount of magnetic substances in the human body for setting the final goal of the magnetic impurity level in this approach. The total amount of magnetic substances in the blood of the human body is about 50 mg/kg, 50 ppm.[9] Accordingly, the fluctuating amount of magnetic content in the human body, which influences MR images, is estimated to be 5 ppm, assuming that the fluctuation caused by blood flow is about 10%. Therefore, it would be worth pursuing further a reduction of the magnetic impurity level in MRI-compatible US probes, up to an order of magnitude.

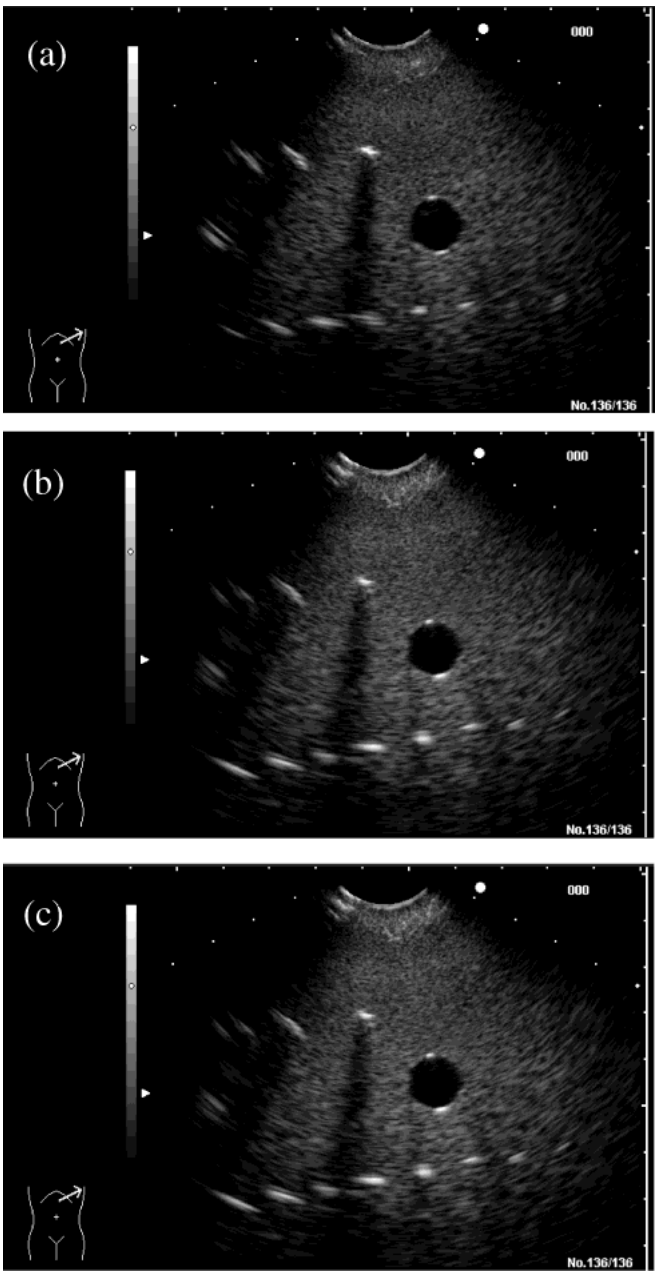

Fig. 8. B-mode images of US phantom taken by (a) conventional US probe, (b) zirconia US probe and (c) tungsten US probe.

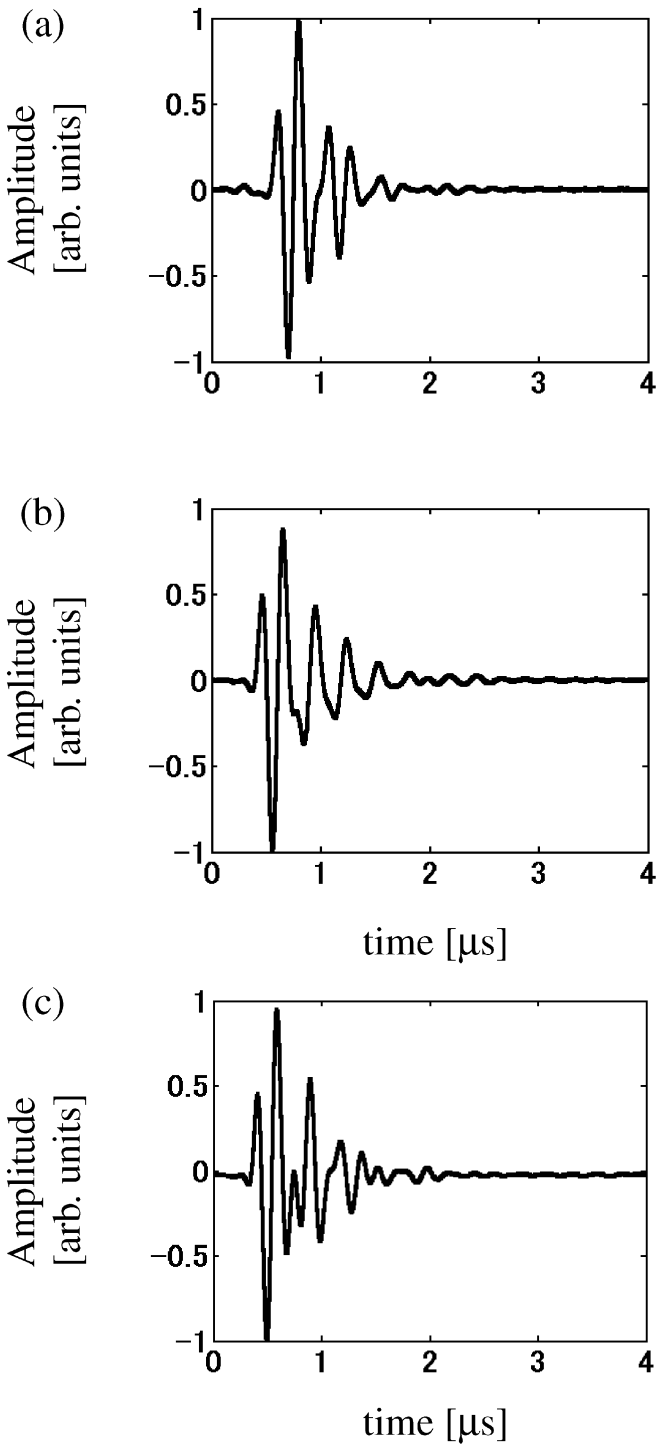

Fig. 9. Pulse responses of (a) conventional US probe, (b) zirconia US probe and (c) tungsten US probe.

4.2 Influence on US images

Figure 8 shows three phantom images taken using three different US probes. All three were identical except for the backing material. The first probe used the conventional backing material, the second one used zirconia, and the third one used tungsten. As shown in these images, the alteration of the backing material did not affect the speckle pattern, the darkness in the cyst-mimicking hole, or the point-reflector shape. The pulse response of each probe is compared in Fig. 9. While the round-trip pulsewidth with the zirconia probe is almost the same as that with the conventional probe, that with the tungsten probe is 1.2 times longer than the others.

These differences may have been caused by a slight difference in acoustic impedance between the tungsten backing material and the others, but they did not seem to be so large as to give significant influence on the US images. The above-described MRI and US evaluations of the probes confirmed the potential usefulness of the proposed MRI-compatible US probe in minimally invasive therapy.

4.3 MRI temperature measurement of tissue heated with MRI-compatible US applicator

Figure 10 shows a temperature map of the liver immediately after a HIFU irradiation. An increase of temperature of about 20°C at the focal point of the US applicator can be clearly observed.

In coagulation treatment with HIFU, the temperature of the tissue to be treated is increased above the coagulation temperature, typically 50–60°C for a very short period of time on

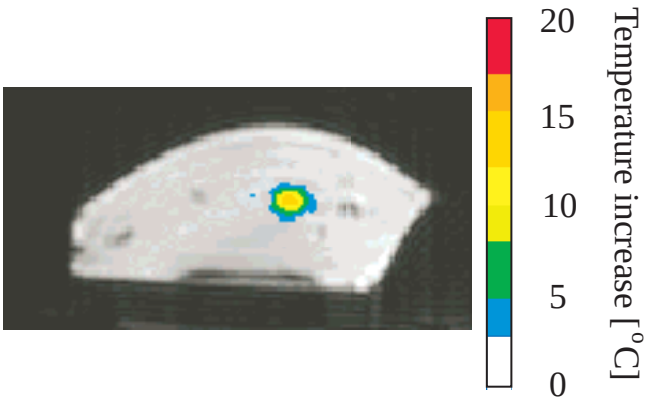

Fig. 10. Temperature map after radiation of HIFU.

the order of several seconds.[10] This corresponds to a tissue temperature elevation of more than about 20°C. The tissue temperature monitoring demonstrated here has sufficient accuracy for such cases. In hyperthemia, on the other hand, the temperature of the tissue to be treated is increased by only several degrees for a very long period of time on the order of an hour. Thus, the amount of magnetic impurity in the US probe must be further reduced to monitor hyperthemia.

5. Conclusions

MRI-compatible US probes, whose backing materials were changed from ferrite to zirconia or tungsten powder, were developed. We evaluated the effects of these US probes on MR images, including their use in tissue temperature measurement, and the influence of material alteration on US images. The newly developed US probe, whose backing material contained 100 ppm ferrite per weight, did not disturb MR monitoring in the region more than a few mm away from the US probe. It can be concluded that we should reduce the amount of magnetic impurity in a US probe to the order of 5 ppm per weight to ensure that it is MR-compatible. Since US imaging and MRI are complementary, we can construct a double-image-guided minimally invasive therapeutic system using the proposed MRI-compatible US probe.

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