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MR Thermometry Near Metallic Devices Using Multi-Spectral Imaging

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Abstract

Purpose—The lack of a technique for MR thermometry near metal excludes a growing patient population from promising treatments such as MR-guided focused ultrasound therapy. Here, we explore the feasibility of multi-spectral imaging (MSI) for non-invasive temperature measurement in the presence of strong field inhomogeneities by exploiting the temperature dependency of the T_1 relaxation time.

Methods—A 2D inversion-recovery-prepared MSI pulse sequence (2DMSI) was implemented for artifact-reduced T_1 mapping near metal. A series of T_1 maps was acquired in a metallic implant phantom while increasing the phantom temperature. The measured change in T_1 was analyzed with respect to the phantom temperature. For comparison, proton resonance frequency shift (PRFS) thermometry was performed.

Results—2DMSI achieved artifact-reduced, single-slice T_1 mapping in the presence of strong off-resonance with a spatial resolution of 1.9 mm in-plane and a temporal resolution of 5 min. The maps enabled temperature measurements over a range of 30 °C with an uncertainty below 1.4 °C. The quality of the resulting temperature maps was independent of the distance from the metal, whereas the PRFS-based temperature measurements were increasingly impaired with increasing off-resonance.

Conclusion—We demonstrated the ability to noninvasively measure temperature near metal using MSI and the T_1 temperature sensitivity.

Keywords

multi spectral imaging; thermometry; temperature; proton resonance frequency shift; T_1 mapping; distortion correction; metallic implants; metal artifacts; metal-induced distortions; susceptibility artifact

Introduction

MR thermometry [1, 2] provides noninvasive temperature monitoring and thus ensures safety and efficacy in MR-guided focused ultrasound (MRgFUS) treatments. Clinical applications of MRgFUS include regional hyperthermia for chemotherapy sensitization [3]

or thermal ablation for tumor control in uterine fibroids as well as breast, prostate and bone cancer [4]. For the latter, MRgFUS has also been established as an effective approach for pain palliation in many of the most common sites of metastases for patients who have failed standard treatment by thermal ablating the tissue at the bone-soft tissue interface of the bone metastases [5]. However, so far there exists no technique for MR thermometry near metallic devices such as implants or surgical clips, which means that a growing patient population is excluded from MRgFUS treatments. Based on approximately half a million new cases of metastatic bone cancer developed in the US annually [6, 7] and the frequent use of metallic devices to stabilize bone at the site of the cancer, we estimate that MRgFUS treatment of bone metastases is currently impractical in tens of thousands of candidates. A technique for MR thermometry near metallic devices would also serve as a tool to evaluate thermal characteristics of devices such as total joint replacements during MRI. Although implants are frequently scanned when MRI is indicated [8], heating of the devices from induced radio-frequency currents is still a major safety concern [9, 10].

MR thermometry commonly exploits the temperature sensitivity of the proton resonance frequency. The proton resonance frequency is determined by the local magnetic field that protons experience, which depends on the screening of the macroscopic field by the electrons of the water molecule. For pure water, the screening increases approximately linearly with increasing temperature over a wide range of temperatures, from -15°C to 100°C [11]. In proton resonance frequency shift (PRFS) thermometry, the temperature-induced frequency shift is determined from the resulting phase change in gradient-recalled echo (GRE) images and is converted to a temperature change using the linear relationship [1, 12]. A number of other MR tissue properties including the relaxation times T_1 and T_2 or the proton density show a linear temperature dependency [2], and thus potentially offer a variety of alternative thermometry mechanisms. The proton density is expected to vary linearly over a wide temperature range [13], whereas the exponential temperature dependency of T_1 and T_2 can be approximated by a linear relationship over a narrow temperature range [14]. As the relaxation times also depend on the tissue microstructure, thermal coagulation leads to additional nonlinear and irreversible changes in the temperature behavior [15]. Whereas the PRFS is expected to provide the highest temperature sensitivity in aqueous tissues [16], the temperature sensitivity of the T_1 relaxation time is $1\text{--}2\text{ }^{\circ}\text{C}^{-1}$ [17], still high compared to that of the proton density ($0.3\text{ }^{\circ}\text{C}^{-1}$) [13] or the T_2 relaxation time ($<0.2\text{ }^{\circ}\text{C}^{-1}$) [17]. The T_1 and T_2 relaxation times and the proton density can be assessed with standard spin echo (SE) techniques.

MR thermometry near metallic devices is challenging. It requires a) a technique for imaging near metal and b) a thermometry mechanism that is accessible with this technique. Metal-induced field inhomogeneities cause both in-plane and through plane distortion artifacts [18]. For GRE-based imaging techniques the impairment is even more severe, as field inhomogeneities cause signal dephasing, which can lead to complete signal loss. Existing multi-spectral imaging (MSI) approaches such as SEMAC [19], MAVRIC [20] and hybrid methods [21] generally suppress most of these artifacts near metallic devices. They are SE based and resolve slice distortions by additional encoding along the slice dimension. So far, the prolonged acquisition time resulting from three-dimensional encoding has limited the flexibility of these MSI techniques. However, a recently introduced two-dimensional MSI

approach (2DMSI) offers the potential for artifact-reduced imaging near metal with a significantly shorter acquisition time [22]. Without the need for three-dimensional encoding, 2DMSI allows faster data acquisition by reducing the number of acquired slices.

In this work, we propose the use of 2DMSI for non-invasive temperature measurement near metallic devices (MSI thermometry), by exploiting the temperature dependency of the T_1 relaxation time. Artifact-reduced T_1 mapping in close proximity to the metal is achieved by a 2DMSI approach with inversion recovery (IR) preparation. We demonstrate the feasibility of MSI thermometry in a phantom experiment and compare this technique to the commonly used PRFS thermometry approach.

Methods

Experimental Setup

Imaging was performed on a 3T whole-body MRI system equipped with a single channel transmit-receive head coil (GE Healthcare, Milwaukee, WI, USA). The coil was loaded with a dedicated phantom, containing a total shoulder replacement implant (Biomet, Warsaw, IN, USA) embedded in agarose gel (Bacto Agar, 2% w/v, Becton, Dickinson and Company, MD, USA). Doping of the agarose gel with CuSO_4 (Copper Sulfate Solution, 1.0 mmol/l, Fisher Scientific Company, VA, USA) resulted in a T_1 of 697 $\pm$ 16 ms ($T_{1\text{Base}}$) at a room temperature of 19.4 °C. The implant consisted of a titanium humeral stem and a cobalt-chromium-molybdenum alloy humeral head. Plastic tubes were embedded in the agarose gel on two sides of the phantom and, containing water at 79 °C, served as a heating source to build up a temperature gradient through the phantom. Four fluoroptic temperature probes with a precision of 0.1 °C (LumaSense Technologies, Santa Clara, CA, USA) were placed at distances between 4 mm and 44 mm from the implant to monitor the phantom temperature. In addition, four oil-filled reference samples were placed in proximity to each corner of the phantom for monitoring the variation of the background phase during PRFS thermometry. The entire setup was insulated with towels to maximize heating efficiency.

2DMSI

2DMSI is a single-shot fast spin echo technique that enables artifact-reduced imaging near metallic devices by excitation of finite spectral and spatial regions that can be imaged with minimal artifact [22]. Figure 1a shows the 2DMSI pulse sequence diagram. For a given slice, the inverted amplitude of the slice selection gradient between excitation and refocusing pulses results in the selection of a “frequency bin”. The bandwidth of the excitation and refocusing pulses limits the spectral distribution of the spins during excitation and refocusing and thus minimizes dephasing during these processes. The resulting spatial overlap of the excitation and the refocusing profile limits the region of selected spins to a region with limited field inhomogeneities also in the absence of the slice selection gradient. This reduces distortions of the read encoding gradient and, further supported by an encoding of the SE train with a high readout bandwidth, avoids the need for view angle tilting [23]. Figure 1b demonstrates excitation and refocusing profiles, as well as the spatial profile of the frequency bin determined by the overlap. Repetition of the acquisition with different radio frequency modulations results in imaging of adjacent frequency bins. Finally, the composite

image is built up by root-sum-of-squares combination of the individual bin images (Fig. 1c). Through this approach, the 2DMSI technique avoids the need for additional time-consuming phase encoding along the slice dimension [19]. The acquisition time is proportional to the number of acquired frequency bins and slices.

T₁-based MSI Thermometry

For T₁-based MSI thermometry, IR-prepared 2DMSI images with varying inversion times (TI) were acquired during heating of the phantom. A series of T₁ maps were retrospectively reconstructed from the images, each representing a different temporal phase during the heating process. To determine the temperature sensitivity, the measured change in T₁ was correlated to the temperature variation provided by the fluoroptic temperature probes. Based on this calibration, the T₁ maps were converted to temperature maps.

For each TI, a set of 12 frequency bins was acquired with a 4500 ms repetition time, 30 ms echo time, 240 mm × 240 mm field-of-view, 3 mm slice thickness, 128 × 128 matrix size (70 phase encodes after half-Fourier) and 1.9 kHz/pixel readout bandwidth, resulting in an acquisition time of 54 sec per IR-2DMSI image. The excitation bandwidth for each frequency bin was 1.3 kHz (full-width-half-maximum; SLR pulse, time-bandwidth product: 1.5, duration: 1.2 ms) and the bandwidth of the slice-selective adiabatic inversion was 1.2 kHz. The slice widths of the excitation, refocusing and inversion pulse were matched. The bin center frequencies were spaced by 0.9 kHz, ranging from −4.5 kHz to +5.4 kHz. The spacing was determined experimentally to achieve an optimized composite slice profile. In total, IR-2DMSI images for five different TIs varying between 50 ms and 4000 ms were acquired, resulting in an acquisition time of about 5 min per T₁ map. The acquired TI values corresponded to 0.1 T_{1Base}, 0.7 T_{1Base}, 1.4 T_{1Base}, 2.1 T_{1Base} and 5.7 T_{1Base} and were optimized to cover T₁ values between 1 T_{1Base} and 3 T_{1Base}. During a heating period of about 400 min, data for a total of 18 T₁ maps were acquired, roughly each time after a temperature change of about 5 °C.

Data analysis was performed in Matlab (MathWorks, Nattick, MA, USA). For each T₁ map, the signal variation over TI was extracted from the local-bin images (magnitude data) on a voxel-by-voxel basis and fitted for T₁ using a four-parameter model also accounting for flip-angle imperfections [24]. The local bin is defined as the frequency bin providing the highest image intensity for a given voxel, similar to a maximum intensity projection along the bin direction. It was preferred to the combination of all frequency bins due to reduced noise bias in the corresponding signal magnitude courses. For signal-to-noise ratio (SNR) quantification, a SNR map was calculated from the IR-2DMSI image with longest TI by dividing the voxel signal by the image noise. The latter was determined from the magnitude image background and the Rayleigh variance was converted to a Gaussian variance. A region-of-interest (ROI) of 3 × 3 voxels (corresponding to an area of 5.6 mm × 5.6 mm) was placed at each temperature probe location and both the mean T₁ value and its variation (standard deviation) were determined. For each ROI, the mean T₁ value was plotted over the temperature provided by the corresponding temperature probe and a linear regression was applied to fit the slope, yielding the temperature sensitivity m. The fit data were also used to evaluate the uncertainty with which the T₁-based MSI thermometry technique was able to

detect T₁ changes and correspondingly temperature changes. As a measure for the thermometry uncertainty ΔT , the standard deviation of the differences between the temperature values provided by the temperature probe and the temperature values provided by the fit was determined. Multiplying the thermometry uncertainty with the temperature sensitivity yielded the corresponding T₁ uncertainty ΔT_1 . Finally, the T₁ maps were converted to absolute temperature maps using the average temperature sensitivity calculated from the four ROIs and by adding the room temperature (19.4 °C) at which the baseline T₁ map was acquired.

PRFS Thermometry

For PRFS thermometry, a radio-frequency-spoiled GRE image of the same slice was acquired after each T₁ map (repetition time: 100 ms, echo time: 4.2 ms, flip angle: 36°, field-of-view: 240 mm × 240 mm, slice thickness: 3 mm, matrix: 128 × 128, excitation bandwidth: 1.3 kHz, readout bandwidth: 0.5 kHz/pixel, acquisition time: 16 sec). The echo time was kept shorter as the echo times usually used for PRFS and represents a tradeoff between PRFS sensitivity and signal loss due to dephasing. The phase change $\Delta\Phi$ during heating was determined from the complex images on a voxel-by-voxel basis and converted to a change in electron screening (Δs) according to

$$\Delta s = - \frac{\Delta\Phi}{T_{Echo} \times \gamma \times B_0} \quad [1]$$

as described in [1]. The variable T_{Echo} denotes the echo time, γ the gyromagnetic ratio and B_0 the static magnetic field. Note that here Δs combines both temperature-induced changes of the actual electron screening and temperature-induced changes of the susceptibility. Both are expected to increase linearly with increasing temperature [12]. The measured Δs might also include temporal variations of the background phase. To compensate for this, the latter was determined by averaging Δs over the oil-filled reference samples and subtracted from the Δs maps. Within oil, temperature does not influence the PRFS due to the absence of hydrogen bonds [11]. The subsequent analysis of the temperature-induced phase change, as well as the conversion of the Δs maps into temperature maps was performed using the same methods as for the T₁-based MSI thermometry (including the same ROIs).

Results

Imaging Near Metal

Figure 2a shows a set of 12 bin images for a given TI, demonstrating the various frequency components within the phantom. The red line indicates the boundary of the metallic implant and was determined from the composite IR-2DMSI image. The on-resonant bin (0.0 kHz) covers most parts of the phantom up to the lower lateral part of the humeral stem. The structures visible next to the right edge of the phantom are part of the heat source tubing. The four circular areas represent the oil-filled reference samples for background phase monitoring. The off-resonance bins reveal the areas closer to the implant, which were not covered by the on-resonant bin. Overall, the implant-induced field inhomogeneities cause a

frequency spread of about ± 4.5 kHz. The corresponding local-bin map (Fig. 2b) indicates, for each voxel, the frequency bin providing the highest image intensity. It represents a coarse off-resonance map, although reduced to 12 frequency steps. The local-bin map also demonstrates the composition of the final T_1 map from the individual bin T_1 maps. The gray dots indicate the positions of the four temperature probes and the gray contour outlines the area of the heat source.

T_1 and PRFS mapping

Figure 3 shows both the measured T_1 maps (a) and the measured ρ maps (b) over the entire experiment for selected time points. The T_1 increase starts from the two sides facing the heat source and eventually spreads over the entire phantom. The quality of the T_1 maps appears similar over the entire phantom, independent of the distance to the metallic implant. With a baseline T_1 ($T_{1\text{Base}}$) of 697 ms at 0 min, the maximum detected T_1 change corresponds to about 250 %. Globally, the ρ maps show a similar pattern as the T_1 maps, although there is an additional increase in ρ at the lower part of the phantom as well as small dipole patterns coming up in the lower right corner for ongoing heating. The quality of the ρ maps is significantly reduced in the off-resonance areas adjacent to the humeral head and the lower part of the humeral stem. Adjacent to these noisy areas, the heating pattern deviates from the expected shape.

Temperature Sensitivity Calibration

Figure 4 shows both the measured change in T_1 and the measured change in ρ over the temperature within the four ROIs for all 18 time points. The squares in Fig. 4a & b indicate both size and location of the ROIs, with the latter matching the location of the temperature probes. The maps themselves show the SNR variation within the underlying IR-2DMSI images (a) and the GRE images (b), respectively. For the IR-2DMSI image acquired with the longest TI, SNR varies between 40 and 140. The bands of reduced SNR indicate the boundaries between the individual bins and result from the fact that the images were composed of the local bin only. The maximum SNR within the GRE images is comparable, but decreases to the background noise level in the off-resonance areas next to the humeral head and the lower part of the humeral stem.

For each ROI, Fig. 4c shows the average T_1 value against the corresponding probe temperature. The T_1 variation within the ROIs primarily depends on the local temperature gradient but also increases with increasing temperature. For ROI I, the variation ranges from 2 % to 7 % and for the ROI IV from 1 % to 3 %. The ongoing heating during data acquisition determines the uncertainty of the temperature probe measurement. It is typically below 0.5 °C with the exception of the very first data points for ROI I, where it reaches up to 1.8 °C. For all four ROIs, including those next to the metallic implant, the data reveals a linear increase of T_1 with increasing temperature over a temperature range of 30 °C, corresponding to a T_1 change of about 90 %. T_1 values outside this linear range (data marked in gray) were excluded from the fitting process. The average slope of all four fits yields 21.8 ± 1.3 ms/°C, corresponding to a temperature sensitivity of $m = 3.1 \pm 0.2$ %/°C with respect to $T_{1\text{Base}}$. The thermometry uncertainty varied between $\Delta T = 1.4$ °C

and $T = 0.8\text{ }^{\circ}\text{C}$ corresponding to a T_1 uncertainty between $T_1 = 4.9\%$ and $T_1 = 2.4\%$, with the lowest value in the area of lowest temperature gradient (ROI IV).

Figure 4d reveals the corresponding results for the ρ measurements. Due to the short acquisition time for the GRE images, the uncertainty of the temperature probe measurements is determined by the precision of the probes of $0.1\text{ }^{\circ}\text{C}$. For the two ROIs most distant to the metallic implant (I and IV), there is a clear linear increase of ρ with increasing temperature over the entire temperature range of $50\text{ }^{\circ}\text{C}$. The average slope is $0.83 \pm 0.06\text{ E-}8/^{\circ}\text{C}$ and the resulting thermometry uncertainty is $T = 0.4\text{ }^{\circ}\text{C}$ or lower. Also for the two ROIs next to the metallic implant (II and III), the fits reveal a comparable slope. However, the individual measurements are strongly influenced by noise, resulting in a thermometry uncertainty as high as $T = 6.4\text{ }^{\circ}\text{C}$.

Temperature Maps

Figure 5 compares the resulting absolute temperatures maps derived from both the T_1 measurement (a) and the PRFS measurement (b) at similar time points. For conversion of the T_1 map into a temperature map, the average slope from all four ROIs was used. The PRF-based temperature map was calculated from the average of the two ROIs most distant from the metallic implant (marked by a black circle). In both maps, the temperature within the area of the heating source was set to $79\text{ }^{\circ}\text{C}$, the temperature of the water used as a heat source for the experiment. As the direct comparison shows, the PRFS-based temperature map appears smoother than the T_1 -based map, provided a certain distance to the off-resonance areas around both the humeral head and the lower part of the humeral stem is maintained. This is in accordance with the thermometry uncertainty T determined within the ROIs. The temperature patterns of both maps are globally comparable, although the heating seems to have propagated slightly further through the phantom in the PRF-based map. Some pixels of the T_1 -based map show temperatures higher than the heat source temperature. However, all temperatures above $50\text{ }^{\circ}\text{C}$ are outside the linear range for which this experiment was calibrated and therefore should be disregarded. The PRFS-based map also yields temperatures above that of the heat source in the lower part of the phantom. Evaluating the performance of the temperature mapping near the metallic implant, the PRFS-based map reveals no reliable temperature information in the off-resonance areas up to about 2 cm distant to the humeral head or the lower part of the humeral stem. In the areas adjacent to the implant exhibiting the highest off-resonance frequencies, the temperature map is dominated by noise. For moderate off-resonance (see arrows in Fig. 5b), the resulting temperature pattern is inconsistent including values below the start temperature. The deviations from the expected temperature pattern are as high as $10\text{ }^{\circ}\text{C}$. In contrast, the T_1 -based temperature map yields a reasonable temperature pattern over the entire phantom with no signs of disturbance around the metallic implant and also no increased noise even in close proximity to the implant.

Discussion

In this work, we have introduced T_1 -based MSI thermometry for non-invasive measurement of temperature near metallic devices and demonstrated its feasibility in a phantom

experiment. The technique uses a novel 2DMSI approach for faster artifact-reduced imaging and exploits the temperature dependency of the T_1 relaxation time. We also compared the performance of the technique with the commonly used PRFS thermometry technique and demonstrated the limitations of the latter in the presence of metallic devices.

T1-based MSI Thermometry

Our IR-2DMSI technique detected T_1 changes of up to 250 % with a performance that was independent of the distance to the metallic implant. The calibration with the temperature probes revealed an essentially linear T_1 increase over a temperature range of 30 °C. This is in accordance with previous results in both agarose gel [25] and tissue samples [26]. The increase corresponds to a T_1 temperature sensitivity of the doped agarose gel of about 3 %/°C, which is higher than typical in vivo sensitivities of 1–2 %/°C [2]. We assume that the observed deviation from the overall exponential behavior for temperatures above 60°C (ROI I) reflects the change in temperature sensitivity that occurs when the doped agarose gel starts to transition into an aqueous solution [14, 25, 27].

The uncertainty of the T_1 -based temperature measurement depends on both the T_1 temperature sensitivity and the uncertainty of the T_1 determination. For the given IR approach, the latter is determined by the SNR in the IR-2DMSI images and the sampling pattern of the IR curve. The T_1 determination becomes more precise with increasing SNR but primarily with increasing coverage of the IR curve towards higher T_1 values, as the latter allows for a better estimation of the equilibrium signal. Conversely, the uncertainty of the T_1 determination varies when measuring a changing T_1 , as can be seen by the increasing T_1 variation within the analyzed ROIs for increasing temperature. Note that the measure for the thermometry uncertainty used in this work is also affected by the temperature gradient within the analyzed ROI. For the given T_1 temperature sensitivity of $m = 3.1$ %/°C, the T_1 uncertainty $\Delta T_1 = 2.4$ %, achieved in the ROI with lowest temperature gradient (ROI IV), resulted a thermometry uncertainty of $\Delta T = \Delta T_1 / m = 0.8$ °C. Taking into account in vivo T_1 temperature sensitivities (1–2 %/°C), this T_1 uncertainty would correspond to a thermometry uncertainty between $\Delta T = 1.2$ °C and $\Delta T = 2.3$ °C. For the given setup, the dynamic range of the temperature measurement is defined by the range of linear T_1 increase over temperature rather than the dynamic range of the T_1 determination. Although this non-linear behavior might be calibrated for, it is the irreversible changes occurring during thermal coagulation that would limit the dynamic range in practical applications.

PRFS Thermometry

As the comparison between both techniques shows, PRFS thermometry generally features both a shorter acquisition time and a lower thermometry uncertainty than the T_1 -based approach, which is in accordance with previous reports [16]. The calibration data also revealed a linear increase over the entire temperature range of 50 °C. Deviations from the expected heating pattern in the lower part of the phantom (Fig. 3b) are likely to originate from susceptibility differences between the agarose gel and the tubes of the heating source, similar to the dipole patterns surrounding the air bubbles in the lower right corner which were formed at higher temperatures by gas emission.

PRFS thermometry is feasible near metallic devices, provided that both shape and alignment of the device with the static magnetic field cause limited field inhomogeneities, as the temperature map next to the middle part of the humeral stem shows. A similar behavior can be observed for T_1 -based thermometry using GRE techniques [26, 28]. However, as the temperature map demonstrates, PRF thermometry is increasingly impaired with increasing field inhomogeneity. The failure of the temperature measurement is obvious in areas of strong off-resonance variation, where the strong signal dephasing results in purely noisy maps. Field map analysis revealed that these areas corresponded to gradients of approximately 300 Hz/pixel and higher. In areas of moderate field inhomogeneities, SNR is higher, although obviously systematic errors appear (see e. g. arrows in Fig. 5b). These errors might result from slice distortions, with the measured temperature being correct but erroneously mapped to a different slice. However, taking into account the position of the heat source, this would neither explain the increased temperature (for instance to the left of the humeral head), nor the temperature below the baseline temperature next to it. A potential reason for these errors might be local, temporally varying field inhomogeneities due to temperature induced susceptibility changes. Reduction of the echo time of the GRE acquisition would allow one to reduce the signal dephasing and the associated signal loss, although it would not affect the slice distortions. But even with this short echo time of 4.2 ms, PRFS thermometry starts to fail about 2 cm away from the implant.

Temporal Resolution Considerations

The introduced T_1 -based MSI thermometry technique enables non-invasive temperature mapping near metallic devices also in the presence of strong field inhomogeneity. Both its precision and dynamic range may be promising for the detection of larger temperature changes. The underlying 2DMSI approach provides the basis for artifact-reduced imaging without time-consuming three-dimensional encoding. However, with an acquisition time of about 5 min per T_1 map, the temporal resolution of the current implementation is still limiting for most clinical applications.

For the current IR approach, the acquisition time could be reduced by optimizing the data acquisition by adapting the TI measurements and interleaving the bin acquisition. Assuming typical in vivo T_1 values in the range of 1000 ms [29] and a dynamic range of 20 °C (4 TI times, $TI_{\max} = 3 TI_{\text{Base}}$), we expect the acquisition time per T_1 map to drop to about one minute. This temporal resolution would be sufficient for applications with slower temperature changes such as hyperthermia or the monitoring of the bone radiation field during ablation treatments. The application of parallel imaging might allow for a further reduction in acquisition time, although the SNR dependency of the introduced technique would require further investigation. Provided the off-resonance distribution around the metallic device is known, we expect additional acceleration by adapting the number of acquired bins. For example, an off-resonance coverage off about ± 2 kHz would require only 5 bins instead of 12, thus reducing the acquisition time by another 60 % to 25 s per T_1 map. With typical sonication duration in the range of 20 s, this temporal resolution would be already sufficient for monitoring the result of an actual ablation. Provided the desired spot allows for coverage with one or two bins only, ultimately a temporal resolution in the range of 10 s might be reached. We expect the current matrix size of 128×128 to provide

sufficient spatial resolution for the above-mentioned clinical applications. In addition, the high readout bandwidth supports an increase in resolution along the read dimension. If also a higher resolution along the phase dimension is required, we recommend the use of a multi-shot approach with the discussed acceleration strategies applied to limit the associated increase in acquisition time.

For applications requiring both a high temporal resolution and broad off-resonance coverage, the findings of this work encourage the exploration of 2DMSI for faster T_1 mapping approaches or other thermometry mechanisms. The benefits of the T_1 relaxation time for MSI thermometry are that it is accessible with SE techniques and that it provides a useful range of linear temperature behavior with good sensitivity. However the T_1 temperature sensitivity is tissue type dependent, and irreversible changes may occur when tissue structure changes. A potential alternative mechanism might be the proton density, which exhibits lower but tissue independent temperature sensitivity [13]. For adipose tissue, the T_2 relaxation time is a promising candidate [30].

Conclusions

This work shows the feasibility of T_1 -based MSI thermometry for non-invasive temperature measurement near metallic devices, in regions where conventional methods such as PRFS mapping fail. The technique exploits a novel 2DMSI approach for artifact-reduced imaging in combination with quantitative T_1 mapping. Both the precision and the dynamic range of the technique are promising for monitoring temperature changes as small as 1–2 °C.

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Abbrev

MRI	Magnetic Resonance Imaging
MSI	Multi Spectral Imaging
MRgFUS	MR-guided Focused UltraSound
GRE	Gradient-Recalled Echo
SE	Spin Echo
ROI	Region of Interest
SNR	Signal-to-Noise Ratio
IR	Inversion Recovery

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Figure 1.

a) Pulse sequence diagram of the 2DMSI sequence, including the first two echoes of the spin echo train. The slice selection gradient is inverted for the excitation pulse (see red circles).

b) Slice profiles for excitation and refocusing with the inverted slice selection gradient. The overlap of both profiles yields the profile of the frequency bin, which contributes signal.

c) The overlay of all frequency bins shows the individual contributions to the composite slice profile, which can be improved by adjusting the overlap between adjacent bins.

Figure 2.

a) Individual bin images of an IR-2DMSI acquisition and the corresponding bin center frequency. The red line indicates the contour of the implant. **b)** Local-bin map, showing the frequency bin that provides the strongest signal at each voxel. The gray dots mark the positions of the fluoroptic temperature probes, the gray line the area of the heat source.

Figure 3.

a) Series of T_1 maps acquired at different time points of the heating process. **b)** Series of Δs maps acquired right after the T_1 maps.

Figure 4.

a) SNR map of the IR-2DMRI local-bin image acquired with the longest TI. The squares show both size and location of the ROI used for the analysis in c) and d). The ROI location matches the location of the temperature probes. **b)** SNR map of the GRE images used for PRFS mapping. **c)** Average T_1 over the measured probe temperature for all four ROIs and resulting fits. Values marked in gray were excluded from the fitting process. The slope m represents the absolute temperature sensitivity; the value in brackets corresponds to the relative sensitivity with respect to $T_{1\text{Base}}$. The thermometry uncertainty ΔT is derived from the differences between fitted and measured temperature values. **d)** Similar analysis for the Δs maps, with α describing the slope.

Figure 5.

Absolute temperature maps based on the T_1 measurements (a) and the PRFS measurements (b) at comparable time points. The temperature in the area of the heat source was set to the temperature of the heating water. Black circles indicate the location of the temperature probes used for calibration. The arrows point to areas of inconsistent temperature pattern in the PRFS temperature map.