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Dual echo z-shimmed proton resonance frequency-shift MR thermometry near metallic ablation probes: Technique and temperature precision

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Abstract

Purpose—To improve the precision of proton resonance frequency-shift MR thermometry near ablation probes, by recovering near-probe image signals that are typically lost due to magnetic susceptibility-induced field distortions.

Methods—A dual-echo gradient-recalled echo sequence was implemented, in which the first echo was under- or over-refocused in the slice dimension to recover image signal and temperature precision near a probe, and the second echo was fully-refocused to obtain image signal everywhere else in the slice. A penalized maximum likelihood algorithm was implemented to estimate a single temperature map from both echoes. Agar phantom and ex vivo experiments with and without microwave heating at 3 Tesla evaluated how much temperature precision was improved near a microwave ablator compared to a conventional single-echo scan as a function of slice and needle orientation in the magnet.

Results—The number of near-probe voxels with temperature standard deviation $\sigma > 1$ °C was decreased by 51% in the phantom experiment, averaged across orientations, and by 31% in the pork. Temperature maps near the probe were more smoother and more complete in all orientations.

Conclusion—Dual echo z-shimmed temperature imaging can recover image signal for more precise temperature mapping near metallic ablation probes.

Keywords

MR thermometry; interventional MRI; z-shimming; ablation; microwave

Introduction

Proton resonance frequency-shift (PRF) MR thermometry has been widely applied to monitor temperature change during thermal ablation [1]. However, in RF, microwave and

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Supporting Information: Dual echo z-shimmed proton resonance frequency-shift MR thermometry near metallic ablation probes

laser ablations, the magnetic susceptibility difference between the ablation probe and surrounding tissue creates B_0 field distortions that lead to through-plane and in-plane phase gradients through a voxel that cause signal loss and pile-up, and prevent precise temperature monitoring near the probe. This is especially a problem when the target lesion size is small relative to the size of the signal hole. One example is in minimally-invasive ablation of the hippocampus in the treatment of medically intractable epilepsy, using a curved metallic needle inserted through the foramen ovale of the skull that carries a laser fiber or RF electrode [2, 3]. PRF-shift thermometry near implanted electrodes is also desirable for assessing their RF safety, but suffers the same problem [4].

Several sequences and reconstructions have been proposed to mitigate signal loss and distortions caused by metallic interventional devices and implants. The classic method which addresses in-plane distortions in spin echo imaging is view angle tilting [5, 6]. More recently, MAVRIC and SEMAC imaging have been developed for spin echo imaging near metallic implants [7, 8], and are based on a combination of view angle tilting with slice-dimension phase encoding. This approach has further been applied to T_1 -based thermometry [9] near implants. Multi-frequency reconstruction methods have been developed to alleviate in-plane distortions in center-out radial gradient-recalled echo (GRE) sequences [10]. Through-plane signal loss in GRE sequences has been addressed in functional MRI using z-shimming, in which multiple images are acquired with different slice-dimension phase encode gradient moments that are tuned to compensate through-plane phase gradients in brain tissue above the sinuses and ear canals [11–13]. Quadratic phase excitation pulses have also been applied for this purpose, but come at the cost of decreased signal away from the recovered regions [14]. The white marker method is an application of z-shimming to interventional MRI, for the purpose of localizing an interventional device [15]. In that method, an acquisition is performed with an unbalanced slice rewinder, which suppresses signals from all tissues except those adjacent to the device. The white marker method has been incorporated into a real-time dual-echo SSFP sequence, to simultaneously obtain a wire image and a background tissue image to overlay it on [16].

In this work we apply z-shimming to GRE PRF-shift MR thermometry, to compensate through-plane phase gradients and recover temperature precision near an ablation probe. Analogous to the white marker method [16], we propose a dual-echo sequence in which the first echo is z-shimmed to recover image signal near the probe, and the second echo is fully-refocused to obtain high image signal everywhere else. An associated penalized likelihood approach is used to reconstruct a single temperature map from both echoes. Phantom and ex vivo experiments were performed with and without microwave heating to characterize image signal recovery and temperature precision improvement using the sequence and reconstruction, compared to a conventional single-echo MR thermometry scan. Overall the method aims to improve the precision of PRF-shift MR thermometry near metallic ablation probes.

Methods

Pulse Sequence

Figure 1 shows the proposed dual echo sequence, in which one echo is z-shimmed to compensate though slice phase accrual due to off resonance and obtain higher image signal near the probe, while the other echo is not z-shimmed to obtain higher image signal elsewhere. The dependence of the PRF-shift phase difference signal-to-noise ratio (SNR) on echo time (TE) indicates which echo should be z-shimmed. Specifically, accounting for intra-voxel off-resonance variation $\omega(\vec{x})$ and neglecting terms that are independent of TE , the phase difference SNR can be written as [1]:

$$\text{SNR}_{\Delta\phi} \propto TE e^{-\frac{TE}{T_2^*}} \left| \int e^{-i\Delta\omega(\vec{x})TE} dV \right|, \quad (1)$$

where the volume integral is taken over the voxel which is approximated as a cube of uniform intensity. For voxels without significant off-resonance, the volume integral is approximately one and SNR is highest when $TE = T_2^*$ [1]. However, the optimal TE will be shorter for voxels with significant off-resonance variation, since the volume integral will decay with increasing TE . Thus, the first echo in the dual-echo sequence should be z-shimmed to obtain higher image signal near the probe since, even with z-shimming, in-plane and non-linear through-plane off-resonance gradients shorten the optimal TE there. In addition, because temperature will be highest near the probe a shorter TE may be necessary there to avoid phase wrapping and consequently inaccurate temperature values.

In the proposed sequence the first echo is z-shimmed by setting the first rephasing gradient trapezoid after slice selection to $p\%$ of its fully-refocused area. This cancels the linear component of the through-slice phase accrued between the excitation pulse and the echo. p is set to maximize average image magnitude in a region-of-interest (ROI) near the probe. In this work, the ROI was defined as the set of voxels where the fully-refocused second echo image's magnitude was less than 70% of the mean magnitude elsewhere in the phantom. Since the z-shim will reduce image signal away from the probe where there is no significant intra-voxel off-resonance, a second slice refocusing gradient is placed between the first and second echoes with $(100 - p)\%$ of the full refocusing gradient area, and the second echo has a longer conventional $TE \approx T_2^*$ echo time [1]. The echoes are measured with the same readout gradient polarity so that chemical shift occurs in the same direction [17]. The sequence was implemented on a 3T scanner (Philips Achieva, Philips Healthcare). Unless otherwise stated below, scans with the sequence used 8 ms z-shimmed TE , 16 ms fully-refocused TE , 50 ms TR, 25° flip angle, $160 \times 160 \text{ mm}^2$ FOV, 176×176 reconstructed matrix size, 4 mm slice thickness and 220 Hz pixel bandwidth. The slice rewinder area for a 100% refocused slice was 4.3 cycles/cm. The body coil was used for excitation and a 32-channel head coil was used for signal reception.

Temperature estimation

The multi-echo hybrid multibaseline and referenceless thermometry algorithm [18, 19] was used to estimate a single temperature map from both echoes. The algorithm models each echo's image as:

$$y_j = \left(\sum_{b=1}^{N_b} x_{b,j} w_b \right) e^{i2\pi(\{Ac\}_j + f_j)TE} + \epsilon_j, \quad (2)$$

where y_j is the signal of the j th image voxel, x_b is the b th complex image from the baseline library containing N_b images, which are combined to form a single baseline image via a weighted sum with weights w , A and c are a matrix of polynomial phase basis functions and its coefficient vector, respectively, f_j is the heat-induced frequency shift at location j , and ϵ_j is complex Gaussian noise. In this work a single baseline was used for each echo, along with a zeroth-order (constant) polynomial frequency shift, and the model in Eq. 2 was fit to the dual echo images y_1 and y_2 jointly using a descent algorithm that minimizes a penalized-likelihood cost function:

$$\psi(\mathbf{f}, c) = \sum_{j=1}^N \frac{1}{2} \left| y_{1,j} - x_{1,j} e^{i2\pi(c+f_j)TE_1} \right|^2 + \frac{1}{2} \left| y_{2,j} - x_{2,j} e^{i2\pi(c+f_j)TE_2} \right|^2 + \lambda |f_j|, \quad (3)$$

where λ is an ℓ_1 sparsity regularization parameter and N is the number of image voxels. The sparsity regularization reflects a priori knowledge that heat will be localized around the probe, which helps separate heating-induced frequency shifts from the center frequency drift. The algorithm works by alternately updating $\mathbf{f}$ and c to reduce $\psi(\mathbf{f}, c)$ using gradient descent with optimization transfer-derived step sizes [20]. It also enforces a non-positivity constraint on $\mathbf{f}$ since temperature increases cause frequency to decrease. Because the ℓ_1 penalty introduces a small downward bias in temperature, after minimizing Equation 3 the minimization is repeated with $\lambda = 0$ and $\mathbf{f}$ is updated only in voxels where estimated temperature is above 0.1°C . Once fit, the frequency shifts are converted to temperature according to:

$$\Delta T_j = \frac{2\pi f_j}{\alpha \gamma B_0}, \quad (4)$$

where $\alpha = -0.01 \text{ ppm}/^\circ\text{C}$, γ is the gyromagnetic ratio, and B_0 is the main field strength in Tesla. MATLAB code for the algorithm to process the porcine muscle dual-echo microwave heating dataset (described below) can be downloaded from https://bitbucket.org/wgrissom/thermometry_near_probes_paper. Compute times were approximately 0.25 seconds per 176×176 map, using a 3 GHz Intel Core i7 laptop with 16 GB RAM (Apple Inc, Cupertino, CA, USA).

Experiments

Image signal recovery versus— p To illustrate how the recovered image signal near the probe depends on p , images of an agar phantom were acquired with a nitinol microwave probe embedded in it, without heating. The probe (Eco Inc., Nanjing, China) had a 3 mm diameter and a metal heat source ring located 1 cm from its tip. The slice was positioned to approximately coincide with the heat source ring. Images were acquired with p values ranging from 30 to 160%, and a spin echo image was acquired in the same slice position, to evaluate how well z-shimming closed the signal loss hole around the probe.

Recovery of temperature precision—To measure how much temperature precision was recovered near the same microwave ablation probe as above, 100 dynamic images of the agar phantom were acquired without heating. Four sets of images were acquired with the probe parallel and perpendicular to B_0 , and with the slice parallel and perpendicular to the probe. For perpendicular slice-probe orientations, the slice was positioned approximately at the heat source ring plane, and for parallel slice-probe orientations the slice was positioned in the same plane as the probe. Temperature maps were estimated from both echoes and from the conventional fully-refocused echo, and temperature standard deviation was calculated across time for each voxel to assess precision.

Agar phantom heating—To characterize recovery of actual temperature maps during heating, four agar phantoms were heated using a microwave generator (Eco Inc., Nanjing, China) and the nitinol microwave probe described above, at 25 Watts for five minutes. During the heating, four temperature scans were acquired with the same four probe and slice orientations as above. Temperature maps were estimated from both dual-echo and conventional fully-refocused single-echo gradient echo images.

Ex-vivo porcine muscle heating—Temperature standard deviation and heating experiments were conducted on a piece of porcine muscle tissue (psoas major) to evaluate signal recovery in actual tissue. The scans used $p = 50\%$, and the same sequence parameters as above except with 21° flip angle, $100 \times 120 \text{ mm}^2$ FOV, 240×240 reconstructed matrix size and 542 Hz pixel bandwidth. Images were acquired with the probe parallel to B_0 and with the imaging slice perpendicular to the probe and approximately in the heat source ring plane. The sample was heated at 25 Watts for three minutes. During heating, dynamic images were acquired to measure the temperature change. 100 dynamic images were also acquired before heating to calculate temperature standard deviation maps.

Results

Image signal recovery versus p

Figure 2a shows that the value of p determines how much and where image signal is recovered in the first echo image, and Figure 2b plots the mean voxel magnitudes in sum-of-squares combinations of the dual-echo images as a function of p , in a near-probe ROI defined as the region where the fully-refocused second echo image had less than 70% of its mean intensity elsewhere in the phantom. The magnitudes are normalized by the mean image signal in that same ROI in the second echo image. The plot shows an optimal value of

$p = 140\%$, corresponding to a through-slice phase gradient of -1.7 cycles/cm. Images acquired with $p = 140\%$ were used to form the sum-of-squares-combined image in Figure 2c. From the images in Figure 2c it is evident that the first echo image shows enhanced signal around the probe, while the second echo gives a conventional gradient echo image with signal everywhere else. When the magnitudes of the first echo and second echo images were combined by sum-of-squares, the size of the signal loss hole was reduced: the full-width-at-half-max (FWHM) of the hole in the spin echo image was 4.4 mm, the FWHM of the hole in the conventional fully-refocused echo image was 11.6 mm, and the FWHM of the combined image was 7.5 mm, which was 35% narrower than the conventional image.

Recovery of temperature precision

Figure 3 shows images and temperature standard deviation maps across probe and slice orientations. The values of p are indicated for each case, and correspond to phase gradients of (starting at the top row) $+1.7$, $+2.2$, $+2.2$, and -1.7 cycles/cm. The optimal values of p are different in each orientation because the linear component of the z-directed susceptibility-induced field distortion is different in each orientation. Image magnitudes (first echo, second echo, and sum-of-squares-combined dual-echo) are shown on the left. Recovery is evident in each orientation and is reflected by the relative ROI signal in the combined images, defined as above for Figure 2b. Temperature standard deviation maps derived from fully-refocused single echo and dual-echo images are shown to the right. In all cases the dual-echo maps contain fewer low-precision voxels around the ablator, and similar or better precision elsewhere in the phantom. The number of low precision voxels with $\sigma > 1^\circ\text{C}$ is indicated for each case. Averaged across orientations, the number of low precision voxels decreased by 51% in the dual-echo maps. The mean single-echo background temperature standard deviation was 0.17°C . The white arrows in the σ maps in the fourth row (parallel needle/perpendicular slice) point to a representative voxel where standard deviation was reduced from 2.8°C to 1.1°C using the dual-echo scan, and the temperature in that voxel is plotted across dynamics in Supporting Information Figure S1. The conventional fully-refocused single-echo sequence yielded a maximum error of 15.0°C in this voxel, while the z-shimmed dual-echo sequence had a maximum absolute error of 2.6°C .

Agar phantom heating

Figure 4 shows temperature maps after five minutes of heating in each of the four probe/slice orientation profiles. The single-echo temperature maps contain irregular temperature edges and holes where the estimated temperature was negative, either due to low SNR or phase wrapping at temperatures above 24.5°C . The dual-echo maps are smoother and more complete in every case.

Ex-vivo porcine muscle heating

Figure 5 shows the results of *ex-vivo* heating experiments on porcine muscle. Figure 5a confirms that the best value of p was 50% (equivalent to a $+2.2$ cycle/cm phase gradient). First and second echo magnitude images of the dual-echo sequence are shown in Figure 5b. Temperature standard deviation maps (Figure 5c) demonstrate that the dual echo maps contain 31% fewer high- σ voxels near the metallic probe. Figure 5d shows temperature maps at low heat ($< 24.5^\circ\text{C}$, before phase wrapping would occur in the single echo map) and

high heat, as well as maximum temperature across the ablation. Temperature errors in the low heat single-echo map are due to low SNR only, while errors in the high heat single-echo map are due to both low SNR and phase wrapping. In the low-heat single-echo map, 17% of the voxels in which the dual echo map had temperatures larger than 15°C were negative, reflecting the single-echo map's low precision near the probe.

Discussion

Summary of results

This work presented a dual echo sequence with z-shimming, which can recover image signal for more precise temperature mapping near metallic ablation probes and wires. An associated penalized likelihood approach to estimate a single temperature map from both echoes was also described. The first echo was z-shimmed to measure near-probe image signals by changing the slice rephasing lobe immediately preceding it to $p\%$ of its full-refocusing area. Image signals away from the probe were measured with a typical PRF thermometry echo time, after fully refocusing the slice using a second refocusing gradient placed between the two echoes. The dual echo sequence has the same scan time as a conventional single-echo 2DFT PRF temperature mapping sequence, since the added echo occupies an otherwise empty time period between the excitation and readout.

Temperature precision in a phantom was measured in multiple microwave probe and slice orientations using dynamic images acquired without heating, which showed that the number of voxels near the probe with greater than 1° C uncertainty was reduced 51% on average using two echoes versus one. That experiment also showed that the best value of p depends on the probe and slice orientation with respect to B_0 , and would need to be changed if the relative orientations change. Phantom microwave heating experiments then showed that in each orientation, the dual-echo method effectively filled in temperature in the signal loss hole near the probe. Finally, experiments with a piece of porcine muscle tissue validated the method in heating of actual tissue. Interestingly, the best value of p in the porcine muscle was approximately opposite that of the phantom experiments with the same probe and slice orientation (140% for the phantom versus 50% for the porcine muscle). Though we made an effort to line up the slice with the metal heat source ring in both cases, this difference may have been due to opposite slice position errors with respect to the ring between the two cases. Further work is needed to verify to what degree these results apply to other probes, with different sizes and conductor geometries. Like other PRF-shift thermometry measurements in aqueous tissue, the accuracy of the proposed method depends primarily on the PRF shift coefficient of the tissue. In future work accuracy could be verified using fiber optic probe measurements.

Tuning the z-shim (p) efficiently

The z-shim parameter p was tuned to maximize signal magnitude recovery near the probe, as a proxy for optimizing temperature SNR. This approach was based on the fact that temperature SNR is proportional to signal magnitude [1], and it was implemented by acquiring images across a range of p values. While this approach is feasible if the probe and slice orientations stay approximately fixed throughout a treatment (as would be the case in

ablating the hippocampus along its length), it would be impractical to repeat it frequently within a single treatment. Tuning may be accelerated using methods developed for z-shimmed functional MRI, such as a procedure that fits a k-space slice profile to a set of images acquired across a sparse set of p values [21]. It may also be possible to use a lookup table if the probe is tracked, since for a fixed TE , the optimal p value is likely to be insensitive to tissue type. Finally, a possible extension to obviate case-by-case tuning would be to add more echoes to the sequence, each with a different value of p . Different values of p recover different regions, so this acquisition could more effectively fill in the signal loss region. The echoes could further be acquired with a high pixel bandwidth to minimize in-plane distortions, and estimating temperature from all echoes jointly could compensate for precision lost due to the increased bandwidth [18]. This may further enable simultaneous water/fat separation, for monitoring ablations in fatty tissues such as breast and diseased liver.

Possible extensions

The simplicity of the proposed sequence could make many other extensions and improvements possible. The TE of the first echo is a degree of freedom that could be explored to further improve precision. TE and p could be jointly tuned by imaging across a set of p and TE pairs, and selecting the pair that maximizes the product of the average ROI signal amplitude and the TE . If a center-out radial imaging readout is used instead of a Cartesian readout, the method could be combined with off-resonant reconstruction to simultaneously reduce in-plane distortions near the probe [10]. Many ablation probes have thermocouples in their tips to provide temperature measurements at the point of ablation; the improved near-probe temperature measurements yielded by the proposed method may make it possible to accurately interpolate temperature over the remaining tissue next to the probe whose signal is not fully recovered, using the thermocouple temperature as a data point in the middle of that tissue. In that case, the combination of PRF measurements with thermocouple measurements could make absolute temperature monitoring possible.

Conclusion

A dual echo z-shimmed temperature imaging sequence and an associated temperature reconstruction algorithm were described, which can recover image signal for more precise temperature mapping near metallic ablation probes. This is particularly important when the target lesion size is small relative to the region of signal loss around the probe, such as when ablating small tumors or small structures such as the hippocampus with metallic ablation probes. The method may also find application in assessing RF tissue heating near implanted electrodes.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1.
The proposed dual echo z-shimmed PRF-shift thermometry pulse sequence.

Figure 2.

(a) First echo images in an agar gel phantom with different values of p show different levels and distribution of signal recovery near the probe. Zoomed near-probe regions are shown in the lower left insets. The green highlighted region in the zoomed $p = 100\%$ image is the near-probe ROI for measuring signal recovery. The ROI was defined as the set of voxels around the probe where the second echo image had less than 70% of the mean signal amplitude in the phantom. (b) Mean voxel magnitudes in the near-probe ROI in sum-of-squares combined dual-echo images, normalized by the mean intensity in the same ROI in the second echo (100%) image. ROI signal is maximized at $p = 140\%$. (c) Magnitude images of the first echo ($p = 140\%$), the second echo and the combined two-echo images. The region near the probe is zoomed in for each case. A spin echo image is also shown to illustrate how close the recovery comes to filling in the hole around the probe.

Figure 3.

Image signal recovery and temperature map precision in agar gel phantoms with different probe and slice orientations and no heating. The value of p used is reported for each case. Magnitude images of the first echo and second echo images are shown in the first two columns, and their combined magnitude is shown in the third column. For the combined images, mean signal in the low-signal ROI around the wire is reported as a percentage of the mean ROI signal in the second echo images. The positions of the perpendicular slices are indicated by the yellow lines in the magnitude images in the parallel slice images. Maps of temperature standard deviation (σ) across time are shown in the fourth and fifth columns, for the fully-refocused echo alone and for the dual-echo scan. The number of voxels with $\sigma > 1^\circ\text{C}$ is reported on an inset zoomed-in σ map for each case; smaller numbers are better. Figure S1 plots the temperature curve over time for the recovered voxel indicated by the white arrow in the parallel probe/perpendicular slice σ map, to illustrate the range of temperature errors with single- and dual-echo scans.

Figure 4.

Temperature maps in the agar gel phantoms after five minutes of heating in each needle and slice orientation using single echo baseline subtraction and the proposed dual echo method. The positions of the perpendicular slices are indicated by the yellow lines in the magnitude images in the parallel slice images. To facilitate comparison, the single echo maps were masked to zero where the dual echo maps were zero. Zoomed near-probe regions are shown in the lower left insets. In all cases the dual-echo maps are smoother near the probe.

Figure 5.

Porcine muscle heating experiment. (a) Mean image signal recovery in the near-probe ROI as a function of p . The signal is maximized at $p = 50\%$. (b) First echo ($p = 50\%$) and second echo images from the dual-echo sequence. (c) Temperature standard deviation across 100 dynamics before heating. The number of voxels with $\sigma > 1^{\circ}\text{C}$ is reported; smaller numbers are better. (d) Temperature maps at low heat ($< 24.5^{\circ}\text{C}$, before temperature wrapping would occur in the single echo map), and high heat, as well as maximum temperature across the 180 second ablation at 25 Watts, using single echo baseline subtraction and the proposed dual echo method. Zoomed near-probe regions are shown in the lower left insets in (b), (c), and (d). To facilitate comparison, the single echo maps were masked to zero where the dual echo maps were zero.