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Respiration Artifact Correction in 3D PRF MR Thermometry using Phase Navigators

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Abstract

Purpose—To develop reliable 3D segmented echo-planar imaging (seg-EPI) proton resonance frequency (PRF) temperature monitoring in the presence of respiration induced B0 variation.

Methods—A free induction decay (FID) phase navigator is inserted into a 3D seg-EPI sequence before and after EPI readout, to monitor B0 field variations. Using the field change estimates, the phase of each k-space line is adjusted to remove the additional phase from the respiratory induced off-resonance. This correction technique is evaluated while heating with MR guided focused ultrasound (MRgFUS) in phantoms with simulated breathing and during non-heating conditions in healthy in vivo breasts.

Results—With k-space phase correction, the standard deviation of magnitude images and PRF temperature measurements in breast from 5 volunteers improved by an average factor of 1.5 and 2.1, respectively. Improved accuracy of temperature estimates was observed after correction while heating with MRgFUS in phantoms.

Conclusion—Phase correction based on two FID navigators placed before and after the echo-train provides promising results for implementing 3D monitoring of thermal therapy treatments in the presence of field variations due to respiration.

Keywords

PRF Temperature; respiration correction; phase navigator

INTRODUCTION

The proton resonance frequency (PRF) shift method (1) has been widely adopted to measure temperature changes in tissue during thermal therapy treatments due to its linearity over the temperature range of interest, the constant of proportionality being largely independent of tissue type (except adipose tissue) (2), and its ability to produce temperature maps with the spatial and temporal resolution required to monitor treatments in real time. PRF temperature mapping interprets any phase change as a change in temperature and is therefore susceptible to errors from any phase changes that are not temperature induced, such as motion. Even when there is no motion within the imaged volume, such as when imaging a stationary

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breast or brain, phase changes induced by respiratory motion have been shown to produce artifacts (3–9). Respiratory induced motion of the abdomen and lungs change the distribution of magnetic susceptibility, altering the B0 field and corresponding resonant frequency distribution throughout the subject including the non-moving imaged volume. Each frequency offset results in an added linear phase evolution during readout that changes between excitations, causing ghosting artifacts. This was investigated in the breast by Peters *et al* (4), who measured an average field fluctuation of 0.13 ppm during regular respiration throughout the breast. Similar but smaller field fluctuations have also been observed in the brain (7,10).

Several methods have been proposed to correct the respiration induced B0 variation in the breast (3,6,8). While these correction schemes were successful, they all required a library of baseline images that adequately characterized the entire respiration period and were all limited to 2D single slice imaging, limiting the monitored region. While this 2D field of view results in a short temporal resolution that can adequately capture the respiratory cycle, it restricts the volume of tissue monitored during thermal therapies.

Three-dimensional PRF sequences offer some important advantages over 2D methods, but have several challenges that must be overcome. Three-dimensional acquisitions can provide high spatial resolution sampling over the target volume and a larger field of view, which is advantageous for thermal therapy techniques, in general, to assess the effectiveness and safety of the treatment. However, 3D acquisitions can require longer acquisition times, reducing temporal resolution and making the images more susceptible to artifacts from subject motion. Echo-planar imaging (EPI) sequences are often used to increase temporal resolution while maintaining the larger field of view. Because of the low phase encoding sampling bandwidth, EPI sequences can have increased sensitivity to phase errors due to B0 variations. The amount of off-resonance varies continuously with respiration and if the image acquisition is faster than the field variation the phase error in each image could potentially be corrected using a multi-baseline method. However, when the total acquisition time covers a significant portion of one or multiple respiratory cycles, the phase variations cause ghosting. In addition the slice orientation of the imaging volume during EPI acquisitions has a strong impact on respiration-induced artifacts (7).

The B0 shift due to respiration can be measured using navigator acquisitions. Several navigator methods have been used for motion correction (11,12), to reduce signal fluctuation (13), and improve image quality (14,15). Some methods use free induction decay (FID) navigators (16) while others applied a gradient during navigator readout (11,15,17) or extracted the navigator information from the k-space data itself (self-navigated) (10,18,19). These methods use navigator information to estimate the phase at the signal echoes and subsequently remove the unwanted phase.

This paper presents an improved method that corrects respiratory induced phase variations in a 3D seg-EPI sequence by collecting two internal FID phase correction navigators, one before the EPI readout and one after, to estimate the respiration induced field change. This technique improves upon other methods by acquiring a 3D image volume with a relatively short temporal resolution without the need for a library of baseline images. The ability of the

phase correction navigators to estimate respiration induced field shifts in in vivo breast is demonstrated in normal volunteers. The accuracy of temperature measurements using the correction technique during MRgFUS is demonstrated using phantom studies.

METHODS

Phase Navigator

A 3D seg-EPI sequence was modified to include flyback EPI readout and two internal phase correction navigators as shown in Figure 1a. The flyback readout ensures that the fat chemical shift is always in the same direction and eliminates the need for standard EPI phase correction. The navigators have no gradients during readout and provide an estimate of the average phase over the sensitive volume of each channel in the receiver array. Phase correction is performed by assuming a uniform shift in B0 and resulting frequency offset over the sensitive area of each coil. While it has been shown that the B0 offset varies spatially (4), making this assumption slightly erroneous, using an array of coils as is done in this work, does provide some spatial sensitivity to the off-resonance measurement.

An outline of this correction method is shown in Figure 1. The phase difference between the two FID's consists of a constant phase difference due to the offset of the image and a variable phase due to the frequency variation with respiration. An example of the phase of the two FID's is shown in Figure 1b. The navigator signal measured by the j^{th} coil after removing the Larmor frequency is described in Eq. 1,

$$S_j(t) = \int_{\text{volume}} c_j(\mathbf{x}) m(\mathbf{x}, t) e^{-i\gamma \int_0^t \Delta B(\mathbf{x}, \tau) d\tau} d\mathbf{x} \quad [1]$$

where c_j is the j^{th} coil sensitivity, $m(\mathbf{x}, t)$ is the magnetization distribution, and $\Delta B(\mathbf{x}, t)$ is the field shift. The phase difference between the two navigators was calculated by multiplying the second navigator by the complex conjugate of the first navigator using equation 2, where it is assumed that the field shift is constant over space and time during each TR.

$$\Delta\varphi_j = \angle(S_j^2 S_j^{1*}) = \angle(e^{-i\gamma \Delta B(t_2 - t_1)}) \quad [2]$$

Measurement error is reduced by taking the average phase difference of corresponding samples in each FID. Figure 1c shows an example of the phase difference during one image acquisition for each of the eight receiver coils. In addition to the phase variation with respiration, there is a constant phase difference between the channels due to the small spatial variation in B0 (imperfect shim) between sensitive volumes. The constant phase difference between each coil is removed by a baseline correction. The average phase difference of the first image is used as a baseline, subtracting from all subsequent image acquisitions for each receiver coil. Note that this method also measures and corrects for a weighted average B0 field drift for each coil (weighted somewhat by the coil sensitivities).

After removing the constant phase difference $\phi_{j, \text{const}}$, the remaining phase variation is primarily due to temporal B0 variation, as shown in Figure 1d. The field shift is calculated using equation 3

$$\Delta B_j = \frac{\Delta \phi_j - \Delta \phi_{j, \text{const}}}{\gamma \Delta t} \quad [3]$$

where ϕ is the phase difference, γ is the gyromagnetic ratio, and t is the time difference between navigator readouts. The field shift is shown in parts per million (ppm) in figure 1e.

After excitation, phase accumulates linearly in time by $\phi(t) = B * \gamma * t$. The phase of the k-space lines acquired each excitation is adjusted to remove the extra phase accumulated due to respiration at their individual read time in the echotrain, incorporating echo-shifting. The field offset B is calculated and used for correcting each coil separately. After adjusting the signal phase, the corrected signal is described by equation 4,

$$S_j(k_x, k_y) = \int_{\text{volume}} c_j(\mathbf{x}) m(\mathbf{x}, t) e^{-i\gamma \int_0^t [\Delta B(\mathbf{x}, \tau) - \Delta B_j] d\tau} e^{-i2\pi(k_x x + k_y y)} d\mathbf{x} \quad [4]$$

where B_j is measured every excitation.

Experiments

To demonstrate the ability of the phase navigators to correct respiration induced artifacts, two types of experiments were performed. The first experiment evaluated the correction in vivo breast during non-heating conditions, and the second experiment evaluated the correction technique during MRgFUS heating in both a gelatin and salt pork phantom. In this situation, a male volunteer was positioned above the phantom setup to create the respiration artifact while heating with MRgFUS. All experiments were performed in a Siemens TIM Trio 3 Tesla MRI scanner (Siemens Medical Solutions, Erlangen, Germany) using a breast-specific MRgFUS system (Figure 4a & b) with an integrated 8-channel RF coil and an MRI-compatible phased array transducer (256 elements, 1 MHz frequency, 13 cm radius of curvature, Imasonic, Besançon, France and Image Guided Therapy, Pessac, France) (20–22). All human studies were approved by the local Institutional Review Board (IRB) and performed with informed consent.

In vivo breast non-heating experiments—Five normal female volunteers (age 29 – 50) were positioned in the breast-specific MRgFUS device in a head first prone position. After localization, multiple 3D segmented-EPI GRE image volumes were acquired in the coronal orientation with no ultrasound applied while the volunteer was free breathing (Voxel spacing = 1×1×3 mm, FOV = 224×154×24 mm, Matrix = 224×154×10, Flip Angle = 20°, TR/TE = 47/15 ms, EPI Factor = 7, 8 slices with 25% oversampling, readout BW = 744 Hz/Px, phase encoding BW = 56 Hz/Px. Spectral fat saturation pulse applied before every TR, 30 Repetitions, 10.3 seconds per image). Volunteer 1 moved several times during the first nineteen image acquisitions and volunteer 2 moved during the seventh image

acquisition, causing image misregistration, and therefore only the last 11 and 23 images were used for these two volunteers, respectively. PRF temperature estimates were calculated using the first of the remaining images as the reference phase. Three-point Dixon data were acquired using the same imaging parameters with TE = 16/17.2/18.4 ms and no fat saturation pulses. These data were used to calculate separate water and fat images. After shifting the fat images to compensate for the large chemical shift artifact present in seg-EPI data, these images were used for tissue segmentation. Only the PRF temperature estimates within water and glandular tissue were used to analyze the effectiveness of the correction method.

MRgFUS phantom experiments—The correction technique was evaluated during heating conditions using both a gelatin phantom (23) and an excised pork sample that contained significant amounts of both fat and aqueous tissues. In order to simulate respiration effects, a male volunteer lay prone above the breast-specific MRgFUS device breathing freely during the ultrasound sonications. A separation gap of approximately 5 mm was intentionally left between the phantom and the chest of the volunteer eliminating any possible heating of the volunteer, and reducing the possibility of bulk phantom motion during the heating period while allowing respiration-induced variation of B0. The 3D imaging volume was prescribed in a coronal orientation with the same imaging parameters that were used in the in vivo breast non-heating experiment. The gelatin phantom used the same 3D seg-EPI sequence with the same MR parameters as the breast volunteers except (FOV = 224×154×24 mm, Matrix = 224×154×10, TR/TE = 32/15 ms, no fat saturation, 7 seconds per image). Four sets of data were collected for both the gelatin and excised pork phantoms. For comparison, two image sets were acquired without the volunteer placed above the phantom. The first image set served as a control without FUS heating or breathing artifact. The second set provided a baseline of the MRgFUS heating (25 acoustic W, 60 seconds) without the breathing artifact. The third and fourth image sets repeated imaging sets one and two with the volunteer above simulating respiration effects. The pork phantom was unintentionally displaced ~1–2 mm while placing the volunteer above, causing a slight position shift between the breathing and non-breathing situation. The PRF temperatures were determined using the average phase of 5 baseline images obtained with no ultrasound heating as the reference phase.

RESULTS

In the 5 female volunteers, it was found that the field shift amplitude was patient specific. While the field shift values shown in Figure 1d (Volunteer 5), as well as volunteer 3, oscillated with an amplitude of approximately 0.2 ppm, volunteer 4 had an amplitude of 0.15 ppm, and volunteers 1 & 2 was 0.1 ppm likely indicating a variation of inhalation volume.

Results for the 5 female volunteers are shown in Figure 2. All images are of the center acquired slice for each volunteer. The top two rows show the uncorrected (top) and corrected (bottom) magnitude images. The middle two rows show the standard deviation of the magnitude, as a percent of the average magnitude value, for uncorrected (top) and corrected (bottom). The bottom two rows show the PRF standard deviation in °C for uncorrected (top) and corrected (bottom) for water and glandular tissue. The incorporation of navigator phase

correction reduces the ghosting artifact in the magnitude images for all volunteers as shown by the reduction of the magnitude standard deviation. The standard deviation of magnitude within the breast was improved by an average factor ranging from 1.34 – 1.84 for the 5 volunteers. The PRF temperature precision was also improved. The standard deviation of PRF temperature estimates within glandular tissue was improved by an average factor ranging from 1.67 – 2.45 for the 5 volunteers. Values for each volunteer are displayed in Table 1.

The PRF temperature precision images for the phantom studies without MRgFUS heating comparing the cases with and without a volunteer above are shown in Figure 3. PRF temperature estimates vs. measurement number for an example single voxel near the center of each phantom for gelatin and pork are shown in figures 3b & c respectively. The average PRF standard deviation inside the gelatin and pork in each case is shown in Table 1.

Figure 4a shows the phantom results during MRgFUS heating. The PRF temperature map for the peak temperature voxel measurement, with and without simulated respiration for both the gelatin phantom and excised pork are displayed. The off-resonance due to respiration introduces greater error to the PRF measurements of the sonicated area as shown in Figure 4b and 4c.

DISCUSSION

These results have shown that image magnitude, phase, and PRF measurement artifacts caused by respiration induced B0 variations during 3D seg-EPI image acquisition can be reduced using two internal FID phase navigators that are acquired at the start and end of each echo-train. In the breast of the five volunteers there were improvements to the standard deviation of image magnitude by an average factor of 1.52, as well as the standard deviation of the temperature estimate by an average factor of 2.1. The inclusion of two internal FID phase navigators had a minimal impact on scan time by adding approximately 1 and 2 ms to TE and TR, respectively. The impact on scan time could be minimized further by acquiring fewer samples in the FID.

The field shift oscillation amplitudes of the breast volunteers before respiratory correction are in the same range and agree with findings by Peters *et al* and Bolan *et al* (4,24). In the phantom studies, the PRF precision after correction is comparable to the control cases without the volunteer above and therefore no respiratory artifacts. When heating with MRgFUS, the breathing with correction temperature estimates were nearly identical to the non-breathing case within the sonicated area in the gelatin phantom. The pork phantom also saw significant improvement to temperature estimates within the sonicated area. It is assumed the efficiency of this correction method decreases as the spatial variation of the off-resonance increases. Each receiver channel measures its own field shift for its region of sensitivity. This study focused on coronal acquisitions within the breast, which did not have motion within the imaging volume. This method may not be as effective with a transverse or sagittal slab in the breast, as respiratory motion (a different artifact) may be within the field of view, as well as possible decreased uniformity of the field shift over the sensitive region of each coil.

While there are several published techniques that used a navigator, many were focused on motion correction (11,12), or improving fMRI signal (17,18,25,26) and were either self-navigated or used a single navigator per TR. In our method, the phase changes due to echo-shifting during seg-EPI readouts (to reduce abrupt phase transitions in k-space) necessitated the use of two internal navigators to calculate the phase difference, instead of simply calculating the difference between navigators in subsequent TRs.

While 2D image acquisitions offer greater time resolution, which is very important in thermal therapy treatments, they often lack the spatial coverage needed to measure temperature in all areas that could be affected during treatment. The method presented here offers 3D images corrected for respiration artifacts, which could increase patient safety by allowing more volume coverage during treatment. The temporal resolution of the in vivo breast images and pork phantom studies was 10.3 s and the gelatin phantom was 7 s. These could be reduced using parallel acquisition methods (27–29), temporally constrained reconstruction (30,31), model predictive filtering (32,33), or decreased resolution (34). Phase navigators could be used with any of these methods as long as the phase correction was applied before the reconstruction algorithm and would theoretically not interfere with their effectiveness. A time reduction factor of 2 or 3 could potentially be sufficient for near real-time monitoring. This method could also theoretically be applied to 2D acquisitions, supplementing or removing the need for a multi-baseline library.

CONCLUSIONS

The described method provides promising results for implementing 3D monitoring of thermal therapy treatments while allowing free breathing. PRF temperature precision was improved using phase correction navigators in both phantom studies and in vivo breast. Ghosting artifacts in the magnitude and phase images were mostly removed. This technique could apply to other stationary targets with nearby motion in addition to breast.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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FIG. 1.

a) Modified segmented EPI sequence with flyback readout. FID phase navigator readouts are acquired before and after EPI readout. b) Example of phase during navigator readout for a single TR. c) Example of average phase difference estimate for 8 coils during one image acquisition. d) Example of phase variation about constant phase difference for 8 coils during one image acquisition. e) Example of field shift (in ppm) for one coil over 30 image acquisitions.

FIG. 2.

In vivo breast images for 5 volunteers (columns). Top two rows – Magnitude images uncorrected above and corrected below. Middle two rows – Standard deviation of magnitude uncorrected above and corrected below. Color scale displays percent of average signal value. Bottom two rows – Standard deviation of PRF temperature estimates in water and glandular tissue (°C) uncorrected above and corrected below.

FIG. 3.

Phantoms under non-heating conditions a) PRF temperature precision images for gelatin (top) and pork (bottom) comparing the no breathing (left), breathing without correction (middle), and breathing with correction (right) cases without FUS heating. c) PRF temperature change of a single voxel in gelatin phantom for all 3 cases. d) PRF temperature change of a single voxel in pork phantom for all 3 cases. Time lapse videos of temperature in gelatin and pork for all 3 cases under non-heating conditions is available as supporting information in the online version of this publication.

FIG. 4.

MRgFUS phantom experiments. a) PRF temperature change images at the time of the peak temperature for gelatin (top) and pork (bottom) comparing the no breathing (left), breathing without correction (middle), and breathing with correction (right) cases with MRgFUS heating. b) PRF temperature change of a single voxel in gelatin phantom for all 3 cases. c) PRF temperature change of a single voxel in pork phantom for all 3 cases. Time lapse videos of temperature in gelatin and pork for all 3 cases under FUS heating conditions is available as supporting information in the online version of this publication.

Table 1

Comparison of all results

Volunteer #	Average Magnitude Std Improvement Factor	Maximum Magnitude Std Improvement Factor	Average PRF Std Improvement Factor	Maximum PRF Std Improvement Factor
Breast 1	1.36 ± 0.52	12.5	1.67 ± 1.14	27.4
Breast 2	1.84 ± 0.97	13.7	2.45 ± 2.67	64.2
Breast 3	1.40 ± 0.52	5.4	2.14 ± 1.86	25.0
Breast 4	1.34 ± 0.37	5.0	2.16 ± 1.63	23.3
Breast 5	1.70 ± 0.58	6.7	1.92 ± 1.59	25.2
Phantom Results				
Case	Gelatin PRF Std °C		Pork PRF Std °C	
No Breathing	0.17 ± 0.09		0.25 ± 0.10	
Breathing without Correction	1.36 ± 0.28		1.01 ± 0.66	
Breathing with Correction	0.39 ± 0.23		0.45 ± 0.32	