Published in final edited form as:

Magn Reson Med. 2014 August ; 72(2): 563–569. doi:10.1002/mrm.24932.

Phase reconstruction from multiple coil data using a virtual reference coil

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Abstract

Purpose—This paper develops a method to obtain optimal estimates of absolute magnetization phase from multiple-coil MRI data.

Methods—The element-specific phases of a multi-element receiver coil array are accounted for by using the phase of a real or virtual reference coil that is sensitive over the entire imaged volume. The virtual-reference coil is generated as a weighted combination of measurements from all receiver coils. The phase-corrected multiple coil complex images are combined using the inverse covariance matrix. These methods are tested on images of an agar phantom, an in vivo breast, and an anesthetized rabbit obtained using combinations of four, nine, and three receiver channels, respectively.

Results—The four- and three- channel acquisitions require formation of a virtual-reference receiver coil while one channel of the nine-channel receive array has a sensitivity profile covering the entire imaged volume. Referencing to a real or virtual coil gives receiver phases that are essentially identical except for the individual receiver channel noise. The resulting combined images, which account for receiver channel noise covariance, show the expected reduction in phase variance.

Conclusions—The proposed virtual reference coil method determines a phase distribution for each coil from which an optimal phase map can be obtained.

Keywords

MRI; phased array; image combination; noise correlation

INTRODUCTION

Although most MRI applications require only magnitude images, there are a growing number that require measurements of phase and phase changes, as indicated by some recent examples in phase contrast flow velocity measurements (1–4) and proton resonant frequency (PRF) temperature measurements (5–9). For a single receiver coil, the phase of the measured signal includes the phase of the magnetization added to the phase of the receiver coil sensitivity. When multiple RF coils are used, each coil could be used for a determination of

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the magnetization phase, but this would require removal of the receiver coil phase. For situations where the complex coil sensitivities (the relative phase and amplitude of each receiver coil throughout the imaging volume) are known, the method of Roemer, which is equivalent to the parallel imaging method, SENSE, with reduction factor equal to one, can be used to obtain the optimal combined signal magnitude and phase (10,11). Because the complex coil sensitivities are usually not known, various methods for obtaining sensitivity estimates from the complex images have been evaluated (12,13). These methods typically result in optimal magnitude images, but phase information is distorted or lost.

When only the difference in phase between an initial and final time is desired, the coil phase can be removed by first taking the difference in phase between time frames in each coil and then combining the resulting phase differences (14). Bernstein weighted the phase differences by the square of the individual coil signal magnitudes (15) but did not account for noise correlations. Thunberg et al. and Lu et al. both considered noise correlations and demonstrated that for regions of low SNR, the SENSE-like optimal reconstruction that takes into account the noise covariance resulted in reduced phase variance over the weighted mean methods (16,17). The phase variance reduction was small and decreased as SNR increased.

When phase subtraction is not possible and the receiver coil phase is not known, some method must be used to estimate and remove the relative receiver coil phase before phase measurements are combined. Self-calibration methods developed from parallel imaging techniques can be used to obtain coil sensitivities (18,19). The adaptive combine method uses localized estimates of the coil sensitivities to combine the complex images from which image phase can be estimated (20). These methods work well for magnitude images, but tend to experience inconsistencies and failures in phase determination (14,21,22) such as at points where the phases between the channels vary sufficiently to cause signal cancellation and points of undefined combined phase. Other methods for combining phases from multiple coils include phase filtering, where a high pass filter is used to eliminate the slowly varying receiver phase (after phase unwrapping of each receiver channel) (23) and multiple coil combination after adding a constant phase offset to each channel to equalize the phase at a point and yield approximate phase equalization throughout the image volume (24). The latter technique shows promise, but loses SNR away from the point of phase equalization.

This paper extends the phase equalization technique (24) to create a virtual reference receiver coil to which the measurements of each receiver coil are referenced and then combined to obtain an optimal phase distribution estimate. The method is demonstrated using multiple coil scans of an agar phantom, a human breast, and a rabbit. Comparisons are made with the adaptive combine method.

THEORY

Phase difference from optimal coil combine

Roemer et al. demonstrated that the signals from multiple coils can be combined in an optimal way to provide a resulting complex image with maximum SNR at every image voxel (10). When phase is determined from the real and imaginary components of the signal, the variance in phase is directly proportional to $1/\text{SNR}^2$ and therefore the Roemer method,

which combines the signals to maximize SNR, will result in the minimum variance in phase. With the transverse magnetization defined as $m(x)^- = m_x + im_y = |m|e^{i\theta}$ and the complex sensitivity of the j^{th} coil defined as $b_j(x)^- = |b_j(x)|e^{i\vartheta_j(x)}$, the measured signal, including added random noise, n_j , can be defined as $p_j(x)^- = b_j m + n_j$. Roemer demonstrated that the combined image with maximized SNR could be obtained using a spatially dependent set of complex weights ($\alpha_j(x)^- = a_j e^{i\vartheta_j}$):

$$P(\vec{x}) = \sum_{j=1}^{N_c} \alpha_j(\vec{x}) p_j(\vec{x}) \quad (1)$$

where:

$$\alpha_j = \sum_{q=1}^{N_c} \frac{1}{\lambda} b_q^* R_{q,j}^{-1}, \quad (2)$$

N_c is the number of receiver coils (channels), R is the noise covariance matrix and λ is a normalization factor. When two images are obtained ($^1P, ^2P$), the optimal estimate of object phase difference can be obtained as (16,17):

$$\Delta\theta = \text{angle}(^2P^1P^*), \quad (3)$$

where $\text{angle}(A+iB) = \arctan(B/A)$. Methods that perform phase subtraction after combining individual coil images into a composite image (17) require knowledge of the phase of the individual coil sensitivities. Coil sensitivity phases are not required if phase subtraction is performed before combining individual coil images (16).

When the coil sensitivities are not known, there is a very simple method to obtain the optimal phase difference between two sets of noisy measurements, 1p_j and 2p_j :

$$\Delta\theta = \text{angle} \left(\sum_{j,q=1}^{N_c} ^2p_j \frac{1}{\lambda} R_{j,q}^{-1} (^1p_q)^* \right). \quad (4)$$

This relation can be obtained by defining:

$$\psi_j = \text{angle}(p_j) = \vartheta_j + \theta + \eta_j, \quad (5)$$

where ϑ_j is the coil phase, θ is the true tissue phase and η_j is a random component due to thermal noise in the individual RF coil measurement. At a later time the tissue phase may have evolved by $\Delta\theta$ and the random component will have changed by $\Delta\eta_j$ giving:

$$^2\psi_j = ^1\psi_j + \Delta\theta + \Delta\eta_j \quad (6)$$

Eq. (4) arises directly from using $^1\psi_j$ and $^2\psi_j$ for the phases of 1p_j and 2p_j and assuming η_j is small. Eq. (4) does not require coil sensitivities, and uses the optimal combination of the relative phases of each coil to directly find the optimal phase difference.

Absolute phase from a single image

To obtain a consistent measurement of absolute phase from the multiple coil images of a single acquisition when the complex coil sensitivities are not known, the adaptive combine method has been proposed and is in current use (20). In this method the phase of the coil sensitivity is estimated from a local matched filter and the image SNR obtained is near optimal. The adaptive combine method works well when there is reasonable homogeneity throughout the imaged volume. However, where there are differences in tissues (fat/water, etc.) local variations in phase can interfere with the determination of the RF coil phase.

In this work a direct approach for phase determination is proposed. When the set of measurements includes a receiver coil (call it coil “r”) that is sensitive over the entire imaged volume, the phase difference between the k^{th} and r^{th} coil images eliminates the tissue phase, leaving the difference between coil phases and the difference in the random component:

$$e^{i\Delta_{rk}} = e^{i\psi_r} e^{-i\psi_k} = e^{i(\vartheta_r - \vartheta_k + \eta_r - \eta_k)}. \quad (7)$$

Because the coil phase varies slowly while the thermal noise is uncorrelated from voxel to voxel, an appropriate low pass filter will suppress the thermal noise while having minimal effect on the coil phase difference: $\langle e^{i\Delta_{rk}} \rangle = e^{i(\vartheta_r - \vartheta_k)}$. Here $\langle \rangle$ implies low pass filtering (12). Adding this filtered difference back to the k^{th} RF coil image gives:

$$e^{i\hat{\psi}_k} = e^{i(\vartheta_k + \theta + \eta_k)} e^{i(\vartheta_r - \vartheta_k)} = e^{i(\vartheta_r + \theta + \eta_k)}, \quad (8)$$

where $\hat{\psi}_k$ is the phase map of coil k with the phase of the k^{th} coil sensitivity replaced by that of the reference coil. With this conversion the phase distribution is the sum of the desired object phase, the random phase error of the original images, and the phase of the reference coil.

This reference coil phase is combined with the original measurements: $p_k^{\hat{}} = p_k e^{-i\hat{\psi}_k} = p_k e^{-i(\vartheta_r + \theta + \eta_k)}$. For each coil, k , the pixel value differs from the original value by an added phase term that is independent of k except for the noise term. Using this $p_k^{\hat{}}$ in Eq. (4) gives:

$$\langle \hat{\psi} \rangle = \text{angle} \left(\sum_{j,k=1}^{N_c} p_j \frac{1}{\lambda} R_{j,k}^{-1} (\hat{p}_k)^* \right) = \vartheta_r + \theta + \text{angle} \left[\sum_{j,k=1}^{N_c} p_j \frac{1}{\lambda} R_{j,k}^{-1} p_k^* e^{i\eta_k} \right]. \quad (9)$$

To the extent that the extra coil specific noise term is small, the last term on the right is just the optimal sum of squares of the coil measurements, and will have negligible phase.

The phase referencing of Equations (7) and (8) works well only if the reference coil has sensitivity covering the entire volume of interest. In many phased arrays designed for

parallel imaging, none of the individual elements have sensitivity that sufficiently overlaps all of the other RF coils. To overcome this limitation, a virtual reference coil, v , can be formed as a linear combination of the phase of all of the receiver coils:

$$V = \sum_{j=1}^{N_c} w_j p_j = \sum_{j=1}^{N_c} |w_j| e^{i\varphi_{refj}} e^{i(\vartheta_j + \theta + \eta_j)}. \quad (10)$$

When the weights, w_j , are chosen based upon the SNR of p_j this expression can result in a virtual reference coil that has higher SNR throughout the imaging volume than any one single coil. To avoid loss in SNR in the virtual reference coil, the phase of the weights, φ_{refj} , must also be chosen to minimize the phase difference between the weighted signal measurements throughout the image volume. Because the coil phases vary slowly spatially, φ_{refj} can be chosen to null the phase of p_j at a single point and the phase difference between the weighted coil measurements will be small throughout the image volume (24).

$$\varphi_{refj} = -\psi_j(x_o). \quad (11)$$

To reduce measurement error the average phase can be determined:

$$\varphi_{refj} = -\text{angle} \left(\frac{\sum_x p_j(x)}{\sum_x |p_j(x)|} \right), \quad (12)$$

where x are points around the point x_o or even throughout the entire image volume. The common point, x_o , can be determined from the region where there is maximum overlap of the signals from each coil. If there is no region where sufficient SNR from all coils overlap, a daisy chain of referencing can be performed such that the first pair of coils are adjusted based upon where their signals overlap, and then subsequent coils are adjusted to match their phase with one of the prior coils at their point of maximum overlap.

Including phase referencing, the weights are then obtained as: $w_j = |p_j| e^{-i\varphi_{refj}} / \sum_k |p_k|$ and the measurements are combined using Eq. (10) to obtain the virtual coil. The phase of the virtual reference coil is:

$$\text{angle}(v) = \theta + \vartheta_{virtual}, \quad (13)$$

where

$$\vartheta_{virtual} = \text{angle} \left(\sum_j \frac{|p_j|}{\sum_k |p_k|} e^{i(\vartheta_j - \vartheta_j(x_o) - \theta(x_o) + \eta_j(x_o))} \right). \quad (14)$$

With $\vartheta_{\text{virtual}} = \vartheta_r$ the original measurements, p_k , are then referenced to the virtual reference coil as in Eq. (8) and the combined phase is determined using Eq. (9). Thus, although the object phase, θ , is not obtained directly, the reconstructed complex images have a stable phase that includes θ . From these images with stable phase, accurate phase difference images can be obtained.

Thus absolute phase can be determined from multiple coil measurements by: 1) Creating a virtual reference coil that is sensitive over the entire imaged volume using a complex weighted sum of the individual coil measurements; a) the phase of the weights rotate the data from each coil such that all coils have the same, hopefully zero, phase where the coil sensitivities have maximum overlap; b) the weight magnitudes vary with the relative local SNR of each coil; 2) Replacing the individual coil phase with the virtual reference coil phase while maintaining the object phase and image noise of each coil by a) taking a low pass filter of the phase difference between each coil and the reference coil; b) subtracting this low-pass-filtered phase difference from the individual coil measurements; 3) using the inverse noise covariance to combine the phase referenced complex measurements from all coils.

METHODS

Image acquisition

Three sets of images were acquired on a Siemens TIM Trio MRI scanner (Siemens Health Care AG, Erlangen, Germany) to illustrate the points of this paper. First, a cylinder filled with agar was scanned twice using a 2D gradient echo sequence with four receiver coils, two pads of two coils each, placed around the cylinder. MRI parameters were TR/TE = 50/2.8 ms, Matrix 192×174. Second, a breast of one female volunteer was scanned using a 3D gradient echo pulse sequence in an 11 channel RF coil (25) used in conjunction with a system for MRI-guided high intensity focused ultrasound (HIFU) of the breast (26). The RF coil array around the breast chamber is shown in Figure 1. The coil consists of a single loop that sits against the chest wall and a 10-channel ladder-array that surrounds the treatment cylinder. The imaging parameters were: BW = 1149 Hz/pixel, TR = 40 ms, TE = 2.61, flip angle = 15°, matrix = 256×184×32 with 25% slice oversampling, 1mm isotropic resolution. Two of the 10 RF ladder channels were eliminated because of artifacts. Third, an anesthetized white New Zealand rabbit was scanned using an interleaved 2D GRE sequence with 3 receiver coils and TR/TE = 98/2.9 ms, BW = 446 Hz/pixel, 320mm FOV, and 1 × 1 × 3 mm voxels. The noise covariance matrices were calculated based upon measurements at the edge of k-space where noise greatly dominated any signal. All animal experiments were approved by the Institutional Animal Care and Use Committee.

Data analysis

Four- and three- channel coil arrays—For the agar phantom and rabbit studies, an image consisting of the minimum intensities of the receiver channels at each position was calculated, and a point of maximum intensity in this image was selected to determine the phase offset, $\phi_{\text{ref},j}$, to null the phase of each receiver coil at that point. A virtual reference coil was calculated as:

$$v = \frac{\sum_j e^{-i\varphi_{refj}} p_j}{\sum_k |p_k|}. \quad (15)$$

The phase difference between the virtual reference coil image and each coil image was calculated, low pass filtered and used to adjust the phase of that receiver coil image. The optimal combined image phase was then calculated using Eq. (9).

Breast coil array—For the breast study, the phase of each of the coils around the treatment cylinder is referenced to the phase of the single chest loop coil. This method works because the chest loop sensitivity is sufficiently high over the entire imaging volume.

RESULTS

Images obtained from the cylinder are shown in Figure 2. Without initial phase offset adjustment, the different phases of each coil shown in Figure 2a result in the wrap-around phase discontinuity shown in Figure 2b. Although this phase image is obtained using the adaptive combine algorithm on the MRI scanner, similar results would be obtained using any algorithm because of the large (nearly 180°) phase differences between the individual receiver coils. The rotated phase maps shown in Figure 2c result in the virtual reference coil phase shown in Figure 2d. The individual coil phases mapped to the virtual reference coil (Figure 2e) are then used to obtain the optimal phase reconstruction (Figure 2f). Neglecting phase spikes, which occur without the corrections proposed in this paper, the standard deviation of the phase is 0.0735 radians for the adaptive combine and 0.064 for the optimal method. The standard deviation of the virtual coil phase before low pass filtering is 0.0775.

Images from the multiple receiver-coil breast GRE study are shown in Figure 3. The magnitude images (Figure 3a) indicate the variation in coil sensitivity within this specific slice. The phase images (Figure 3b) show phase wraparound in many of the images. The differences between the image phase of each channel and the corresponding phase of the chest loop coil image (Figure 3c) eliminates tissue phase. The chest loop coil was used as reference instead of a virtual reference coil for all receivers because it had sufficient uniformity across the image volume. A 21-point (21×21) median filter applied to smooth the coil sensitivity difference (Figure 3d) leaves only the difference in phase distribution between the corresponding pairs of receivers. Adding the filtered phase difference back to each receiver channel phase (Eq. (8)) results in the image phase in each channel being corrected to the base phase of the chest loop channel (Figure 3e). The images, consisting of the individual channel magnitudes and the corrected phase are then combined using Eq. (9) to obtain the combined image shown in Figure 4b. Phase images obtained using the adaptive combine method (Figure 4c) did not show a branch line because the original coil images had similar phases but do show some subtle differences, including some blurring, that may be due to an adaptive combine matched filter. For the rabbit study, a comparison of the phase map with branch line and singular phase point from the adaptive combine reconstruction with the improved phase map from the virtual reference coil method are shown in Figure 5.

DISCUSSION

This work presents a simple method for obtaining the optimum phase or phase difference from multiple receiver coil images when the coil sensitivities are not known. The method is based on the observation that an optimal estimate of the subject phase difference from multiple coil images can be obtained by incorporating the two sets of multiple coil images directly in the inverse covariance sum of squares formalism. This work then uses this observation to develop a method to obtain the optimal estimate of combined phase from a single set of multiple coil images.

The experiment with the agar cylinder demonstrated the importance of eliminating large (180°) phase differences between the multiple receiver coils before calculating the reference (virtual) image phase. Without the phase adjustment, when these coils are combined to form a virtual or reference image, the rotating phase can cause destructive interference near the point of greatest overlap as well as a branch line where the virtual phase angle passes through $+\pi$. Although the effect of the branch line can in principle be reduced when using complex subtraction, the loss in image SNR results in increased phase variance and the end of the branch line becomes a point of undefined combined phase, resulting in bad measurement conditioning. Fortunately, this problem can be almost completely eliminated throughout the image volume by appropriate choice of the phase of the weights.

The similarity between Figure 2d and 2f would suggest that the simpler method of Hammond et al. (24) could be used to obtain the combined phase distribution. The differences are subtle but important. First, the coils that contribute to the virtual coil phase only have the same phase at a single point in the 3D phase distribution. In the proposed method, all measurements are referenced to the virtual coil phase and there is no signal loss due to phase differences away from the point of zero phase. Second, the weights for generating the virtual coil phase do not include the inverse noise covariance weighting.

Although it is not always large, an improvement in phase measurement precision (reduction in phase variance) should always occur when optimal coil combination with the inverse covariance matrix is used. In the example given, an improvement of nearly 16% was observed, and the branch line and singular point were completely eliminated. We did not do more experiments to demonstrate this fact, but it has been well documented by others (16,17).

The proposed method does have some limitations. The definition of the virtual reference coil assumes that the object phase at the reference point is stable. Variations in the reference point or phase at the reference point as a function of time will result in a constant phase offset throughout the image volume. In the absence of motion, this variation can be avoided if the virtual reference coil is determined once and used for all subsequent images in time. Localized phase changes due to heating, flow, or physiological motion will result in potential errors. Finally, if there is a large range of receiver coil sizes, it is possible that the simple method of phase referencing may not be sufficient to eliminate branch lines and more complicated phase adjustments may be necessary.

CONCLUSION

The method of virtual reference coil as presented allows the combined image phase to be determined from an optimal combination of the individual receiver coil measurements. The method reduces the potential for branch lines and singular points in the resulting reconstructed phase image and results in phase measurements with minimum variance.

Acknowledgments

Grant Support: NIH R01 CA134599, R01 EB013433, Siemens Medical Solutions, Ben B. and Iris M. Margolis Foundation

The authors acknowledge contributions from Mahamadou Diakite, Henrik Odeen, and other members of the University of Utah HIFU research group. This work has been supported by NIH grants R01s CA134599, EB013433, DC011497, as well as grants from Siemens Medical Solutions, the Cumming Foundation, the Ben B. and Iris M. Margolis Foundation, and the Mark H. Huntsman Endowed Chair.

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Figure 1. The breast imaging and treatment cylinder with RF coils used in this study. Shown with (a) and without (b) treatment platform. The chest loop coil location at the top of the treatment cylinder is indicated by an arrow.

Figure 2.

- a) Phase images from a coronal acquisition of the agar cylinder with the four receiver coils positioned around the cylinder phantom.
- b) Combined phase image using adaptive combine.
- c) Phase images after rotation by ϕ_{refj} (Eq. (11)) before combining to create the virtual coil.
- d) Phase of reference (virtual) coil image (Eq. (15)).
- e) Phase after referencing to phase of virtual coil image (Eq. (8)).
- f) Phase of optimally combined complex images (Eq. (9)).

Figure 3.

Magnitude and phase images from each of the nine receiver coils of a sagittal 3D GRE study of a human breast scan, located approximately mid breast. See text for acquisition details. a) Magnitude image. The scale is equivalent for all images.

b) Phase distribution of the same slice for all 9 receiver channels.

c) Phase difference between each individual coil and chest loop (9th) coil (lower right in part b).

d) Smoothed phase difference (median filter). This smoothed phase difference is used as a correction phase to transform to the base phase of the chest loop coil.

e) Phase of all 9 receiver channels after correction for phase difference relative to the chest loop coil. Note all phase images are scaled from $-\pi$ to π .

Figure 4.

Phase (a, b, and c) and magnitude (d) images obtained from the data of Figure 3. Images are from the chest loop coil (a), the combined images using Roemer's equation with phase referencing to the chest loop coil (b), and from the adaptive combine algorithm on the MRI scanner (c). (d) Magnitude image corresponding to (b).

Figure 5.

Phase images obtained from a HIFU study of a rabbit reconstructed with adaptive combine (a) and the virtual reference coil method (b). Coil placement results in a signal null (dashed arrow). The branch line and singular point in (a) (solid arrow) are eliminated in (b).