Optimal Method for Obtaining Dynamic Phase Images from Multi-Channel EPI Data

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INTRODUCTION: Although MRI predominately uses magnitude information for the formation of images for most techniques, there are several MRI methods that rely on phase information (e.g. thermometry[1], velocity mapping, water-fat imaging, and susceptibility-weighted imaging). In BOLD-weighted EPI imaging, the phase information has been used to separate intra- and extra-vascular BOLD signal[2] and to track the state of the field homogeneity and respiration over time[3,4]. From a signal processing standpoint, the phase and magnitude have a clear meaning in the context of phase sensitive signal detection (i.e. the phase and amplitude of the MR signal can be used to enhance SNR). However, since the phase of the detected signal can be different for each RF channel, the concept of phase at the level of the reconstructed image becomes somewhat confusing in the situation of multiple element coils. The combination of multiple coil images is generally performed without any weighting, which is not optimal[5]. Additionally, the MR signal has an arbitrary phase offset for each voxel and channel due to the relative location of each coil element with respect to each voxel. Finally, due to the bounded nature of the arctangent, calculations using phase images can be complicated. We implement a reconstruction method that results in an optimal determination[5] of complex, phase sensitive data that correctly weights and combines multi-channel data. As an example of the method, we demonstrate the technique by applying a Maxwell correction to the combined data and produce a corrected phase image.

COIL COMBINATION: Multiple RF receive coils can be used to increase SNR, but each coil is sensitive only to a smaller region of the full field of view. Equations 1-4 describe the typical mathematical descriptions of MR magnitude and phase. The first step is to obtain the magnitude of the complex signal (where Re and Im are the orthogonal components of the complex signal) at each voxel for each coil from the vector length as in Eqn 1. For multiple coils, these are combined to get image signal S(x,y) in Eqn 2 by weighting by the local sensitivity of the coil to that voxel. A common approximation of the sensitivity is the magnitude at each voxel for each coil, in which case W(x,y,c) = Mag(x,y,c), which is the special case of sum-of-squares coil combination, but W(x,y,c) can be any weighting factor. For the applications requiring phase information, polarity must be preserved, so one solution is to sum the complex channels without decomposing to magnitude/phase or weighting by sensitivity, which is shown in Eqn 3. Phase is the angle between the combined complex signals at each voxel as in Eqn 4.

However, shown in Figure 1 is a graphical description of how each coil can see an arbitrary offset due to spatial location w.r.t. each voxel. A particular voxel may give MR phases that have opposing directions as seen by two separate coils, resulting in a diminished SNR of the resultant phase measurement (although in practice the differences are not that large). This can be handled by referencing all subsequent MR complex signal measurements from each coil and voxel to the first measured complex signal. Eqn 5 shows complex conjugate multiplication of the current MR complex signal with the first acquired complex signal (denoted by Re_{ref} and Im_{ref}). This reduces the need for phase-unwrapping after reconstruction.

The referenced complex signal can be weighted by either the magnitude determined in Eqn 1, or any other coil-specific weighting factor that captures the coil sensitivities, which is shown in Eqns 6 and 7 for the orthogonal components of the complex signal. This allows one to retain the polarity and weight the images in the same manner as the magnitude image to improve the SNR of the resultant phase image. Before calculating the phase, distortion correction must be applied to the complex data. Interpolation of phase data does not give a meaningful result, except approximately, because phase data cannot be added in the same way without taking precautions. Finally, after any distortion correction or other interpolation is performed, the phase may be calculated through the arc-tangent of the referenced, combined complex data as in Eqn 8. We compare MR phase images from **1)** naively combined and not referenced, ala Eqn 4), **2)** referenced but naively combined, **3)** optimally combined but not referenced, **4)** referenced and optimally combined.

Fig 1: MR signal vectors at 4 coils

$$\text{Eqn1: } Mag(x,y,c) = \sqrt{Re(x,y,c)^2 + Im(x,y,c)^2}$$

$$\text{Eqn5: } Re' + i*Im' = \sqrt{(Re + i*Im) \bullet (Re_{ref} + i*Im_{ref})^*}$$

$$\text{Eqn2: } S(x,y) = \sqrt{\sum_c Mag(x,y,c) \times W(x,y,c)}$$

$$\text{Eqn6: } Re_w(x,y,c) = \sum_c Re'(x,y,c)*W(x,y,c)$$

$$\text{Eqn3: } Re(x,y) = \sum_c Re(x,y,c)$$

$$\text{Eqn7: } Im_w(x,y,c) = \sum_c Im'(x,y,c)*W(x,y,c)$$

$$\text{Eqn4: } Phase(x,y) = arg(Re(x,y) + i*Im(x,y))$$
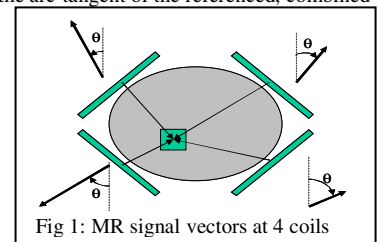
$$\text{Eqn8: } Phase_w(x,y) = arg(Re_w(x,y) + i*Im_w(x,y))$$

DATA: The following scan was performed using a 12-channel RF receive head coil on a Siemens 3T Trio with a blipped EPI pulse sequence. Scan1: whole-brain EPI 130 volumes of 18-4mm thick axial slice TE/TR/flip = 29ms/2000ms/80°, matrix=128×128, 6/8 partial Fourier factor, 256mm ×256mm FOV, BW=250KHz. The data was processed as above into four phase datasets for each of the two scans, but datasets that were not referenced (phase sets 1 and 3 were phase-unwrapped due to the many pi-bounces across the images[6]. The data were compared by correlation with the mean masked slice phase timeseries, with the hypothesis being that much of the phase deviation is due to respiration and therefore global across the slice, and reduction in correlation will be due to non-optimal methods introducing noise into phase measurement.

RESULTS: The mean correlation of every voxel's phase timeseries with the mean slice timeseries is histogrammed in Fig 2a, where it is clear that referencing provides much benefit, but that optimal combination produces a negligible effect. Fig 2b shows the ratio of correlation to mean for dataset 4 to dataset 3 across the image, which shows there is a net increase in correlation of between 1 and 2%. The ratio of dataset 4 to dataset 2 however shows a smaller increase in correlation (tenth of a percent), which is higher in regions of inhomogeneity. The qualitative difference between the phases calculated by the four methods is shown in Figure 3. Weighted combination produces minimal (but noticeable) change between Fig 3b and 3c. An unwrapping algorithm was used to prepare the image in Fig 3d from that in Fig 3b (which appears qualitatively the same as the unwrapped Fig 3c). The improvement due to referencing is evident in Fig 3e.
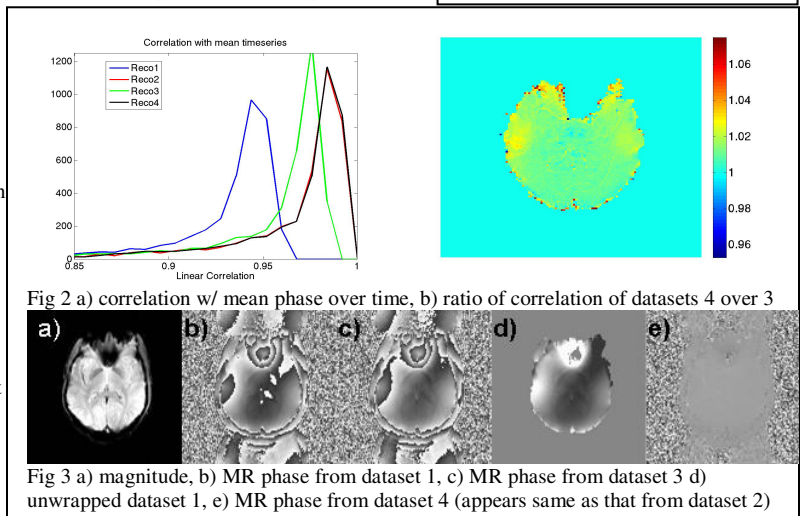

Fig 2 a) correlation w/ mean phase over time, b) ratio of correlation of datasets 4 over 3
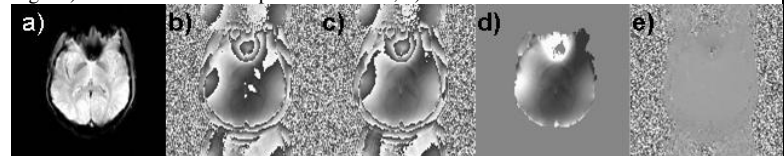

Fig 3 a) magnitude, b) MR phase from dataset 1, c) MR phase from dataset 3 d) unwrapped dataset 1, e) MR phase from dataset 4 (appears same as that from dataset 2)

DISCUSSION: The primary benefit of phase referencing is the removal of the need to do phase unwrapping for every image and a small increase in signal to noise ratio. The SNR improvement may be greater in coils with a larger system electronics phase offset. The dynamic evolution of the field homogeneity can be determined from referenced phase information[3] and used to improve volume-to-volume distortion correction by combining with a fieldmap determined from the initial phase.

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