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The use of Fourier-domain analyses for unwrapping phase images of low signal-to-noise ratio

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Abstract

Purpose: We report a new post-processing procedure that uses Fourier-domain data analyses to improve the accuracy and reliability of phase unwrapping for MRI data of low signal-to-noise ratio (SNR).

Methods: The developed method first identifies the Fourier-domain energy peak locations corresponding to different image-domain areas, from which a robust measurement of image-domain phase gradients can be obtained even for MRI data of low SNR. The phase gradient information measured from critical brain regions, using the above-mentioned Fourier-domain analysis, is then combined with the conventional temporal-domain or spatial-domain phase unwrapping procedure to remove phase wraps. The developed method was tested with MRI data obtained from 30 healthy adult volunteers, and its performance was quantitatively evaluated.

Results: The developed Fourier-domain analysis could robustly quantify image-domain phase gradients even for MRI data with low SNR (e.g., $\text{SNR} \approx 2$). Experimental results show that the Fourier-domain analyses could further reduce phase wrap artifact in data produced by the conventional temporal-domain or spatial-domain phase-unwrapping procedures.

Conclusion: Our results demonstrate that the developed phase-unwrapping method can reduce residual phase wraps resulting from conventional procedures in critical brain regions (e.g., near the air-tissue interfaces), and should prove valuable for studies that require accurate measurements of MRI phase values, such as quantitative susceptibility mapping (QSM), B_0 field mapping, and temperature mapping.

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Keywords

MRI; Phase unwrapping; Field inhomogeneities; Fourier-domain analysis

Introduction

Measurement of MRI image-domain phase values is important for many different types of studies ranging from mapping magnetic field inhomogeneities (1), to echo-planar imaging (EPI) distortion correction (1,2), to quantifying blood flow velocity (3), to characterizing chemical shift effect (4), to temperature mapping (5), to susceptibility weighted imaging (6) and quantitative susceptibility mapping (QSM) (7). Since the phase values derived from imaging data wrap around between $-\pi$ and π , they need to be unwrapped (i.e., by adding integer multiples of 2π) to reveal true phase values.

Conventional phase unwrapping procedures are based on temporal-domain processing (8–11), spatial-domain processing (6,12–18), or Laplacian based operation (19,20). The temporal-domain procedures typically analyze gradient-echo MRI data sets corresponding to different echo times (TEs), on a voxel-by-voxel basis, aiming to reveal true phase values ($\Delta\theta$) that are related to B_0 frequency offset ($\Delta\omega$) and echo time spacing (ΔTE) according to $\Delta\theta = \Delta\omega \times \Delta TE$ without wrap artifact. However, as pointed out by Dagher et al. (9) and Robinson et al. (10), the temporal-domain phase unwrapping may not work robustly when the phase increment is $> \pi$ across two subsequent TEs in data obtained with sparsely sampled TEs. Although the true phase value can be much more robustly measured from multiple optimally chosen TEs with non-equal spacing (9,10), it remains challenging to unwrap MRI phase values in critical brain regions obtained with typical multi-echo gradient-echo MRI protocols (e.g., sampling 8 equally-spaced TEs between 4 and 50 msec in typical QSM scans).

In contrast to temporal-domain phase unwrapping, the spatial-domain phase unwrapping procedures are applicable to MRI data obtained with a single TE. With an assumption that the true phase difference between two neighboring pixels is smaller than π , spatial-domain phase unwrapping can be achieved by 1) measuring the phase difference between neighboring pixels, followed by 2) adding or subtracting 2π to phase values of neighboring pixels where the phase differences are greater than π due to phase wrap. However, it is not always easy to accurately unwrap the phase values, particularly when 1) the signal to noise ratio (SNR) is low, and 2) the phase gradient is large (e.g., $> \pi$ between two nearest neighboring voxels).

Laplacian-unwrapping is based on an observation that, assuming the unwrapped phase is sufficiently differentiable, its local derivatives are most similar to the derivatives of the wrapped phase. Implemented with Fourier analysis operator and mirror symmetrization of data (20–22), the Laplacian-unwrapping procedure can effectively remove spatial discontinuity in phase values, and largely preserves phase variation patterns (with a performance much less dependent on SNR as compared with spatial-domain phase unwrapping procedures). However, the difference between wrapped and Laplacian-

unwrapped phase values may not be integer multiples of 2π . As a result, the Laplacian-unwrapping procedure does not produce quantitatively accurate phase values, especially in brain regions affected by pronounced susceptibility field gradients.

To address the above-mentioned challenges in conventional phase unwrapping procedures, in this paper we report a new procedure that use Fourier-domain analyses to improve the accuracy and robustness of phase unwrapping for imaging data of low SNR.

Theory

Impact of noise on image-domain phase values and their Fourier-domain representation

A 1D mathematical simulation (Figure 1) illustrates the impact of noise on image-domain data and the corresponding Fourier-domain representation, in the presence of image-domain phase gradients. Figures 1a and 1b show image-domain phase values of a 1D mathematical phantom in the presence of $-8 \times 2\pi$ radians/FOV phase gradients and $22 \times 2\pi$ radians/FOV phase gradients, respectively. The magnitude signals of this 1D phantom equal 1 in all image-domain locations (data not shown).

According to the shift theorem, the Fourier transform (FT) of complex-valued signals $\rho(x)$ that originate from an image-domain region will be shifted by Δk when there exists an image-domain phase gradient ($\exp(i 2\pi \Delta k x)$), as shown in Equation 1, where $S(k_x)$ is the Fourier transform of $\rho(x)$, i.e., $S(k_x) = FT[\rho(x)]$:

$$S(k_x - \Delta k) = FT[\rho(x) \exp(i 2\pi \Delta k x)] \quad [1]$$

The Fourier-domain data corresponding to the image-domain data of Figures 1a and 1b are shown with blue line (peaked at $k_x = 8$) and black line (peaked at $k_x = -22$), respectively, in Figure 1c. Figures 1a to 1c illustrate that image-domain phase gradients can be computed with either image-domain analyses (after spatially unwrapping the phase values shown in Figures 1a and 1b) or Fourier-domain energy peak identification (in Figure 1c), in the absence of noise.

As shown in Figures 1d and 1e, after mathematically adding Gaussian noise (with its standard deviation equal to 1) to both the real- and imaginary components of the data shown in Figures 1a and 1b, respectively, the phase gradient patterns are significantly distorted, making it challenging to unwrap these phase values of low SNR. Although the image-domain phase gradient patterns are significantly distorted by noise (Figures 1d and 1e), the corresponding Fourier-domain echo energy peak displacement (i.e., blue and black lines in Figure 1f) can still be reliably quantified. The Fourier-domain representation of phase gradients has a higher tolerance to noise, as compared with the image-domain data, because signals from a group of voxels (with similar image-domain phase gradients) all contribute coherently to the same Fourier-domain echo energy peak.

The phase gradients shown in Figures 1a and 1b are spatially uniform, and thus the corresponding Fourier-domain energy peaks are visually identifiable in Figures 1c and 1f.

However, in brain MRI data, the phase gradients could be non-uniform and nonlinear within a large image-domain region. In the next subsection, we will present a Fourier-domain analysis method that quantifies local phase gradients within a small image-domain region.

Quantification of local phase gradient with Fourier-domain analyses

A numerical procedure, consisting of the following steps, is developed to find the correspondence between local phase gradients (e.g., within each of the small image-domain regions containing 7×7 voxels) and Fourier-domain energy peaks. First, the input complex-valued image (of matrix size: $N_x \times N_y$) is multiplied with a 7×7 mask (or different sizes: see the Methods and Discussion sections), to zero fill the signals outside the chosen small image-domain region. Second, the Fourier-domain signals corresponding to the masked image are produced with 2D Fourier transform. Third, the Fourier-domain echo energy peak is identified, and its coordinates (in terms of deviation from center of the Fourier-domain matrix along x and y directions) are recorded. Fourth, the above-mentioned steps are repeated with different masks (of the same size but different locations, i.e., a sliding window scheme: requiring $N_x \times N_y$ operations of Fourier transform), to quantify the Fourier-domain echo peak coordinates in every image-domain region. The output of these steps includes 1) a Fourier-domain peak displacement map (of matrix size: $N_x \times N_y$) along the x-direction, and 2) a Fourier-domain peak displacement map (of matrix size: $N_x \times N_y$) along the y-direction. This numerical procedure is conceptually similar to the previously reported k-space energy spectrum analysis (23), which identifies the k-space echo energy location through masking the k-space data instead of image-domain data.

The produced Fourier-domain echo displacement maps were then converted to phase gradient maps ($\Delta\theta_x(x, y)$ and $\Delta\theta_y(x, y)$) along the x and y directions with Equations 2 and 3, respectively.

$$\Delta\theta_x(x, y) = \Delta k_x(x, y) \times 2\pi / N_x \quad [2]$$

$$\Delta\theta_y(x, y) = \Delta k_y(x, y) \times 2\pi / N_y \quad [3]$$

The computation of phase gradient based on Fourier-domain analyses (i.e., with Equations 2 and 3) has better tolerance to noise than image-domain analyses, as suggested by Figure 1. The computed phase gradient maps, along both x- and y-directions, can then undergo 2D numerical integration to reveal wrap-free phase values in critical brain regions (e.g., near the air-tissue interfaces and with low SNR) using appropriate boundary conditions (e.g., known phase values in voxels with high SNR and away from the air-tissue interfaces). In our implementation, the MRI phase data are first processed with conventional temporal-domain or spatial-domain phase unwrapping procedures, and the successfully unwrapped phase values (e.g., in voxels with high SNR) are used as the boundary conditions for 2D integration. Our 2D numerical integration procedure is implemented based on matrix

inversion, and is independent of the integration path, as described in the <https://nankueichen.github.io/2d-numerical-integration.html> and Supporting Information S1. The complete reconstruction codes (including Fourier-domain analyses based gradient mapping; 2D numerical integration, in both Matlab and Julia) and test data sets are available in https://github.com/nankueichen/Fourier_space_phase_unwrapping.

Methods

Our developed phase gradient mapping and phase unwrapping methods were evaluated with human MRI data obtained from 30 healthy adult participants (4 participants in Study 1; 26 participants in Study 2) using a 3 Tesla MRI system (Discovery MR750 of General Electric (GE) Healthcare, Waukesha, WI, USA) equipped with an 8-channel RF coil. Human studies were conducted with the approval of the Institutional Review Board (IRB) of Duke University Medical Center.

Study 1 was designed to evaluate the developed methods in reducing phase wrap artifact in manually chosen voxels of temporally-unwrapped phase data, and Study 2 was conducted to assess the Fourier-domain analyses on calculating wrap-free phase values in automatically chosen ROIs of spatially-unwrapped phase data, as detailed below.

Study 1: Reduction of residual wrap artifact in multi-echo gradient-echo MRI phase maps obtained with temporal-domain phase unwrapping

Multi-echo gradient-echo MRI data were obtained from 4 volunteers, using a set of typical QSM scan parameters: FOV = 25.6 cm $\times$ 25.6 cm; in-plane matrix size = 256 $\times$ 256 (full-Fourier sampling); axial-plane slice thickness = 2 mm; number of slices = 74; voxel size = 1 $\times$ 1 $\times$ 2 mm³; readout bandwidth = 62.5 kHz; TEs of 8 echoes = 5.3, 10.4, 15.6, 20.8, 25.9, 31.1, 36.3, 41.4 msec; TR = 100 msec; flip angle = 20 degrees; and in-plane parallel acceleration factor = 2.

Both real- and imaginary-components of complex-valued images produced by the scanner (through GE's parallel reconstruction) were saved, and used as the input of the following data processing steps, aiming to calculate wrap-free phase map at the longest TE (41.4 msec). First, the phase values corresponding to 8 TE values were unwrapped using the temporal-domain unwrapping procedure described by Reber et al. (11). It should be noted that, with the chosen TE increment (5.168 msec), only the voxels with B₀ frequency offset < 96.7 Hz could be successfully unwrapped along the temporal-domain (11). Second, voxels with unsuccessfully unwrapped values were identified visually (based on the noticeable phase discontinuities between nearest neighboring voxels near the air-tissue interfaces), and manually outlined and saved (with the ROI editor tool provided by the mrview program of mrtrix). Third, the phase gradients along both x- and y-directions were calculated using the Fourier-domain processing procedures described in the Theory section. Based on the simulation and experimental data (Supporting Information S2: summarized in the next paragraph), 7 $\times$ 7 mask was used in the computation of phase gradients. Fourth, wrap-free phase values of the voxels within critical brain regions (identified in the second step) were calculated from the phase gradient maps (from the third step) and the boundary conditions

(i.e., the temporally unwrapped phase values immediately outside the critical brain region), using our implemented path-independent 2D numerical integration program.

Supporting Information S2 illustrates that 1) the performance of the proposed phase gradient computation depends on mask size, spatial frequency of phase gradient variations, extension of image-domain areas with homogeneous phase gradients, phase values, and SNR; and 2) a mask size 7×7 is a reasonable choice for Fourier-domain analyses for MRI data with a wide range of SNR. Specifically, Supporting Information Figure S1 shows that the residual error is more pronounced in areas with fast-switching phase values of a mathematical phantom. Supporting Information Figure S2 quantitatively compares the residual error in Fourier-domain-computed phase gradient maps corresponding to 5 different mask sizes and 6 levels of mathematically added noise. Simulation results shown in Supporting Information Figure S2 suggest that a mask size 7×7 is a reasonable choice for Fourier-domain analyses for data with a wide range of SNR, although a smaller mask size would be needed to more accurately calculate fast-changing phase gradients (in data with high SNR). Supporting Information Figure S3 measures residual error in Fourier-domain-computed phase gradient maps corresponding to different mask sizes and different levels of SNR in human brain MRI data, confirming that the optimal mask size depends on the SNR of input images (e.g., a 7×7 filtering mask for images with SNR=2). Supporting Information Figure S4 compares phase gradient maps obtained from data with different SNR levels, illustrating that our implementation, with a 7×7 filtering mask, has a good tolerance to noise.

Study 2: Reduction of residual wrap artifact in MRI phase maps obtained with spatial-domain phase unwrapping

T2*-weighted MRI data were obtained from 26 volunteers using a two-shot EPI pulse sequence with the following parameters: FOV = 23 cm $\times$ 23 cm; in-plane matrix size = 128 $\times$ 128 (full-Fourier sampling); TR = 5 sec; TE = 26 msec; sagittal-plane slice thickness = 1.8 mm; voxel size = 1.8 $\times$ 1.8 $\times$ 1.8 mm³; readout bandwidth = 125 kHz; and the number of repetitions = 50. The k-space data were saved, and processed with the in-house program to remove Nyquist artifact (24). The TSENSE (25) reconstruction scheme was then used to calculate coil sensitivity profiles directly from EPI data, and produce 100 sets of complex-valued parallel images from the acquired data (2 shots $\times$ 50 repetitions = 100). These 100 complex-valued images were averaged to produce a single data set with high SNR (>70 for the magnitude signals in occipital-lobe ROIs), which was used as the input of our subsequent analyses, as described below.

First, since the input data had high SNR, the phase values could be successfully unwrapped with the PRELUDE procedure (3D mode; without additionally specifying parameters: see https://fsl.fmrib.ox.ac.uk/fsl/fslwiki/FUGUE/Guide#PRELUDE_.28phase_unwrapping.29) (13). PRELUDE-unwrapped phase maps of high-SNR were used as the ground truth.

Second, the ground truth magnitude images and the PRELUDE-unwrapped phase maps from all of the 26 subjects were normalized to the Montreal Neurological Institute (MNI) space with FSL and averaged to produce a mean phase map, from which the critical frontal brain region affected by pronounced field inhomogeneities was identified (with

$|\Delta B_0| > 100\text{Hz}$ at 3 Tesla: with Supporting Information Figure S5 showing the critical frontal brain regions affected by pronounced field inhomogeneities, as detailed in Supporting Information S3). The critical frontal brain ROI with pronounced susceptibility field gradients was projected back to individual subject's space, automatically identifying critical brain regions where the conventional image-space phase unwrapping procedures (e.g., PRELUDE) might not work robustly in data sets with low SNR.

Third, Gaussian noise was then mathematically added to real- and imaginary components of complex-valued data to produce low-SNR data for each of the subjects. The noise level was chosen in such a way that the $\text{SNR} \approx 2$ for the magnitude signals in the occipital-lobe ROIs. The PRELUDE program (3D mode; without specifying parameters) was then used to unwrap the low-SNR phase images. For noisy MRI data, the PRELUDE method could perform reliably in most brain regions but might fail to produce accurate phase values in brain regions affected by pronounced susceptibility field gradients (e.g., within frontal brain regions shown in Supporting Information Figure S5).

Fourth, phase gradient maps along the left-right (x) and anterior-posterior (y) directions were calculated from the complex-valued MRI with mathematically added noise, using the Fourier domain procedures described in the Theory section. The Fourier-domain analyses provided phase gradient maps were compared with the phase gradient maps calculated from PRELUDE-unwrapped phase images (of low SNR) from the third step, to identify voxels whose phase values were not successfully unwrapped by PRELUDE (with difference between two phase gradient maps greater than π). Afterward, wrap-free phase values in the identified voxels were calculated using our implemented 2D path-independent numerical integration, with Fourier-analyses provided phase gradient maps and appropriate boundary conditions (i.e., nearby voxels whose phase values were successfully unwrapped by PRELUDE) as the input.

Fifth, root-mean-square errors of the unwrapped phase values (with low SNR), obtained with PRELUDE and the developed Fourier-domain procedures, were calculated from all the voxels within critical frontal brain ROI (shown in Supporting Information Figure S5) using high-SNR PRELUDE-unwrapped phase images as the ground truth.

In order to further evaluate the Fourier-domain phase unwrapping method on complex-valued images that were not reconstructed with parallel MRI procedure (as described above), one of the k-space data sets acquired in Study 2 went through another (non-parallel MRI) reconstruction pipeline, comprising 1) Nyquist artifact correction, 2) 2D Fourier transform of full k-space data of both EPI segments, 3) removal of RF channel-specific background phase using information obtained from coil-sensitivity calibration scan, 4) averaging the complex-valued data across 8 RF receiver channels and 50 repetitions, and 5) phase unwrapping with the reported method. The unwrapped phase maps obtained with parallel and non-parallel MRI reconstruction pipelines were then compared.

Results

Using the developed Fourier-domain analyses (i.e., Equations 1 to 3), phase gradient maps could be computed from complex-valued MRI data obtained with different pulse sequences from 30 participants. For example, Figures 2a and 2b show a set of magnitude and phase images of T2*-weighted MRI, respectively, and the corresponding Fourier-domain data are shown in Figure 2c (showing magnitude only). With the procedure described in the Theory section (i.e., taking 2D Fourier transform of masked image-domain signals), the Fourier-domain data corresponding to the yellow, orange and white masks (see Figure 2a) were computed, as shown in Figures 2d, 2e, and 2f, respectively, from which the Fourier-domain peak coordinates could be measured. After measuring the Fourier-domain peak coordinates corresponding to all possible mask locations (i.e., a sliding window scheme), the Fourier-domain peak displacement maps along x- and y-directions were generated, as shown in Figures 2g and 2h, respectively, which could be immediately converted to phase gradient maps with Equations 2 and 3. As described in the Theory section, wrap-free phase values in a chosen ROI (e.g., outlined by red line in Figure 2i) could be computed from phase gradient maps with 2D numerical integration.

Our Julia and Matlab implementations of phase gradient mapping took about 0.58 sec and 2.19 sec, respectively, for processing an image with 78×128 matrix size in a Linux desktop equipped with 8-core CPU (Intel i7-7700 @ 3.6GHz) and 16GB memory. Our implemented 2D numerical integration for computing wrap-free phase values in a 2D critical brain ROI (e.g., comprising 530 pixels) took about 1.28 sec and 0.18 sec with Julia and Matlab, respectively (https://github.com/nankueichen/Fourier_space_phase_unwrapping).

Study 1: Reduction of residual wrap artifact in multi-echo gradient-echo MRI phase maps obtained with temporal-domain phase unwrapping

Figures 3a to 3d show T2*-weighted MRI data with TE = 41.4 msec, obtained from 4 healthy volunteers using a typical QSM protocol (multi-echo gradient-echo MRI with 8 TE values ranging from 5.3 to 41.4 msec). Columns 1 to 3 show the magnitude images, phase maps obtained with temporal-domain phase unwrapping, and phase maps further processed with the developed Fourier-domain data analysis procedure, respectively. It can be seen that, because of the pronounced local susceptibility field gradients in critical brain region (within red contours), the magnitude signals have low SNR (column 1) and the phase values cannot be successfully unwrapped (column 2) with temporal-domain unwrapping procedures.

Phase gradients along both x- and y-directions were computed with Fourier-domain analyses, and then used to produce wrap-free phase values within critical brain regions using 2D numerical integration with phase values outside the red contours (see Figure 3) as the boundary conditions. It can be seen that the phase discontinuities (artifact) column 2 of Figure 3 are successfully removed in phase maps produced with our developed procedures (column 3 of Figure 3).

Study 2: Reduction of residual wrap artifact in MRI phase maps obtained with spatial-domain phase unwrapping

Table 1 lists the root-mean-square errors of phase values, calculated from all of the voxels within the critical frontal brain ROI (shown in Supporting Information Figure S5), in data obtained with the PRELUDE procedure and Fourier-domain based analyses. Table 1 is sorted according to the reduction of root-mean-square errors achieved with the Fourier-domain analyses. The mean value of the error reduction is 0.425 (radian), and the standard deviation is 0.390. It can be seen that the residual phase wrap artifact is reduced by the developed Fourier-domain analysis procedures in 25 of 26 data sets (except for data from subject ID 12, in which high spatial-frequency components of phase gradients might need to be measured with a smaller mask size).

Figures 4a and 4b show the magnitude and phase images (with mathematically added noise), respectively, from 4 of the 26 subjects (including subject ID 12: the only data set that the Fourier-domain analyses slightly increase the error). Figure 4c shows the PRELUDE-unwrapped phase maps, and Figure 4d shows phase images obtained with the Fourier-domain analyses. It can be seen that, where significant phase wraps exist (e.g., indicated by arrows in ID 15, ID 6, ID 2), the wrap artifact could be reduced by our developed Fourier-domain unwrapping procedures. Figure 4e shows the ground truth phase images (without mathematically added noise).

In addition, we found that the Fourier-domain analysis of complex-value images obtained with parallel and non-parallel MRI reconstruction pipelines produced very similar results (data not shown).

Discussion

Here we demonstrate that the developed Fourier-domain analyses can be used to reduce residual artifact in phase maps produced by the conventional temporal-domain and spatial-domain phase unwrapping procedures. The capability of robustly unwrapping MR phase images should prove highly valuable for many applications, including B_0 field mapping, QSM, and temperature mapping among others. The developed phase unwrapping procedures can also be applied to non-MRI data, such as optical imaging data.

In our current implementation, the Fourier-domain analysis based phase gradient maps (e.g., Figures 2g and 2h) are used to calculate phase values only within a small area (instead of the full FOV) using a path-independent 2D numerical integration, with the boundary condition provided by the conventional phase unwrapping procedures (e.g., PRELUDE). Therefore, this implementation relies on an initial successful phase unwrapping in a large brain region, before the Fourier-domain analyses provided information could be used.

In our studies the 2D phase gradient maps were computed from each of the slices, and then used to produce phase values in critical brain ROIs using 2D numerical integration on a slice-by-slice basis. Our 2D procedures could be extended to 3D processing. First, it is straightforward to implement a 3D Fourier-space analysis based 3D phase gradient mapping, which is similar to our implemented 2D procedure, through multiplying 3D Fourier-space

data with cuboidal masks. Second, our 2D numerical integration procedure could also be extended to 3D.

The developed Fourier-domain phase unwrapping method has been evaluated in parallel multi-echo gradient-echo MRI in Study 1 and T2*-weighted EPI data Study 2, in which complex-valued images were produced with parallel MRI reconstruction. To evaluate the Fourier-domain phase unwrapping method with complex-valued images that were not reconstructed with parallel MRI procedure, the k-space acquired in Study 2 went through another (non-parallel MRI) reconstruction pipeline, comprising 1) Nyquist artifact correction, 2) 2D Fourier transform of full k-space data of both EPI segments, 3) removal of RF channel-specific background phase using information obtained from coil-sensitivity calibration scan, 4) averaging the complex-valued data across 8 RF receiver channels and 50 repetitions. We found that the Fourier-domain analysis of the reconstructed complex-value images (without undergoing parallel MRI reconstruction) could produce phase gradient maps that were very similar to the ones obtained with the TSENSE produced complex-valued data shown in the Results section.

An important component of our reported phase unwrapping method is the identification of voxels whose phase values cannot be successfully unwrapped with conventional procedures. As shown in our studies, these voxels could be identified in multiple ways. In our Study 1, voxels with unsuccessfully unwrapped values were identified visually and then manually outlined. However, this procedure is labor intensive and may not be practical for processing a large number of data sets. In our Study 2, critical frontal brain ROI with pronounced susceptibility field gradients was projected from MNI space back to individual subject's space for 26 healthy adults, automating the phase unwrapping procedure without manual input. However, image normalization (between MNI and subject's native space) may not always be robust, especially for patient MRI data. A robust and fully automated procedure for identifying unsuccessfully unwrapped voxels in clinical MRI data remains to be developed.

In summary, in this paper we report a procedure that can robustly unwrap MR phase images, even in critical brain regions with low SNR and affected by pronounced susceptibility field gradients. The developed method should prove valuable for MR phase mapping, and can also be applied to other types of imaging data (e.g., optical imaging data).

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1:

(a) shows image-domain phase signals in the presence of $-8 \times 2\pi$ radians/FOV phase gradients. (b) shows image-domain phase signals in the presence of $22 \times 2\pi$ radians/FOV phase gradients. (c) The Fourier-domain data corresponding to (a) and (b) are shown with blue line (peaked at $k_x = 8$) and black line (peaked at $k_x = -22$), respectively. (d) shows phase signals of (a) with mathematically added noise (to both real- and imaginary-components of complex-valued data). (e) shows phase signals of (b) with mathematically added noise. (f) The energy peaks in Fourier-domain data corresponding to (d) and (e) are identifiable even in the presence of mathematically added noise.

Figure 2:

(a) and (b) are the input magnitude and phase images, respectively. (c) shows the Fourier-domain signals for all the voxels in (a). (d), (e), and (f) show the Fourier-domain signals corresponding to only voxels in the yellow, orange and white masks (displayed in (a)), respectively. (g) and (h) are the phase gradient maps along the x and y directions, respectively, obtained with Fourier-domain analyses. (i) shows the unwrapped phase map produced with our implemented 2D numerical integration.

Figure 3:

Rows (a) to (d) show data obtained from 4 healthy volunteers. Column 1 shows the T2*-weighted gradient-echo magnitude image. Column 2 shows the temporally-unwrapped phase map. Column 3 shows the phase map obtained with the developed Fourier-domain processing procedures.

Figure 4:

(a) and (b) are the magnitude and phase images (with mathematically added noise), respectively, from 4 of the 26 subjects. (c) shows the spatially-unwrapped phase maps. (d) are phase images obtained with the developed Fourier-domain processing method. (e) are the ground truth phase images.

Table 1:

Root-mean-square errors (RMSE) in phase maps calculated with the PRELUDE (3D mode; no additionally specified parameters) and our Fourier-domain analyses from MRI data obtained from 26 healthy adult volunteers. The table is sorted based on the reduction in RMSE.

Subject ID	RMSE: prelude	RMSE: integrated	Reduction in RMSE	Subject ID	RMSE: prelude	RMSE: integrated	Reduction in RMSE
15 *	2.93	1.62	1.31	1	1.44	1.16	0.27
24	8.84	7.69	1.15	13	2.58	2.31	0.27
25	3.30	2.27	1.03	16	0.92	0.69	0.23
14	2.28	1.31	0.97	2 *	1.76	1.56	0.20
21	2.80	1.93	0.88	22	1.36	1.18	0.18
7	3.21	2.44	0.78	10	1.35	1.24	0.10
20	4.79	4.02	0.77	26	2.43	2.33	0.10
5	2.98	2.40	0.58	9	2.21	2.12	0.10
4	4.08	3.57	0.51	23	1.33	1.24	0.08
6 *	2.74	2.31	0.43	17	2.91	2.86	0.05
18	4.73	4.30	0.43	11	0.89	0.85	0.05
3	1.52	1.19	0.33	8	0.89	0.85	0.04
19	1.57	1.26	0.31	12 *	1.66	1.76	-0.10 [§]

* Their phase images are shown in Figure 4.

[§] The phase errors increase after processing data with the Fourier-domain analyses.