Magnetic Resonance Imaging 20 (2002) 455–462

Motion artifact reduction technique for dual-contrast FSE imaging

Eugene G. Kholmovski[a,*], Alexei A. Samsonov[a], Dennis L. Parker[a,b]

[a]Department of Physics, University of Utah, Salt Lake City, Utah, 84112, USA
[b]Department of Radiology, University of Utah, Salt Lake City, Utah, 84132, USA

Received 18 May 2002; accepted 17 June 2002

Abstract

There is considerable similarity between proton density-weighted (PDw) and T2-weighted (T2w) images acquired by dual-contrast fast spin-echo (FSE) sequences. The similarity manifests itself in image space as consistency between the phases of PDw and T2w images and in k-space as correspondence between PDw and T2w k-space data. A method for motion artifact reduction for dual-contrast FSE imaging has been developed. The method uses projection onto convex sets (POCS) formalism and is based on image space phase consistency and the k-space similarity between PDw and T2w images. When coupled with a modified dual-contrast FSE phase encoding scheme the method can yield considerable artifact reduction, as long as less than half of the acquired data is corrupted by motion. The feasibility and efficiency of the developed method were demonstrated using phantom and human MRI data. © 2002 Elsevier Science Inc. All rights reserved.

Keywords: Magnetic resonance imaging; Dual-contrast fast spin-echo; Motion artifact; POCS; Phase encoding scheme

1. Introduction

Proton density-weighted (PDw) and T2-weighted (T2w) images contain complementary information that may significantly improve the diagnostic value of MRI scans. Dual-contrast fast spin-echo (FSE) sequences, based on the rapid acquisition with relaxation enhancement (RARE) technique first introduced by Henning et al. [1], are widely used to simultaneously acquire high-resolution PDw and T2w images [2–4]. Images acquired using dual-contrast FSE sequences do not require post-acquisition spatial coregistration, which greatly simplifies the application of multiple image analysis techniques such as multivariate segmentation.

Typical acquisition times for dual-contrast FSE scans can exceed several minutes thereby making such scans vulnerable to image quality degradation caused by accidental patient motion. The appearance of artifact in reconstructed images strongly depends on the k-space trajectory used for data acquisition. In spin-warp (Fourier) FSE imaging, motion predominantly leads to blurring along the motion direction and ghosting along the phase encoding directions [5]. Any temporary deviation of the imaged object from the baseline object position during FSE imaging may result in serious degradation of the reconstructed images. This strong sensitivity to motion is primarily caused by the specific nature of the FSE phase encoding scheme, where each echo train acquires low frequency components of k-space data. As a rule, inconsistency in the central (low frequency) part of the MR data causes severe image degradation. Thus, motion-caused inconsistency between the data acquired by a small number of echo trains and all other k-space data may negatively impact the quality of the reconstructed image. In this paper k-space data that are inconsistent with the other k-space data because of motion will be referred to as motion corrupted data.

In typical dual-contrast FSE imaging the first half of the echo train is used to acquire PDw k-space data and the second half is used to acquire T2w data [6,7]. The effective echo time (TE) for each image is determined by the echo time of the echoes, which are used to acquire the central part of the corresponding k-space. The standard phase encoding scheme for dual-contrast FSE sequences is implemented in such a way that the same phase encoding views are acquired for PDw and T2w images during a single echo train. The typical time to acquire an echo train, less than two hundred milliseconds, is significantly smaller than the characteristic time of non-cardiac patient motions. Thus, all lines of k-space data acquired by a single echo train may be consid-

* Corresponding author. Tel.: +1-801-581-8247; fax: +1-801-585-3592.

E-mail address: ekhoumov@doug.med.utah.edu (E.G. Kholmovski).

PII: S0730-725X(02)00526-X

ered to be corrupted if the imaged object deviates from the baseline position during the given echo train acquisition. The relative artifact appearance in dual-contrast FSE acquired PDw and T2w images strongly depends on the phase encoding scheme used. In the case of dual-contrast FSE imaging with the standard phase encoding scheme, the same lines of PDw and T2w k-space data will be corrupted by motion, which will result in a similar artifact appearance in the final PDw and T2w images. However, if the phase encoding scheme is modified such that the same phase encoded views of PDw and T2w data are acquired by different echo trains, then different sets of k-space lines of PDw and T2w data will be corrupted by motion. This will result in notably different artifact appearances in the PDw and T2w images.

In the absence of motion there is a significant similarity (redundancy) between PDw and T2w images that manifests itself both in image space and k-space. The goal of this work is to develop a technique that uses the inherent redundancy of dual-contrast FSE acquired images to reduce image quality degradation caused by motion.

2. Materials and methods

2.1. Phase consistency and k-space similarity

The PDw and T2w images of a stationary object, acquired by the dual-contrast FSE sequence, may be described by the following equations:

$$I_{PD}(\boldsymbol{r}) = I(\boldsymbol{r})\Phi_{PD}(\boldsymbol{r}) + N_{PD}(\boldsymbol{r}) = J_{PD}(\boldsymbol{r}) + N_{PD}(\boldsymbol{r}), \tag{1}$$

$$I_{T2}(\boldsymbol{r}) = I(\boldsymbol{r})W(\boldsymbol{r})\Phi_{T2}(\boldsymbol{r}) + N_{T2}(\boldsymbol{r}) \tag{2}$$

where $I(\boldsymbol{r})$ is the magnitude of the noise-free PDw image, $J_{PD}(\boldsymbol{r})$; $\Phi_{PD}(\boldsymbol{r})$ and $\Phi_{T2}(\boldsymbol{r})$ are the phase variation factors in noise-free PDw and T2w images; $W(\boldsymbol{r})$ is the T2 weighting factor (real positive definite function); and $N_{PD}(\boldsymbol{r})$ and $N_{T2}(\boldsymbol{r})$ are complex noise contributions in the PDw and T2w images. The real and imaginary components of $N_{PD}(\boldsymbol{r})$ and $N_{T2}(\boldsymbol{r})$ may be characterized by zero mean Gaussian probability density functions with the same standard deviation and can be considered to be mutually uncorrelated.

Equations (1) and (2) show that the difference between noise-free PDw and T2w images of a stationary object may be characterized by T2 weighting and phase variation factors. These factors have specific features which may be used to reduce image degradation by motion artifacts.

The spatial phase variation factors $\Phi_{PD}(\boldsymbol{r})$ and $\Phi_{T2}(\boldsymbol{r})$ are defined by $\Phi_{PD}(\boldsymbol{r}) = \exp\,(i\phi_{PD}(\boldsymbol{r}))$ and $\Phi_{T2}(\boldsymbol{r}) = \exp\,(i\phi_{T2}(\boldsymbol{r}))$, where $\phi_{PD}(\boldsymbol{r})$ and $\phi_{T2}(\boldsymbol{r})$ are the spatially dependent phases of the noise-free PDw and T2w images. One of the main requirements for obtaining high-quality MR images using an FSE sequence is the stability of the spatial phase variations during the data acquisition. The image phase distribution (neglecting phase encoding) must be the same for all echo trains and all echoes that compose each echo train. When this condition is not satisfied (e.g. object motion occurs during the scan) the acquired images have artifacts. In this work, we assume that the FSE data acquisition is designed to meet the requirement of phase variation stability during imaging of a static object. Therefore, the phase variations in the PDw and T2w images of a stationary object acquired using the dual-contrast FSE sequence should be the same: $\Phi(\boldsymbol{r}) = \Phi_{PD}(\boldsymbol{r}) = \Phi_{T2}(\boldsymbol{r})$. This observation will be referred to as phase consistency in this paper. Additionally, in the case of FSE imaging of a stationery object, the image phase may be described by a slowly varying function.

The relationship between the k-space of the noise-free PD-weighted image and the k-space of the T2-weighted image of a stationary object is given by:

$$I_{T2}(\boldsymbol{k}) = J_{PD}(\boldsymbol{k}) \otimes W(\boldsymbol{k}) + N_{T2}(\boldsymbol{k}). \tag{3}$$

In the case of dual-contrast FSE imaging, a typical $W(\boldsymbol{k})$ consists of a sharp peak in the center of k-space and low values in other areas. The value of $W(\boldsymbol{k})$ in the center of k-space is real and positive and significantly larger than the amplitude of neighboring points. Such properties of $W(\boldsymbol{k})$ lead to considerable similarity between k-space data of noise-free PD- and T2-weighted images. For real MR data the similarity continues to exist in the central part of k-space but is seriously corrupted by noise contamination in the high frequency part of k-space.

2.2. Motion artifact reduction technique

A motion artifact reduction (MAR) method for dual-contrast FSE imaging has been developed. The main assumption of the proposed method is that patient motion occurs only during a fraction of the scan time, and therefore, only a portion of the acquired MR data is corrupted. It is assumed that motion consists of a temporal deviation of the imaged object from the baseline object position. When these assumptions are valid, the acquired k-space data may be divided into two groups: Motion corrupted data and valid data. The above described similarity of PDw and T2w images and k-spaces is violated for the first group but continues to be valid for the second group and may be used to improve the quality of motion corrupted images.

The MAR method, which is based on k-space similarity and phase consistency between PDw and T2w images, consists of two main steps: 1. Identification of k-space lines corrupted by motion; and 2. Recovery of corrupted k-space lines and image reconstruction.

2.2.1. Identification of k-space lines corrupted by motion

In the FSE acquired images, artifact caused by accidental patient motion manifests itself not only as violations of the image structure (consistency between the object and image of the object), but also as the existence of correlated features in the image background [5]. This type of artifact manifestation is common for sequences with periodic structure in the phase encoding arrangement, such as occurs in FSE sequences, where the echo trains are obtained at specific relative positions in k-space [8–10]. The lines of k-space corrupted by motion may be identified based on the correlated features of image background caused by the motion. To achieve this, an image support mask is identified and the Fourier transform of the image is calculated after the object has been masked out. The corrupted lines of k-space appear as peaks when the energy spectrum is integrated over kx.

The motion identification based on correlated features of the image background requires that motion caused violations of k-space data consistency have higher energy than k-space noise contributions. This condition is usually met for FSE imaging, where motion results in errors for each line of k-space acquired during a single echo train and each echo train has a few echoes that acquire low frequency (high energy) components of k-space data. We have found that better motion identification can be made for echoes acquired near the center of k-space, and if a motion error is detected for one echo in the echo train the entire echo train can be considered to be motion corrupted. In the case of dual-contrast FSE imaging the identification is further improved by the existence of two data sets with known relative arrangements of PDw and T2w phase encoding views acquired using the same echo train.

2.2.2. Recovery of corrupted k-space lines and image reconstruction

To recover corrupted k-space lines and reconstruct a corrected image, a priori information about the image should be used as reconstruction constraints. From the first step of the MAR method the spatial support constraint is already known. Another constraint that can be used to recover corrupted k-space lines is that the phase of an FSE acquired image may be described by a smoothly varying function in cases of static object imaging. This smoothly varying phase property has been used to reconstruct images from partial k-space data by methods such as those described by Margosian et al. [11] and Cuppen et al. [12], homodyne detection [13], or projection onto convex sets (POCS) [14–16]. All of these methods use a phase estimate calculated from the central (low-frequency) part of the acquired data to correct for the missing high-frequency k-space data. The problem of reconstructing motion corrupted data may be converted into the same form as the partial data reconstruction problem with the only difference being that the data corrupted by motion are actually acquired, but should then be discarded and replaced by the data recovered during the reconstruction. In the reconstruction of motion corrupted k-space data, a phase estimate cannot be found using the method described above because both low and high frequency data are corrupted by motion. Therefore, a preliminary step of the reconstruction scheme should be implemented to partially recover the corrupted k-space lines at the central part of k-space. This partly recovered k-space data may then be used to determine an image phase estimate that can be used as a reconstruction constraint to recover all motion corrupted k-space lines and thereby reconstruct a corrected image.

In the case of dual-contrast FSE imaging there are two sets of k-space data. Since PDw and T2w images of a static object acquired by dual-contrast FSE sequences have the same phase, the phase of the PDw image may be used as a reconstruction constraint to recover the T2w image, and vice versa, as was shown for the case of a partial data acquisition [17]. However, if motion occurs, the PDw and the T2w images will both have artifacts. For the standard phase encoding scheme, the artifact appearance (spatial distribution) in PDw and T2w images is similar because the same lines of PDw and T2w k-space data are corrupted. To use the similarity between PDw and T2w k-space data for phase constraint identification, the phase encoding scheme of dual-contrast FSE sequences should be changed such that different phase encoding views are acquired for the PDw and the T2w images during a single echo train. A modified phase encoding scheme that allows selection of a time separation interval between acquisition of the same phase encoding views of PDw and T2w k-space data has been developed. If this modified phase encoding scheme is used to acquire MR data, motion will corrupt different sets of k-space lines, allowing the use of k-space similarity between PDw and T2w data to find the phase constraint.

The following method for image reconstruction from motion corrupted MR data in dual-contrast FSE imaging has been developed. The method uses the POCS formalism and the similarity between PDw and T2w images. A detailed description of the reconstruction algorithm for the PDw image is given below. An analogous procedure should be implemented to reconstruct the corresponding T2w image with reduced motion artifact manifestation.

Let $\mathcal{K}$ be a set of k-space positions that correspond to the sampling on the Cartesian grid that gives the desired field of view (FOV) and resolution of PDw and T2w images. $K^{PD}(\boldsymbol{k})$ and $K^{T2}(\boldsymbol{k})$ are the acquired PDw and T2w k-space data. If motion occurs during data acquisition the data should be divided in two groups: Motion corrupted and valid data. $\mathcal{K}^{PD}$ and $\mathcal{K}^{T2}$ are subsets of $\mathcal{K}$ that correspond to k-space positions of motion corrupted PDw and T2w data. The first step of the MAR technique is used to identify $\mathcal{K}^{PD}$ and $\mathcal{K}^{T2}$. Denoting Ω_1 to be the convex set of images whose phase is the same as the phase estimate ϕ; Ω_2 to be the convex set of images for which space support is the same as the reconstructed image support S; and Ω_3 to be the convex set of images whose k-space data agree with the valid PDw k-space data ($K^{PD}(\boldsymbol{k})$ with $\boldsymbol{k} \notin \mathcal{K}^{PD}$), we define the following projection operators onto Ω_1, Ω_2, and Ω_3:

$$P_1\text{: } P_1(\phi)I_n = |I_n| \cos(\phi_n - \phi)e^{i\phi} = (I_n + I_n^* \cdot e^{2i\phi})/2, \tag{4}$$

$$P_2\text{: } P_2(S)I_n = \begin{cases} I_n(\boldsymbol{r}), & \boldsymbol{r} \in S \\ 0, & \text{otherwise}, \end{cases} \tag{5}$$

$$P_3\text{: } P_3 I_n = \mathcal{F}^{-1}\{\mathcal{R} \cdot \mathcal{F}\{I_n\}\}, \tag{6}$$

where $I_n = I_n^{PD}(\boldsymbol{r})$ is the PDw image after the n-th iteration, ϕ_n is the phase of I_n, I_n^* denotes the complex conjugate of I_n, $\mathcal{F}$ and $\mathcal{F}^{-1}$ are the forward and inverse Fourier transforms, and $\mathcal{R}$ is the data replacement operator defined as:

$$\mathcal{R} \cdot I_n(\boldsymbol{k}) = \begin{cases} K^{PD}(\boldsymbol{k}), & \boldsymbol{k} \notin \mathcal{K}^{PD} \\ I_n(\boldsymbol{k}), & \boldsymbol{k} \in \mathcal{K}^{PD} \end{cases} \text{ with } \boldsymbol{k} \in \mathcal{K}. \tag{7}$$

The developed image reconstruction algorithm consists of the following steps:

1. Calculate the phase estimate, $\phi(\boldsymbol{r})$, to be used as a phase constraint.
   To achieve this the corrupted k-space lines of one echo time data set should be replaced by the corresponding k-space lines of the other echo time data set weighted by a correction factor which compensates for the difference between amplitudes of the corresponding data sets. If the same k-space line is motion corrupted in both data sets then the weighted averaged value of the line should be used. In the case of PDw image reconstruction, the combined k-space data is defined by

$$I_o(\boldsymbol{k}) = \begin{cases} K^{PD}(\boldsymbol{k}), & \boldsymbol{k} \notin \mathcal{K}^{PD} \quad \text{and} \quad \boldsymbol{k} \in \mathcal{K} \\ \gamma K^{T2}(\boldsymbol{k}), & \boldsymbol{k} \in \mathcal{K}^{PD} \quad \text{and} \quad \boldsymbol{k} \notin \mathcal{K}^{T2}, \\ K^{A}(\boldsymbol{k}), & \boldsymbol{k} \in \mathcal{K}^{PD} \quad \text{and} \quad \boldsymbol{k} \in \mathcal{K}^{T2} \end{cases} \tag{8}$$

where $K^A(\boldsymbol{k}) = (K^{PD}(\boldsymbol{k}) + \gamma K^{T2}(\boldsymbol{k}))/2$.
The correction factor, γ, may be calculated as a ratio between the magnitudes of the central values of the PDw and T2w image k-space data. Typically, images based on the combined data have a mixed (PDw and T2w) contrast and manifest residual artifacts caused by remnant violations of k-space data consistency. However, these artifacts usually are less visible than the artifacts caused by motion because the violation of k-space data consistency in the combined data is smaller than the corresponding error in the original motion corrupted data. The residual error is mainly concentrated in the high frequency part of k-space. Thus, the phase estimate ϕ may be easily calculated from the central (low-frequency) part of the combined data $I_o(\boldsymbol{k})$. The resulting phase should be close to the phase of the noise-free PDw image because of the phase consistency between the PDw and the T2w images.

2. Reconstruct the PDw image using POCS iterations.
   The resulting PDw image is sought as a point in the intersection of the convex sets Ω_1, Ω_2, and Ω_3 in Hilbert space by alternating projections onto the convex sets:

$$I_n = P_3 P_2 P_1 I_{n-1} \text{ with } I_0 = \mathcal{F}^{-1}\{I_o(\boldsymbol{k})\} \text{ and } n \geq 1. \tag{9}$$

   Typically, an image with visibly reduced motion artifact was obtained after only four to six iterations.

2.3. Imaging experiments

The proposed MAR technique was validated by studying its effectiveness on data from phantom and human subject studies. Informed consent was obtained from all subjects. All imaging studies were performed on a 1.5 Tesla GE SIGNA Lx 8.4 MRI scanner (General Electric Medical Systems, Waukeesha, WI) with NV/CVi gradients (40 mT/m amplitude, 150 mT/m/ms slew rate). The standard dual-contrast 2D FSE pulse sequence was modified to acquire different phase encoding views for PDw and T2w images during the same echo train. The time separation (TS) between the acquisition of the same phase encoding views for the PDw and T2w images may be chosen over a wide range from the time between the echo trains to a half of the slice scan time. Phantom data were acquired using a dual-contrast 2D FSE pulse sequence with the modified phase encoding scheme and with TS equal to half of the slice scan time. If motion occurs during imaging with the given TS value, then the motion will corrupt k-space lines from the opposite halves of the PDw and T2w data sets. The imaging sequence parameters were: TR = 1000 ms, effective TE = 10.5 (94.7) ms for PDw (T2w) image acquisition, TS = 128 TR, echo-train length (ETL) = 16, ±20.83 kHz receiver bandwidth (rBW), FOV = 140 × 140 mm^2, 256 × 256 in-plane acquisition matrix, and 2 mm slice thickness. Two sets of images were acquired: The first set without motion, the second set with motion. The motion was caused by a manual rotation and translation of the phantom during a fraction of the scan time.

The PDw and T2w brain images were obtained using the dual-contrast 2D FSE pulse sequence with the modified phase encoding scheme. One set of images was acquired with TS equal to half of the slice scan time and the following imaging sequence parameters: TR = 3000 ms, effective TE = 10.1 (91.1) ms for PDw (T2w) imaging, TS = 96 TR, ETL = 16, rBW = ±20.83 kHz, FOV = 240 × 180 mm^2, 256 × 192 in-plane acquisition matrix, and 2 mm slice thickness. The other images were acquired with similar imaging parameters except the TS was equal to one quarter of the slice scan time (TS = 64 TR), FOV = 220 × 220 mm^2, and a 256 × 256 in-plane acquisition matrix was used. Two sets of brain images were acquired: The first set was obtained without volunteer motion; during acquisition of the second set the volunteers were asked to temporarily change head position by in-plane or through-plane rotation.

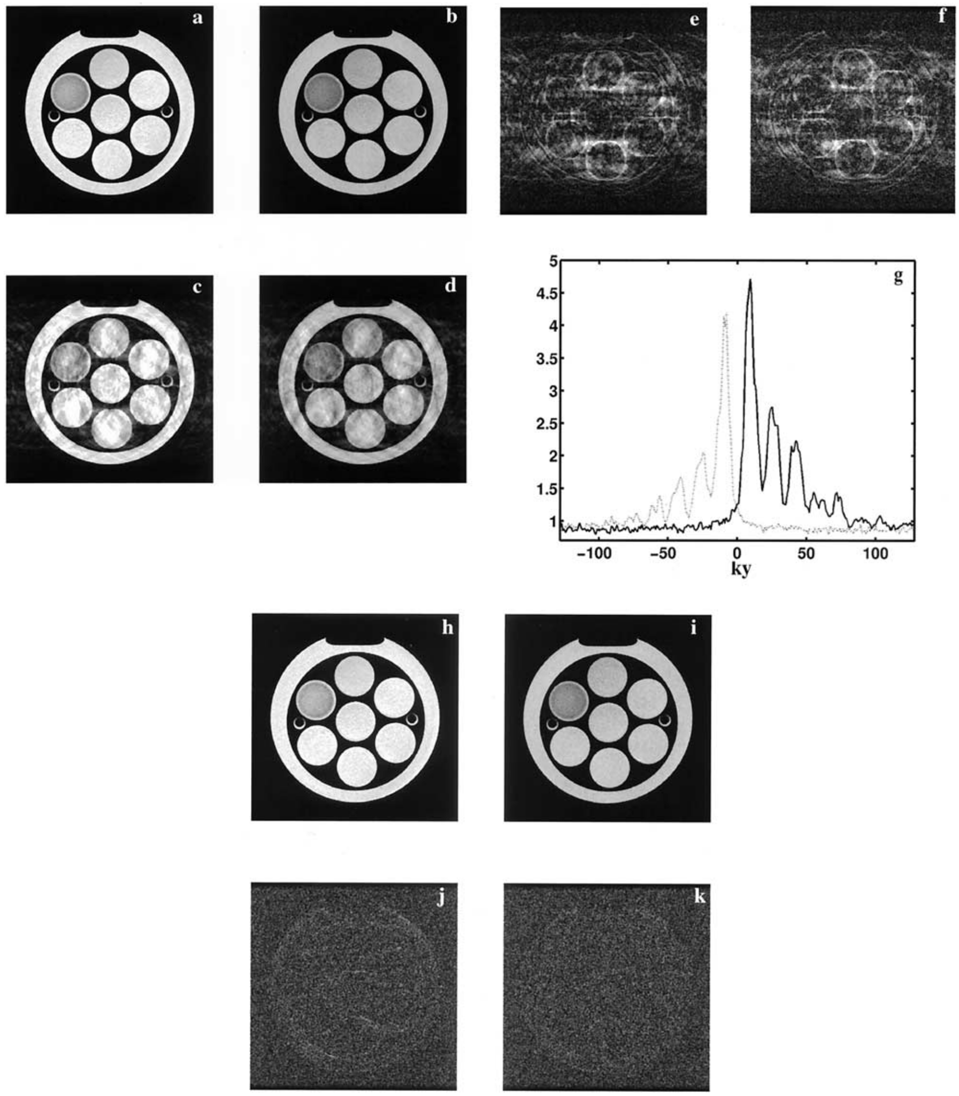

Fig. 1. PDw and T2w phantom images. (a, b) the original (artifact free) images; (c, d) the motion corrupted images; (e, f) the magnitude of the difference between the original and the motion corrupted images; (g) the plot used to identify k-space lines corrupted by motion in the PDw and T2w data sets (solid line—PDw, dotted line—T2w); (h, i) the MAR reconstructed images; (j, k) the magnitude of the difference between the original and the MAR reconstructed images.

The image support was identified manually to be everything but the external background of the image. The phase estimates were calculated using the central part (32×32 pixels) of the corresponding combined k-space data and zeroing the other k-space data. A Hamming window was used to achieve a smooth transition between the zeroed and non-zeroed data.

3. Results

Fig. 1 shows the results of the MAR technique to reduce motion artifacts in phantom images. In this study the motion was predominantly in-plane rotation. In comparison with the original (no motion) images, the motion corrupted images are seriously degraded by blurring and ghosting artifacts. The difference between the original and the motion corrupted images clearly indicates the existence of correlated features in the image background. The corrupted k-space lines of the corresponding data sets are easily detectable as periodic peaks on the plot of the integrated along read-out direction energy spectrum of the image background (Fig. 1g). The significant parts of both data sets (about 35%) can be identified as motion corrupted and should be recovered during reconstruction. The identified k-space lines are located at opposite halves of PDw and T2w k-spaces. The MAR reconstructed PDw and T2w images (Fig. 1h and 1i) have noticeably improved quality with nearly complete removal of motion artifact image degradation. The difference between the original and the MAR reconstructed images shows predominantly noise indicating that the proposed motion artifact reduction scheme almost completely recovered the motion corrupted k-space lines.

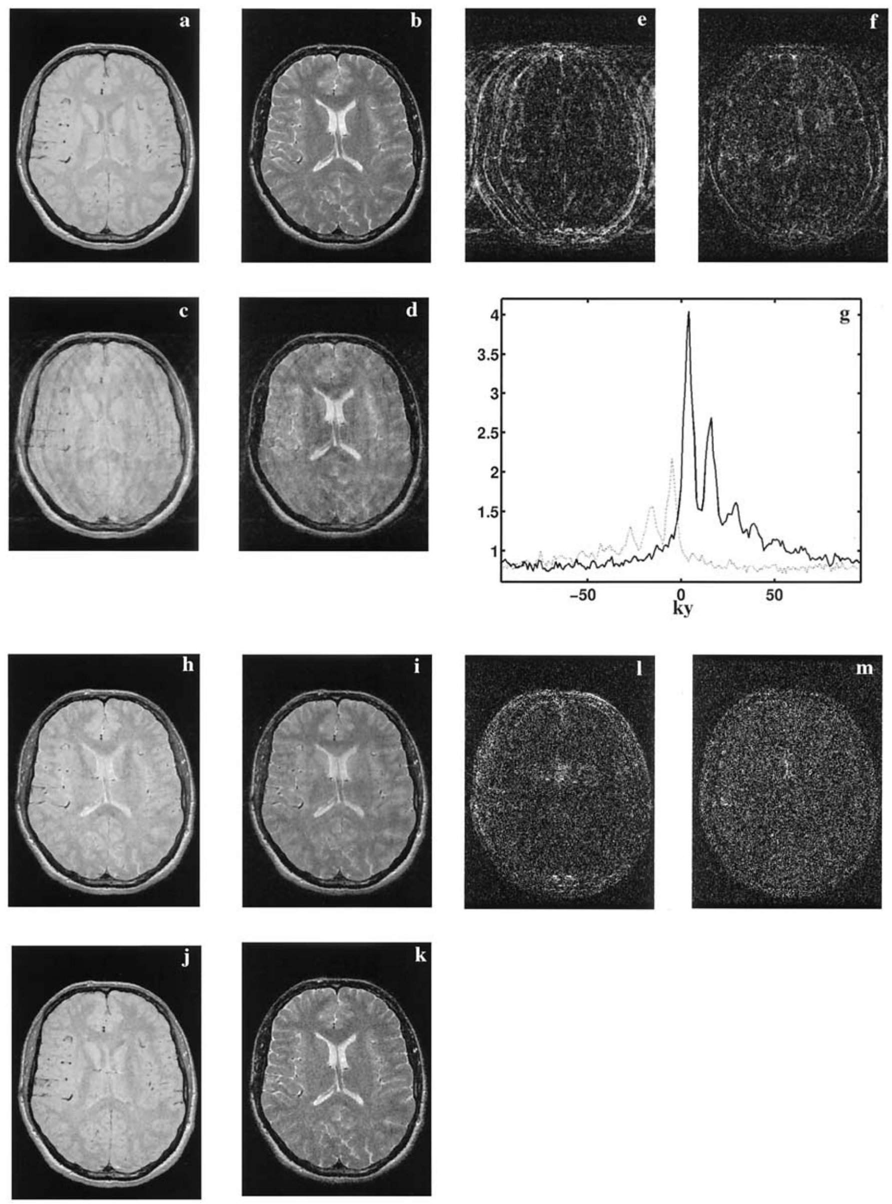

Fig. 2. PDw and T2w brain images. (a, b) the original (artifact free) images; (c, d) the motion corrupted images; (e, f) the magnitude of the difference between the original and the motion corrupted images; (g) the plot used to identify k-space lines corrupted by motion in the PDw and T2w data sets (solid line—PDw, dotted line—T2w); (h, i) initial guess images for POCS iterations; (j, k) the MAR reconstructed images; (l, m) the magnitude of the difference between the original and the MAR reconstructed images.

It was noticed that violation of k-space consistency that occurs due to accidental patient motion may cause serious image degradation and loss of diagnostic information. This can easily be seen in the comparison of the original and motion corrupted PDw and T2w brain images shown in Fig. 2. In this study, the background area is noticeably smaller than the background area in the phantom experiment. However the area is large enough to enable identification of the motion corrupted k-space lines, using the integrated energy plot shown on Fig. 2g. The images reconstructed from the combined k-space data sets (Fig. 2h, 2i), and used as the initial guess images for the corresponding POCS iterations, have clearly mixed contrast. The resulting PDw and T2w images have no motion related artifacts, and the contrast in each of the images corresponds to true PDw and T2w image contrast, which is apparent in the difference images (Fig. 2l and 2m).

Fig. 3 demonstrates the results of MAR reconstruction of PDw and T2w brain data corrupted by primarily through-plane rotation. The motion causes considerable artifact contamination in both images, but it is especially strong in the case of the T2w image where a considerable area of the frontal part of the brain is quite attenuated. The diagnostic value of both images is gravely diminished. The artifact manifestations in PDw and T2w images are distinct from each other because the corresponding k-space data sets were acquired by the FSE sequence with TS equal to one quarter

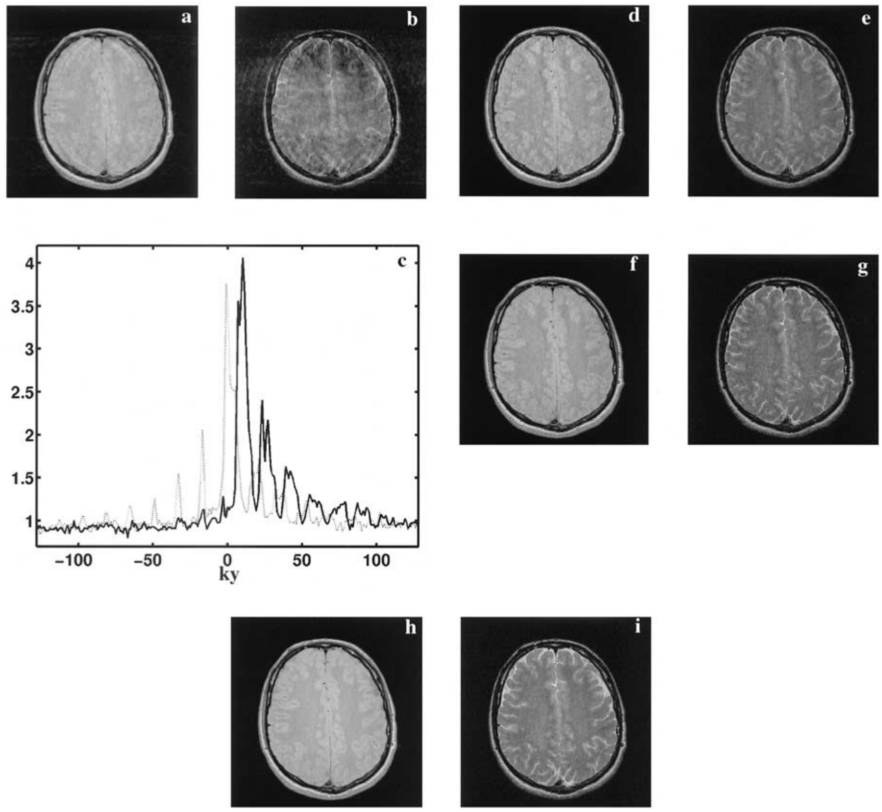

Fig. 3. PDw and T2w brain images. (a, b) the motion corrupted images; (c) the plot used to identify k-space lines corrupted by motion in the PDw and T2w data sets (solid line—PDw, dotted line—T2w); (d, e) the images reconstructed from the combined data sets; (f, g) the MAR reconstructed images; (h, i) the original (artifact free) images. Discrepancies between the original and the MAR reconstructed images were caused mainly by head movement between scans.

of the slice scan time. The motion corrupted k-space lines of PDw and T2w data sets can easily be detected using the described background based method. In contrast to the first two studies, the motion in this study corrupts k-space lines from both halves of T2w data set, including the center of k-space. The images used as initial estimates for the corresponding POCS iterations have residual artifacts and mixed contrast. However, after five POCS iterations the resulting PDw and T2w images show no visible artifacts and the contrast of the images is similar to true PDw and T2w image contrast.

The MAR reconstructed images have lower SNR than the inherent SNR of the corresponding motion corrupted images, but the diagnostic value of the MAR reconstructed PDw and T2w brain images is significantly higher than the diagnostic value of the corresponding motion corrupted images.

4. Discussion

The dual-contrast FSE sequence is the sequence of choice when acquisition of high resolution PDw and T2w images with perfect spatial coregistration is required. However, relatively long dual-contrast FSE imaging times make this sequence vulnerable to motion artifacts, which may cause noticeable degradation of image quality and significant loss of diagnostic information.

A motion artifact reduction technique for dual-contrast FSE acquisitions has been developed. The technique is based on POCS formalism and uses the k-space similarity and phase consistency between PDw and T2w images. The proposed MAR method is easy to implement and computationally efficient. The main limitation of the method is that it can only correct artifacts caused by accidental patient motion that violates only a part of k-space data (less than half). Thus, the method cannot be used to correct image degradation caused by motion that persists during the entire scan. The algorithm also requires that a phase encoding scheme of dual-contrast FSE pulse sequences should be implemented such that the same k-space lines of the PDw and T2w data are acquired by different echo trains. The best results for the described technique may be attained when there is no overlap between motion corrupted parts of PDw and T2w k-space data. In cases with partial overlap, the MAR technique, typically, significantly improves image quality, but the existence of some residual artifact is possible. We note that application of this technique will be limited to cases when images acquired using dual-contrast FSE sequences are corrupted by a global movement of the imaged object, such as a change in head position during brain study, swallowing during neck imaging, and so on. Spatially localized motions may not generate enough artifact energy to be detected by the first step of the MAR technique, thereby making artifact correction impossible. As we have found, the developed technique works quite well in brain studies but has not been tested in imaging other parts of the body.

The main distinctions of the proposed technique from earlier published postprocessing methods of global motion artifact correction [18–24] are the utilization of the similarity between PDw and T2w images and the absence of any assumptions about motion characteristics, except that the motion only occurs during a fraction of the scan time. The developed method discards the lines of k-space that are identified as corrupted by motion, and recovers them using a priori information about the acquired image. To identify motion corrupted k-space lines, the motion artifact manifestation on the image background and a periodic structure of the FSE phase encoding scheme are used. The main disadvantage of discarding corrupted data is a possible loss of valuable information about the image that could have been recovered and used to further improve the reconstructed image quality. However, the process of correcting the corrupted data may be time consuming and computationally difficult (e.g. correction of rotation usually needs interpolation or regridding) and usually requires either a priori knowledge about the motion characteristics (in-plane or through-plane, rotation or translation, or both, and so on) or modification of pulse sequences to incorporate navigator data acquisition for motion trajectory identification. A second disadvantage of the proposed algorithm is the reduced SNR of the corrected image due to the missing (discarded) measurements. But this result is typical for all images reconstructed from partial MR data and is usually acceptable, especially when this SNR loss is accompanied by a significant improvement in image quality.

Acknowledgments

We thank Sean Webb for editing the manuscript. This work is supported in part by NIH grants R01 HL48223 and R01 HL53596, and NIH BISTI grant 1P20HL68566-01.

References

[1] Henning J, Naureth A, Friedburg H. RARE imaging: a fast imaging method for clinical MR. Magn Reson Med 1986;3:823–33.
[2] Jones KM, Mulkern RV, Mantello MT, Melki PS, Ahn SS, Barnes PD, Jolesz FA. Brain hemorrhage: evaluation with fast spin-echo and conventional dual spin-echo images. Radiology 1992;182:53–8.
[3] Patola WB, Coulter BA, Conjoint DA, Chipperfield PM, Lingawi SS. A comparison of conventional spin-echo and fast spin-echo in the detection of multiple sclerosis. J Magn Reson Imaging 2001;13:657–67.
[4] Liu K, Margosian P. Multiple contrast fast spin-echo approach to black-blood intracranial MRA: use of complementary and supplementary information. Magn Reson Imaging 2001;19:1173–81.
[5] Madore B, Henkelman RM. Motion artifacts in fast spin-echo imaging. J Magn Reson Imaging 1994;4:577–84.
[6] Higushi N, Oshio K, Momoshima S, Shiga H, Melki PS, Mulkern RV, Jolesz FA. Two-contrast RARE: a fast spin-density and T2-weighted imaging method. J Magn Reson Imaging 1991;1:147.
[7] Melki PS, Mulkern RV, Panych LP, Jolesz FA. Comparing the FAISE method with conventional dual-echo sequences. J Magn Reson Imaging 1991;1:319–26.
[8] Mulkern RV, Wong STS, Winalski C, Jolesz FA. Contrast manipulation and artifact assessment of 2D and 3D RARE sequences. Magn Reson Imaging 1990;8:557–66.
[9] Constable RT, Gore JC. The loss of small objects in variable TE imaging: implications for FSE, RARE, and EPI. Magn Reson Med 1992;28:9–24.
[10] Kholmovski EG, Parker DL, Alexander AL. A generalized k-sampling scheme for 3D fast spin echo. J Magn Reson Imaging 2000;11: 549–58.
[11] Margosian P, Schmitt F, Purdy D. Faster MR imaging: imaging with half the data. Health Care Instrum 1986;1:195–7.
[12] Cuppen J, van Est A. Reducing MR imaging time by one-sided reconstruction. Magn Reson Imaging 1987;5:526–7.
[13] Noll DC, Nishimura DG, Macovski A. Homodyne detection in magnetic resonance imaging. IEEE Trans Med Imaging 1991;10:154–63.
[14] Youla DC. Mathematical theory of image restoration by the method of convex projection. In: Stark H, editor. Image Recovery: Theory and Application. Orlando: Academic Press, 1987. p. 29–77.
[15] Haacke EM, Lindskog ED, Lin W. A fast, iterative, partial-Fourier technique capable of local phase recovery. J Magn Reson 1991;92: 126–45.
[16] Liang ZP, Boada FE, Constable RT, Haacke EM, Lauterbur PC, Smith MR. Constrained reconstruction methods in MR imaging. Rev Magn Reson Med 1992;4:67–185.
[17] Kholmovski EG, Parker DL. Application of phase consistency to improve time efficiency and image quality in double echo black blood carotid angiography. submitted to J Magn Reson Imaging.
[18] Ehman RL, Felmlee JP. Adaptive technique for high-definition MR imaging of moving structures. Radiology 1989;173:255–63.
[19] Hedley M, Yan H, Rosenfeld D. An improved algorithm for 2D translational motion artifact correction. IEEE Trans Med Imaging 1991;10:548–53.
[20] Wood ML, Shivji MJ, Stanchev PL. Planar-motion correction with use of k-space data acquired in Fourier MR imaging. J Magn Reson Imaging 1995;5:57–64.
[21] Riek JK, Tekalp AM, Smith WE, Kwon E. Out-of-plane motion compensation in multislice spin-echo MRI. IEEE Trans Med Imaging 1995;14:464–70.
[22] Trouard TP, Sabharwal Y, Altbach MI, Gmitro AF. Analysis and comparison of motion-correction techniques in diffusion-weighted imaging. J Magn Reson Imaging 1996;6:925–35.
[23] Zoroofi RA, Sato Y, Tamura S, Naito H. MRI artifact cancellation due to rigid motion in the imaging plane. IEEE Trans Med Imaging 1996;15:768–84.
[24] Atkinson D, Hill DLG, Stoyle PNR, Summers PE, Keevil SF. Automatic correction of motion artifacts in magnetic resonance images using an entropy focus criterion. IEEE Trans Med Imaging 1997;16: 903–10.