Self-Navigated Interleaved Spiral (SNAILS): Application to High-Resolution Diffusion Tensor Imaging

Chunlei Liu,1,2 Roland Bammer,1 Dong-hyun Kim,1 and Michael E. Moseley1*

A fat-saturated twice-refocused spin echo sequence was implemented on a GE Signa 1.5-T whole-body system for diffusion-weighted imaging. Data were acquired using an analytically designed interleaved variable-density (VD) spiral readout trajectory. This flexible design algorithm allowed real-time prescription on the scanner. Each interleaf of the VD spiral oversampled the center of k-space. The oversampling provided an inherent motion compensation capability. The resultant diffusion-weighted images showed good quality without any retrospective motion correction. An iterated motion correction algorithm was developed to further reduce the signal cancellation artifact caused by motion-induced phase error. In this algorithm, a low-resolution phase map was estimated using the oversampled data in the center of k-space in order to correct for phase error in image space.

In vivo diffusion tensor imaging (DTI) studies were performed on the brains of healthy volunteers. High-quality isotropic diffusion-weighted images, trace maps, and FA maps from axial, sagittal, and coronal slices were obtained using a VD spiral readout trajectory with matrix size 256 × 256. To our knowledge, this was also the first time in vivo 512 × 512 DTI results were reported. Magn Reson Med 52:1388–1396, 2004. © 2004 Wiley-Liss, Inc.

Key words: magnetic resonance imaging; high resolution; diffusion; diffusion tensor imaging; variable density spiral; interleaved; motion correction

Diffusion-weighted imaging (DWI) is a unique technique for studying random molecular motion in biologic tissues. Over the past decade, DWI has found routine applications in medical diagnosis, especially in detecting acute cerebral ischemia (1). Most diffusion-weighted images are currently acquired using a single-shot echo-planar imaging (EPI) technique. Single-shot EPI has the advantage of rapid image acquisition and insensitivity to phase error caused by subject motion because the entire k-space is acquired with a single rapid train of gradient echoes. Despite the rapid image formation, single-shot EPI lasts long enough that $T_2^*$-decay limits image resolution and off-resonant spins can still cause serious image degradation.

To shorten the readout time, multishot sequences can be used (2,3); however, they generally suffer from view-to-view phase variations caused by motion during the period when the diffusion-sensitizing gradients are turned on. One approach to correct these variations is to acquire additional navigator data that can be used to resolve the phase error (4–7). The navigator can be implemented to correct for either one-dimensional or two-dimensional phase error. The navigator data are intended to provide a direct measure of the motion-induced phase variations. Under the assumption of rigid body motion, the data can be subsequently corrected for small amounts of motion.

A few studies have recently explored the self-navigating capability of the spiral readout trajectory in multishot DWI (8,9). Magnetic resonance imaging (MRI) based on spiral readout has been found to be effective in various applications, including functional neuroimaging (10) and spectroscopy (11). The spiral trajectory has the merit of moment-nulling motion compensation (12) and efficient use of gradient power (13). Conventional spiral readout trajectories have a limited potential for self-navigation because there are insufficient data in the central k-space, which effectively allows only zero- and (to some extent) first-order compensation. In order to improve the navigating capability, it is necessary to increase the sampling density at the center of k-space. With a recently developed analytical variable-density (VD) spiral design technique, the k-space sampling density can easily be manipulated and prescribed on the scanner hardware in real time (14). With this technique, each spiral trajectory serves effectively as a self-navigator that corrects for motion induced view-to-view phase variations.

There continues to be an increased demand for higher spatial resolution and diminished artifacts in diffusion-weighted imaging, which will eventually provide better lesion delineation in acute ischemic stroke and high-fidelity data for diffusion-tensor-based tractography. The purpose of this work is therefore to develop a reliable multishot diffusion-weighted VD spiral sequence and a reconstruction and navigation algorithm that effectively reduces motion artifacts. Applications to high-resolution (both 256 × 256 and 512 × 512) diffusion tensor imaging (DTI) are also addressed.

METHODS

Data Acquisition

A twice-refocused spin echo (TRSE) sequence was implemented (8) because of its well-documented ability to reduce eddy current distortions (15). A fat saturation pulse is added to reduce the off-resonance effect of the fat signal. Diffusion-encoding was achieved by applying a pair of bipolar field gradients around two 180° refocusing pulses.

*Correspondence to: Michael E. Moseley, Radiological Science Laboratory at the Richard Lucas MRS/I Center, Department of Radiology, Stanford University, 1201 Welch Road, Stanford, CA 94305-5488.
E-mail: moseley@stanford.edu

Received 12 May 2004; revised 12 July 2004; accepted 27 July 2004. DOI: 10.1002/mrm.20288
Published online in Wiley InterScience (www.interscience.wiley.com).

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A variable-density spiral readout trajectory (Fig. 1) that oversamples the origin of $k$-space was prescribed to collect data in $k$-space with intrinsic phase-navigation capabilities.

In Fig. 1, the durations of the four diffusion-sensitizing gradients are $\delta_1$, $\delta_2$, $\delta_3$, and $\delta_4$, respectively. They are chosen such that $\delta_1 = \delta_4$, and $\delta_2 = \delta_3$. The residual eddy currents introduced by switching field gradients can be minimized by varying the durations of the gradients (15). As suggested by Reese et al. (15), one can find the optimal durations for the $\delta$’s experimentally. However, this optimization is rather time consuming and less robust. Therefore, we have set up the timing of the pulse sequence so that maximum diffusion weighting is achieved for a given echo time (TE) without severely compromising the benefit of eddy current reduction. This observation has been confirmed by Reese et al. who demonstrated that close-to-minimum eddy currents can be achieved over a relatively broad range of sequence parameters (15).

The VD spiral trajectory is designed using a simple analytic method as proposed by Kim et al. (14). This method allows flexibility for controlling $k$-space sampling density and enables real-time prescription on scanners. Following Kim’s method, the variable density spiral trajectory can be expressed as

$$\begin{align*} \mathbf{k}(\tau) &= \lambda \tau^e e^{j\omega \tau}, \quad [1] \end{align*}$$

where $\tau$ is a function of time and is in the range of $[0,1]$, $\omega = 2\pi n$ and $n$ is the number of turns of the spiral. Given the desired matrix size $N$, and field of view (FOV), $\lambda$ is given by $\lambda = N/(2 \times \text{FOV})$ when scanning on a $2\times$ grid. The resultant trajectory normally starts in the slew-rate-limited region and ends in the amplitude-limited region as given by (14)

$$\begin{align*} \tau(t) &= \begin{cases} 
\frac{\gamma s_m}{\lambda \omega \alpha^2} \left( \frac{\alpha}{2} + 1 \right) t, & 0 \leq t \leq \min(T_{2s}, T_{a}) \\
\frac{\gamma s_m}{\lambda \omega} \left( \frac{\alpha}{2} + 1 \right) t^{1/(\alpha + 1)}, & T_{2s} \leq t < T_{a} 
\end{cases} \quad [2]
\end{align*}$$

where $\gamma$ is the gyromagnetic ratio; $s_m$ and $g_m$ are, respectively, the maximum achievable slew rate and gradient amplitude; $T_{2s}$ is the transition time from slew-rate-limited region to amplitude-limited region; $T_{a}$ is the ending time of a slew-rate-limited trajectory. $\alpha$ is a constant for controlling the change of sampling density in the radial direction. Note that when $\alpha$ equals 1, the VD spiral trajectory is identical to a regular spiral trajectory.

Figure 2 shows a sample waveform design with the...
following parameters: $N = 256$, FOV = 22 cm, number of interleaves = 28, and $\alpha = 4$. As shown in Fig. 2, the center of $k$-space is oversampled. For example, Fig. 2c illustrates that there are 52 sampling points that will be gridded to the central pixel alone.

**Phase Correction and Image Reconstruction**

The algorithm used for image reconstruction is illustrated in Fig. 3. The data acquired on the interleaved spiral trajectories are first resampled onto a Cartesian grid of size $2N \times 2N$ using a standard gridding algorithm with density precompensation. The sampling density compensation function is calculated using the Voronoi approach (16). After the gridding step, phase correction is performed in image space for each interleaf.

Anderson and Gore have shown that rigid-body motion of a scanned subject during the diffusion sensitizing gradients will introduce a linear phase term in image space (4). While this phase term yields no consequences in single-shot EPI, this phase error is a more severe problem in an interleaved acquisition because the motion-induced phase accrual can be different from interleaf to interleaf. As a result, spatially dependent signal cancellation can occur in the image, causing severe artifacts. Therefore, motion-induced phase errors must be removed before combining all the interleaves.

In the scheme outlined in Fig. 3, the phase error is estimated from individual VD spiral interleaves using a low-resolution image, which can be reconstructed using the center portion of $k$-space. A Gaussian window function is applied before the Fourier transformation in order to reduce ringing effects. Since the center of $k$-space is oversampled, for each interleaf a low-resolution image can be reconstructed without aliasing artifacts. The low-resolution phase is then removed from the aliased high-resolution image. This phase correction procedure can be iterated several times until no significant change in the phase is observed from iteration to iteration. In our study, we found satisfying convergence after two iterations.

As described by both Anderson and Gore and Butts et al., motion-induced phase accrual causes the readout trajectory to usually deviate from the intended pathway and not start from the origin of $k$-space. By correcting the phase in the image space, this displacement in $k$-space can be corrected. As illustrated in Fig. 4, the corrected $k$-space data after two iterations show that after correction: (1) the peak of the signal is brought back to the origin of $k$-space (Fig. 4b and c) and (2) the signal loss in the reconstructed images is also restored (Fig. 4e and f). The image-based phase correction also corrects the nonlinear phase that exists in the low resolution phase estimate (17). Nonlinear phase includes, for example, the phase of the object itself. Nonlinear phase correction in image space results in more than just a simple shift $k$-space. For example, the intensity of the $k$-space data can be modified after phase correction (Fig. 4a–c). Alternatively, one can correct solely for the linear phase by extracting a linear phase.
term from the low-resolution image (7). However, Miller and Pauly show that nonlinear phase correction can further restore signal loss and improve image quality, since nonlinear phase in diffusion-weighted images can arise from nonrigid motion (17).

Experiment

The interleaved VD TRSE diffusion-weighted sequence was implemented on a GE Signa 1.5-T whole-body system equipped with a high-performance gradient set (50 mT/m per gradient axis, slew rate = 150 mT/m/msec). In vivo diffusion measurements were performed on the brains of nine healthy volunteers. All experimental scans were approved by the institutional review board. All subjects signed an informed consent after the nature of the study was explained to them and fully understood. The following parameters were used: TR/TE = 2.5 sec/67 msec, a total of 20 slices with a 6.5-mm slice thickness and a zero

FIG. 4. Comparison of (a–c) the k-space data of one interleaf and (d–f) corresponding reconstructed images before and after phase correction. Two white lines in (a–c) indicate the coordinate system. (a and d) Before correction, the k-space data are displaced due to motion during the diffusion sensitizing gradients and signal cancellation occurs in the image. (b and e) After one iteration of correction. (c and f) After two iterations of correction, the peak has been shifted back to the origin of the k-space and signal loss in the image space has been restored.

FIG. 5. Variable density spiral trajectory has self-navigating capability. (a) $b = 0$ image and three $b = 800$ sec/mm$^2$ images acquired with conventional spiral and reconstructed without motion correction. The three diffusion gradient directions are $[0 1 -1], (-1 1 0), (1 0 -1)$. (b) Images acquired with conventional spiral and reconstructed with motion correction. (c) Images acquired with VD spiral and reconstructed without motion correction. (d) Motion-corrected images reconstructed using the same dataset as (c). The signal loss appears in (c) has been restored. All data were acquired during the same study with the same diffusion encoding.
gap in between, \( b = 800 \text{ sec/mm}^2 \), FOV = 22 cm, acquisition matrix size = \( 256 \times 256 \), BW = ±125 kHz, \( \alpha = 4 \), and number of interleaves = 28. The duration of the four diffusion gradients was: \( \Delta_1 = \Delta_4 = 8.4 \text{ msec} \) and \( \Delta_2 = \Delta_3 = 12.3 \text{ msec} \). The readout time was 13.5 msec for each interleaved acquisition. To acquire the diffusion tensor, the scan time for acquiring six DW images plus an unweighted image was 8.1 min per acquisition for whole brain coverage. This scan was repeated four times, and the total scan time was about 32 min. To illustrate the effect of motion correction, images were reconstructed both with and without applying the algorithm described in the previous section.

To illustrate the benefit of oversampling the center of \( k \)-space, we also implemented a similar sequence with a conventional spiral readout trajectory. The conventional spiral was designed following the algorithm by Glover (18). A total of 16 interleaves were used. The readout time was 13.2 msec per interleaved image, which was comparable with that of the VD spiral. All other parameters were the same as the VD spiral protocol.

To evaluate the potential of the VD spiral sequence for applications of high-resolution DTI, axial slices with a matrix size of \( 512 \times 512 \) were also prescribed. The DW images were acquired with 64 interleaves of VD spiral within a FOV of 24 cm. Each interleaved image had a readout time of 20.3 msec. The slice thickness was 8 mm. All other parameters were kept identical to the \( 256 \times 256 \) protocol. The scan time for acquiring six DW images plus an unweighted image was 18 min per acquisition for whole brain coverage. In order to acquire three scans for averaging, the total scan time was about 55 min.

The diffusion tensor was calculated using six independent diffusion-weighted images and a non-diffusion-weighted image (19). A color-coded fractional anisotropy (FA) map was also created for each slice using the directions of the three eigenvectors (20). Color coding was as follows, for axial slices: red indicated the anterior–posterior direction; green indicated left–right direction; and blue indicated the superior–inferior direction. For sagittal slices red represented the direction of superior–inferior; green represented anterior–posterior; and blue represented left–right. For coronal slices red represented the direction of superior–inferior; green represented left–right; and blue represented anterior–posterior.

**RESULTS**

**Motion Correction**

Figure 5a and b shows images reconstructed with and without motion correction using the data acquired by a conventional spiral readout trajectory. Figure 5a and b clearly demonstrates that motion artifacts were not effectively removed in the diffusion-weighted images because of insufficient sampling of the \( k \)-space origin. As a consequence, severe signal loss caused by phase variations from interleaf to interleaf appears in various regions.

Figure 5c and d shows images reconstructed with and without motion correction using the data acquired by VD spiral. The self-navigating capability of VD spiral is evident. Even the uncorrected images acquired by VD spiral have superior quality compared to motion-corrected images acquired by conventional spiral. By oversampling the center of \( k \)-space, the VD spiral provides an inherent motion compensation capability. Image qualities are comparable when there is no diffusion-weighting (i.e., \( b = 0 \) images in Fig. 5c and d). In the diffusion-weighted images, motion-induced phase error during the diffusion-encoding

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**FIG. 6.** In vivo high-resolution (256 × 256) diffusion-weighted images averaged over four acquisitions. The images are obtained using a 28-interleaf VD spiral TRSE sequence with \( b = 800 \text{ sec/mm}^2 \). (a–f) Diffusion encoding directions are (1 1 0), (1 0 1), (0 1 −1), (−1 1 0), (0 1 1), and (1 0 −1), respectively.
gradients varies from interleaf to interleaf. Without motion correction, all the diffusion-weighted images are corrupted to a certain degree (Fig. 5c). After two iterations of motion correction, signal loss is restored and image artifacts are greatly reduced (Fig. 5d).

High-Resolution Diffusion Tensor Imaging

A full set of diffusion-weighted images from an axial slice is shown in Fig. 6. Because of the high-resolution scan, images were averaged over four acquisitions to increase the signal-to-noise ratio (SNR). Compared to single-shot EPI, these images provide a greater level of detail and better delineation of neuroanatomic structures. One great advantage of the interleaved spiral readout is the remarkable reduction of geometrical distortions in the DW images as shown in Fig. 6.

Figure 7 shows the isotropic-weighted image, the trace map, the FA map, and the color-coded FA map for three axial slices. DTI results from one sagittal slice and two coronal slices are given in Fig. 8. Orientation of the white matter tracts can be clearly identified in the colored FA map.

Figure 9 demonstrates a $512 \times 512$ isotropic DW image, the corresponding trace map, and the FA map. The SNR can be improved by more signal averaging. To our knowledge, this is the highest resolution reported so far for in vivo DTI in humans.

FIG. 7. In vivo high-resolution (256 × 256) DTI results acquired with a VD spiral TRSE sequence in axial slices. (a) $b = 0$. (b) Trace of diffusion tensor. (c) FA map. (d) Color-coded FA map. Red represents the direction of anterior–posterior; green represents left–right; and blue represents superior–inferior. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]
DISCUSSION

We have demonstrated the acquisition of high resolution (both 256×256 and 512×512) diffusion-weighted images using a self-navigated interleaved spiral (SNAILS) sequence. The radial nature of a spiral trajectory reduces artifacts by spreading them evenly in all directions. It also decreases most geometric distortions that appear frequently in images obtained by EPI sequences. The sequence prevents image warping caused by eddy currents, which is a severe problem for single-shot EPI DWI. Image warping in single-shot EPI usually produces considerable misregistration among DWI scans with diffusion encoding along different directions. It also causes misregistration between conventional structural imaging and fMRI scans.

The reconstruction algorithm uses 2D navigator information provided by the VD spiral to correct the data from interleaf to interleaf. Hence, no additional data need to be collected for navigator information. The resultant images have high SNR and greater contrast between white matter and gray matter. As a result, fine structures, such as the layers of the tapetum in the optic radiations (Figs. 6 and 7) and the fiber structures in the thalamus (Fig. 8), can be observed in both the diffusion-weighted images and the FA map.

FIG. 8. In vivo high-resolution (256×256) DTI results from sagittal (first column) and coronal (second and third columns) slices. b = 800 sec/mm². (a) Isotropic DWI. (b) Trace of diffusion tensor. (c) FA map. (d) Color-coded FA map. For sagittal slices, red represents the direction of superior–inferior; green represents anterior–posterior; and blue represents left–right. For coronal slices, red represents the direction of superior–inferior; green represents left–right; and blue represents anterior–posterior.
The SNAILS acquisition of diffusion-weighted images has a substantial benefit in reducing the sensitivity to subject motion. SNAILS is a self-navigated technique in the sense that it oversamples the origin of $k$-space and therefore allows monitoring and correcting for trajectory errors. Since $k$-space shifts introduce a smoothly varying phase in the image domain, they are easier to correct using low-resolution navigator images. When VD spiral is implemented in an interleaved fashion, the center of $k$-space is oversampled and the outer portion of $k$-space is sampled below the Nyquist rate for each interleaf. This motion compensation capability is evident in the comparative evaluation made between VD and conventional spiral. Images acquired by VD spiral and reconstructed without retrospective motion correction have significantly fewer motion artifacts than images acquired by conventional spiral technique and reconstructed with retrospective motion correction (Fig. 5).

Although SNAILS has this inherent motion compensation capability, when images from different interleaves are combined directly without motion correction, signal cancellation still occurs due to destructive interference (Fig. 5c). These image artifacts also depend on the diffusion encoding direction. The most severe artifacts appear when the gradient is applied in the S/I direction (data are not shown here). This is most likely because this is the primary direction of pulsatile motion in the brain (4). These residual motion artifacts can be further reduced by applying phase correction in image space (Fig. 5d). A potential problem of this approach is that this does not correct for the phase of aliased high-frequency component. As pointed out by Pipe et al. (21), in order to allow correction for the aliased component, both the aliased and the unaliased components require the same phase at any location, which generally is not true. Fortunately, this does not cause a serious problem in our case. The aliased high-frequency component has significantly lower energy than the unaliased component. Furthermore, the aliased components are not in phase from interleaf to interleaf because they contain uncorrected motion-induced phased errors that vary from interleaf to interleaf. The aliased energy can be effectively suppressed by averaging over a certain number of acquisitions. Miller and Pauly have shown effective suppressions by averaging over as few as four acquisitions (17).

The proposed iterated phase correction scheme can effectively remove the phase error in the image space for each interleaf. However, misregistration between interleaves caused by gross patient motion remains a potential problem. Since each VD spiral interleaf provides a unaliased low-resolution image, similar coregistration methods such as those used by Pipe (22) can also be applied to compensate for gross involuntary patient movements. Although gross patient motion was not a big problem in our experiment, it is frequently seen in acute stroke patients. If motion occurs, it is unlikely that these coregistered images could allow accurate computation of diffusion tensor measures; however, despite patient movements, coregistration could contribute significantly to the quality of diagnosing and screening stroke patients.

SNAILS also allows flexibility to change the sampling density of the spiral trajectory and prescribe the desired number of interleaves. The interleaved acquisition shortens the readout time and, therefore, helps to reduce blur-

FIG. 9. In vivo 512 x 512 DTI results in one axial slice acquired with 64-interleaf VD spiral. (a) $b = 0$. (b) Isotropic diffusion-weighted image, $b = 800 \text{ sec/mm}^2$. (c) Trace of diffusion tensor. (d) FA map.
ring artifacts that result from off-resonant spins. The sequence can be improved by determining the optimal number of interleave factors in the center of $k$-space, and the VD pitch factor $(\alpha)$ when given the desired image resolution. This optimization requires a tradeoff between SNR and scan time. For an image acquisition matrix of $256 \times 256$, our initial study has found that no significant further improvement is achieved by increasing the number of interleave factors to a number larger than 32 for the particular images shown in Fig. 6. At locations where susceptibility effects are severe, it is beneficial to increase the number of interleave factors and further shorten the readout time. However, these additional interleave factors would prolong the data acquisition time. For more rapid data acquisition, parallel imaging techniques, such as the SENSE technique, can be used in conjunction with the interleaved spiral readout trajectory (23).

The SNAILS DWI technique can find potential applications in other body regions where motion is unavoidable due to its motion compensating capability, such as breasts, prostate, and the solid organs in the abdomen. Moreover, this technique is much better suited to imaging the skull base or posterior fossa than EPI due to its significantly reduced sensitivity to off-resonant artifacts. This dramatic increase in spatial resolution is a great advantage over single-shot EPI methods and will improve both structural DW imaging and fiber tracking. Of course, a greater level of detail requires extra scanning time. In addition to high-resolution SNAILS scans, multi-shot SNAILS scans with identical acquisition matrices as EPI demonstrate much less distortion with only a moderately increased scan time.

In summary, SNAILS DWI in combination with improved navigation and reconstruction techniques demonstrates superior image quality and poses an excellent alternative to single-shot EPI DWI.

ACKNOWLEDGMENTS

The authors thank Dr. Gary Glover at Stanford University for graciously providing the basic spiral pulse sequence upon which our sequence is developed. The authors also thank Karen Y. Chen, MS, and Rebecca Rakow Penner, MS, for proofreading the manuscript.

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