Rapid Isotropic DWMRI at 1.5 Tand 3T with Radial-FSE

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Introduction

Multi-shot radial fast spin-echo (radial-FSE) methods enable diffusion-weighted MRI (DWMRI) to be carried out with high spatial resolution, without sensitivity to bulk motion and/or susceptibility [1]. An interesting feature of radial datasets is that all radial lines pass through the origin of Fourier space and contribute more or less equally to the intensity and contrast in the reconstructed image. When different lines have various weightings, images reconstructed from the full dataset will have the average weighting of all the lines. We have taken advantage of this feature by introducing variations in the diffusion weighting directions within the radial-FSE sequence. If the appropriate diffusion weighting and view ordering are implemented, reconstructed images have intensity that is, in effect, scaled by the trace of the diffusion tensor. Because this sequence employs FSE refocusing, images can be obtained at high magnetic field strengths without accentuating susceptibility artifacts. Imaging of volunteers and patients has been carried out at 1.5 T and 3.0 T for demonstration.

Methods

The isotropic radial-FSE sequence is identical to that previously published [2] except that the diffusion weighting is cycled through different diffusion directions during the scan: XYZ, -XYZ, X-YZ, -X-YZ. Using full gradient strength on all three axes simultaneously allows a maximal b-value to be obtained in the shortest possible TE, which is not possible with other isotropic weighting techniques. The order in which radial lines (views) are collected, needs to be controlled such that 1) all radial lines acquired within a given TR period are spread out about the full 2π radians of Fourier space, 2) there is a high frequency variation of TE and diffusion weighting direction with view angle and 3) there is no correlation between diffusion weighting direction and TE. By accounting for these criteria, images can be obtained that minimize artifacts due to motion, T2 decay and diffusion anisotropy. In addition, the images are self registered because all the data is collected in one scan. If N total views are collected, the number of views corresponding to a particular diffusion direction is N/M where M is the number of diffusion directions. Images reconstructed from such datasets, are essentially an arithmetic average of images obtained with diffusion weighting in individual directions.

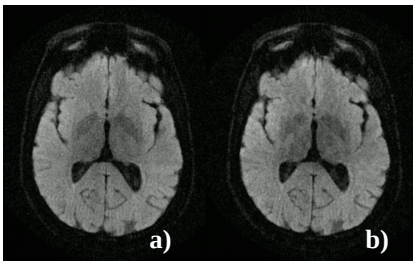

Fig 1. a) Trace image; **b)** Isotropic diffusion-weighted image.

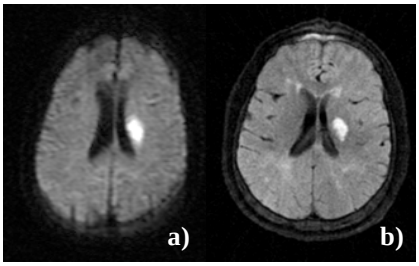

Fig 2. Diffusion-weighted images of a stroke patient acquired with **a)** Trace SSEPI and **b)** isotropic radial-FSE. For a), TE=76ms, TR= 10s, FOV= 30 X 24cm. For b), TE=83ms, TR=1500ms, ETL=4, FOV=26cm.

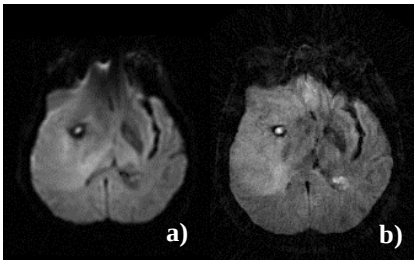

Fig 3. Diffusion-weighted images acquired of a cancer patient using **a)** Trace SSEPI and **b)** isotropic radial-FSE. For a), TE= 72ms, TR= 10s, FOV= 30 X 20cm. For b), TE=75ms, TR=3000ms, ETL=4, FOV=26cm.

Results

Isotropic and conventional diffusion-weighted radial-FSE sequences were implemented on a GE Signa LX 1.5T echospeed and GE Signa VH3 3.0T MRI scanners (General Electric Medical Systems, Milwaukee, WI) with actively shielded gradients capable of 33 and 40 mT/m, respectively. Images of the brain of a normal volunteer at 1.5T are shown in **Fig 1**. The trace image in Fig. 1a is the geometric mean of four individual images obtained with constant diffusion weighting (b = 1000 s/mm^2) in the XYZ, -XYZ, X-YZ, -X-YZ directions. The isotropic image was generated from a single radial dataset within which the diffusion weighting direction was cycled through the above directions throughout the scan. Four identical images were averaged to obtain equivalent SNR as the trace image. The images are visually similar because of the effective averaging of anisotropic structures. Identical ROIs in the right side of the corpus callosum (an area of high anisotropy) were tabulated from each image; the mean signal intensity of the ROI was variable between directions, but the values in the trace image and isotropic image are statistically identical (84.14 in the trace image and 85.86 in the isotropic image).

Images of a stroke patient obtained using isotropic radial-FSE and conventional single-shot echo planar imaging (SSEPI) acquired at 1.5T are shown in **Fig. 2**. The tissue affected by the stroke appears bright due to the reduced apparent diffusion coefficient (ADC) associated with acute ischemia. The SSEPI images exhibit low spatial resolution and the region of ischemia is blurred. Visible ghosting in the SSEPI images is most likely due to misregistration of images with different diffusion weighting. The isotropic radial-FSE image has higher spatial resolution and is self registered, allowing better definition of the region of ischemia. Also, there is no variation in signal intensity due to diffusion anisotropy and there are no significant artifacts due to motion and/or magnetic field inhomogeneities. Diffusion-weighted (b = 1000 s/mm^2) images of a cancer patient with a renal cell carcinoma metastasis in the right hemisphere obtained at 3T are shown in **Fig. 3**. The SSEPI image contains a noticeable magnetic field inhomogeneity artifact in the frontal lobe. Due to the high-resolution and lack of distortion in the isotropic radial-FSE image, diffusion properties in and around the tumor can be accurately determined.

Conclusion

The ability to obtain radial-FSE images with isotropic diffusion weighting in a single exam increases imaging speed by a factor of two, i.e. data for the calculation of ADC maps are obtained in two exams as apposed to four. This reduction in imaging time, combined with the self registered, high-resolution images increases the practical usefulness of radial-FSE exams in clinical DWMRI and makes them particularly useful at high magnetic field.

References: [1] Trouard, *et al*., *M.R.M*., 42, 11-18, 1999. [2] Theilmann, *et al., Proc. of the 10th ISMRM*, 0078, 2002.

Acknowledgements: This work supported by NIH grant R21 AG021624.