Diffusion-weighted imaging of the spine with readout-segmented (RS)-EPI

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Introduction: The development of diffusion-weighted imaging (DWI) of the spine has been hindered by the large off-resonance effects problematic for single-shot EPI. Overcoming these restrictions would open up applications ranging from the diagnosis of spinal cord infarction, vertebral body fractures, lymphoma, trauma, and many other disorders. Navigated interleaved EPI has been shown to overcome this distortion problem both in the brain [1] and in the spine [2]. However, interleaved EPI may lead to undersampling effects in k-space leading to ghosting [3]. Readout-segmented (RS)-EPI has recently been shown to be a promising pulse sequence for high-resolution DWI of the human brain, with significantly reduced distortions [4-6]. RS-EPI uses several 'blinds' (each coupled with a central blind navigator) which are adjacent in the readout direction. The shorter echo-spacing – or, in other words, the increased pseudo bandwidth along the phase encoding direction – reduces both geometric distortions and image blurring. Since each blind itself is consistent (that is, acquired at full FOV), this trajectory also avoids ghosting in the presence of motion that occurs between shots in interleaved EPI.

The objective of this study was to compare DW images acquired with RS-EPI and EPI. Here we produce images with significantly reduced distortions compared with EPI.

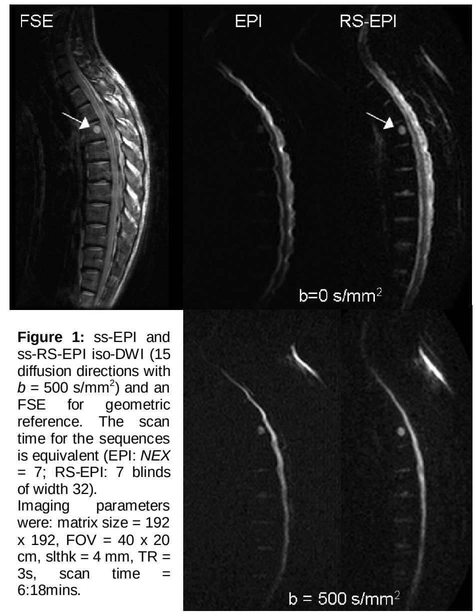

Figure 1: ss-EPI and ss-RS-EPI iso-DWI (15 diffusion directions with b = 500 s/mm^2) and an FSE for geometric reference. The scan time for the sequences is equivalent (EPI: *NEX* = 7; RS-EPI: 7 blinds of width 32). Imaging parameters were: matrix size = 192 x 192, FOV = 40 x 20 cm, slthk = 4 mm, TR = 3s, scan time = 6:18mins.

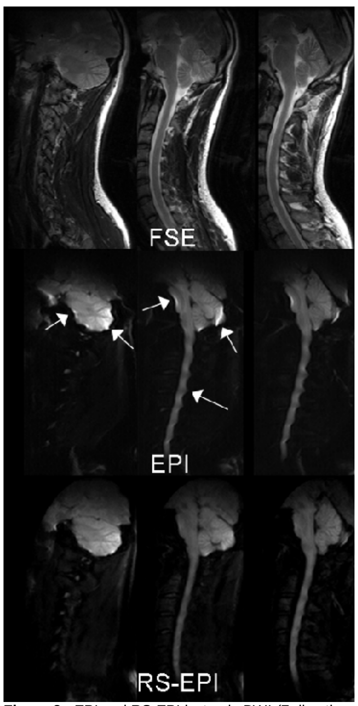

Figure 2: EPI and RS-EPI isotropic DWI (7 directions with b = 500 s/mm^2) and an FSE for geometric reference. As in Fig. 1, the scan time for the sequences is equivalent (2:27mins). Imaging parameters were: matrix size 192 x 192, FOV = 30 x 15 cm, slthck = 4 mm, TR = 3 s. The echo spacing was 316μs for RS-EPI and 728μs for EPI.

Materials & Methods: Diffusion-weighted images were acquired on three volunteers on a 3T whole-body GE Excite system using a four-channel spine coil. One volunteer presented with an asymptomatic hemangioma in the vertebrae. Single-shot (ss)-EPI and ss-RS-EPI images were acquired in the thoracic spine as well as cervical spine region. Parallel imaging was not used due to the coil geometry. Thus, the phase encoding direction was applied in the A/P direction and an anterior graphical saturation pulse was applied to avoid wrapping (due to phase aliasing and motion artifacts from the heart). Both methods used a matrix size of 192x192, TE = minimum (RS-EPI: 61 ms, EPI: 84ms), 32 overscans, slthck = 4 mm, TR = 3s. RS-EPI used 7 blinds of size 32x192 (freq.×phase), and EPI used 7 NEX to keep the scan time equivalent. Images of the thoracic spine were acquired on two volunteers, consisting of 3 b = 0 s/mm^2 and 15 isotropically distributed DW directions with b = 500 s/mm^2, a rectangular FOV = 40 x 20 cm, and a scan time of 6:18mins. For the second volunteer, a b = 0 s/mm^2 image was acquired with a negative phase encoding gradient, and – with the use of the b = 0 s/mm^2 image from the positive phase encoding gradient – this data was used to perform distortion correction [7]. The cervical spine scan consisted of one b = 0 s/mm^2 and seven b = 500 s/mm^2 diffusion directions at a FOV = 30 x 15 cm, and a scan time of 2:27mins. For all scans, the theoretical reduction in distortion with RS-EPI is 57%. The ghost calibration parameters [8] were calculated from the center blind of the center slice of the b=0 volume and were applied to all b = 0 and b = 500 s/mm^2 volumes. This was followed by ramp sampling correction, and phase correction using triangular windowing [9,10]. The partial Fourier data were reconstructed with POCS [11,12], prior to gridding [13] and sum-of-squares over coils. FSE images were also acquired for geometric comparison. Note that for all data, gradient warp correction was not used.

Results: A comparison between the b = 0 s/mm^2 and isotropic b = 500 s/mm^2 EPI and RS-EPI images of the spine is shown in Figure 1. The RS-EPI trajectory significantly reduces the "zig-zag" appearance of the spinal cord compared with EPI (cf. FSE image, left panel). Incidentally, hemangioma in the vertebrae (white arrows) is more clearly delineated with RS-EPI. Likewise, the isotropic RS-EPI-DW images of the cervical spinal cord in Figure 2 have reduced distortions compared with EPI, particularly at tissue-air interfaces. Fig. 3 shows distortion corrected ss-RS-EPI data. As indicated by the red line drawn on the FSE for geometric reference, further improvement in the distortion properties can be observed.

Discussion: EPI-based DWI of the spine has traditionally been difficult because of off-resonance effects and large FOVs. This work shows that RS-EPI can be useful for imaging the spine, with considerably reduced geometric distortions (Figs. 1-3). This method can be used to acquire high resolution DW images in a clinically reasonable scan time. Compared with EPI, the scan time increase is 7-fold. However, compared with PROPELLER FSE – a sequence which may be another useful candidate for spine DWI – the reduction in scan time is significant. Another advantage of RS-EPI is that the target resolution is independent from geometric distortion – it is only dependent upon the blind width and FOV. If further reduction in distortion is warranted, one may decrease the blind width (at the expense of more blinds and a longer scan time), and/or perform distortion correction (Fig. 3) with the use of an image acquired with a negative phase-encoding gradient. Future work could be to replace the spectral spatial pulse used here with selective excitation [14-16] to avoid the imperfect saturation pulse; and to investigate the use of inversion recovery methods such as STIR, SPIR, and SPAIR [17] to achieve better fat saturation.

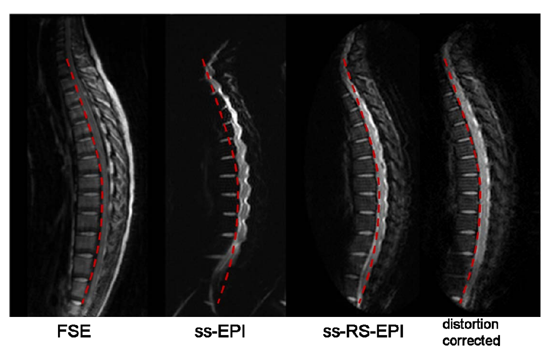

Figure 3: Images (b = 0s/mm^2) on a second volunteer acquired as in Fig. 1, showing a comparison between ss-EPI and ss-RS-EPI. On the far right the ss-RS-EPI image has been distortion corrected with the use of a b = 0 image acquired with a negative phase-encoding gradient.

References: [1] Butts K. MRM 1996;35:763-770. [2] Bammer R. JMRI 2002;15:364-373. [3] Atkinson MRM 2000;44:101-109. [4] Porter D. ISMRM 2004;442. [5] Porter D. ISMRM 2008;3262. [6] Holdsworth SJ. ISMRM 2008;757. [7] Andersson JL. Neuroimage 2003;20(2):870-888. [8] Nordell A. ISMRM 2007:1833. [9] Pipe J. MRM 1999;42(5):963-969. [10] Holdsworth SJ. ISMRM 2008;4. [11] Haacke JMR 1991;92:126-145. [12] Liang Rev MRM 1992;4:67-185. [13] Jackson IEEE TMI 1991;10:473-478. [14] Mansfield J. Phys E: Sci Instrum 1988(21):275-280. [15] Rieseberg S MRM 2002;47(6):1186-1193. [16] Saritas E. MRM 2008;60:468-473. [17] Mürtz Eur Radiol 2007;17:3031–3037. **Acknowledgements:** This work was supported in part by the NIH (2R01EB002711, 1R01EB008706, 1R21EB006860), the Center of Advanced MR Technology at Stanford (P41RR09784), Lucas Foundation, Oak Foundation, and the Swedish Research Council (K2007-53P-20322-01-4). We would also like to thank Greg Zaharchuk for his useful advice.