Diffusion-Weighted Inner-Field-of-View EPI Using 2D-Selective RF Excitations with a Tilted Excitation Plane

J. Finsterbusch[1,2]
[1]Department of Systems Neuroscience, University Medical Center Hamburg-Eppendorf, Hamburg, Germany, [2]Neuroimage Nord, University Medical Centers Hamburg-Kiel-Lübeck, Hamburg-Kiel-Lübeck, Germany

Introduction

2D-selective RF (2DRF) excitations [1,2] can be used to acquire inner field-of-views (FOVs) without aliasing in the phase-encoding direction which, e.g., in echo-planar imaging (EPI) reduces geometric distortions induced by magnetic field and susceptibility inhomogeneities [3]. Recent applications demonstrated the feasibility of inner-FOV EPI for high-resolution diffusion-weighted imaging, e.g. in the human spinal cord [4,5]. In these approaches the unwanted side excitations of the 2DRF excitations appeared either in the phase-encoding or in the slice direction which requires to position them outside of the object or beyond the slice stack to measure, respectively. However, for large objects or many slices, this may yield rather long 2DRF pulses which not only enhances their sensitivity to off-resonance effects but also increases the echo time and related signal losses. In this study, it is shown that tilting the excitation plane to position the side excitations in the dead corner between the slice stack to acquire and the current image section (see Fig. 1) represents a simple and robust method to suppress unwanted signal contributions because in such a setup the side excitations neither are refocussed nor do they overlap with neighboured sections. For larger objects or many slices to acquire, this approach can reduce the 2DRF pulse durations and the echo time considerably and, thus, increase the SNR significantly as is demonstrated in the human spinal cord in vivo.

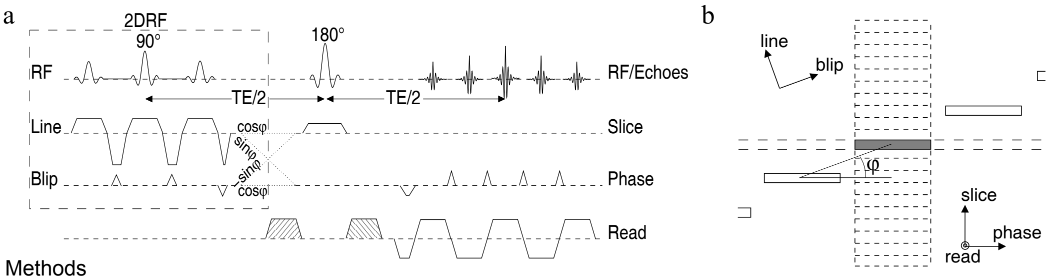

Fig. 1: (a) Basic pulse sequence for diffusion-weighted (hatched gradients) EPI of inner FOVs using a 2DRF excitation based on a fly-back blipped-planar trajectory with the excitation plane tilted by an angle of φ. (b) Geometry of the main excitation (grey filled box), the unwanted side excitations (white filled boxes), the image section (long dashes), i.e. the slice covered by the refocussing RF pulse in (a), and the slice stack (short dashes)

Methods

Measurements were performed on a 3T whole-body MR system (Siemens Magnetom Trio) using a 12-channel head coil (phantom experiments), a four-channel neck coil (cervical spinal cord), and a 18-channel spine array coil (thoracic and lumbar spinal cord). Spatial resolutions of $0.9\times0.9\times5\,mm^3$ and $0.45\times0.45\times5\,mm^3$ were used for the diffusion-weighted and the T2*-weighted (MEDIC) acquisitions. Diffusion-weighted measurements were performed with *b* values of $750\ s\ mm^{-2}$ (cervial/thoracic spinal cord) and $500\ s\ mm^{-2}$ (lumbar spinal cord) in six non-collinear directions, a repetition time of 4.5 s, and 24 magnitude averages yielding a total acquisitions time of 12.6 min. 2DRF excitation (flip angle 90°) were based on a fly-back blipped-planar trajectory (see Fig. 1a) with a resolution of $5\times10\ mm^2$ and were designed under the low-flip-angle approximation [2]. An inner FOV of 40 mm was desired in the phase-encoding directions which required 14 mm oversampling to account for the profile transition region (see Fig. 2). The distance of the side excitations was 290 mm (29 k-space lines, 25.5 ms) for the untilted geometry ($\varphi = 0°$) which is compatible with lower spinal cord acquisitions in heavier patients, and 70 mm (7 k-space lines, 5.9 ms) for the tilted geometry ($\varphi = 20°$) to avoid unwanted signal contributions from side excitations (see Fig. 1b). A water phantom and healthy volunteers from which informed consent was obtained prior to the examination, were investigated.

Results

Profile acquisitions show that the side excitations are suppressed (Fig. 2) and do not interference with other slices of the stack (data not shown). The shorter 2DRF pulses for the tilted geometry are less sensitive to profile distortions in the presence of magnetic field inhomogeneities and yields reduced echo times (-20 ms) which increases the SNR by about 20% (Fig. 3). Figure 4 shows results of the DTI acquisitions in the thoracic and lumbar spinal cord of a healthy volunteer. Some residual geometric distortions can be seen in the EPI images, e.g., in comparison to the MEDIC images.

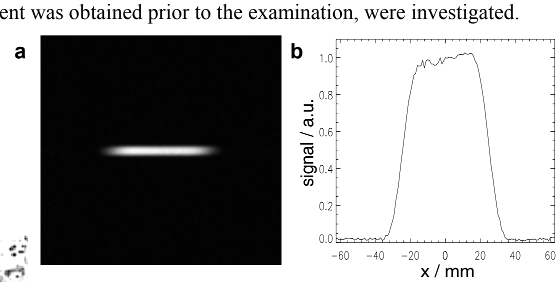

Fig. 2: Profile acquisitions in a phantom: (a) MR image and (b) intensity plot along the FOV direction (left-right).

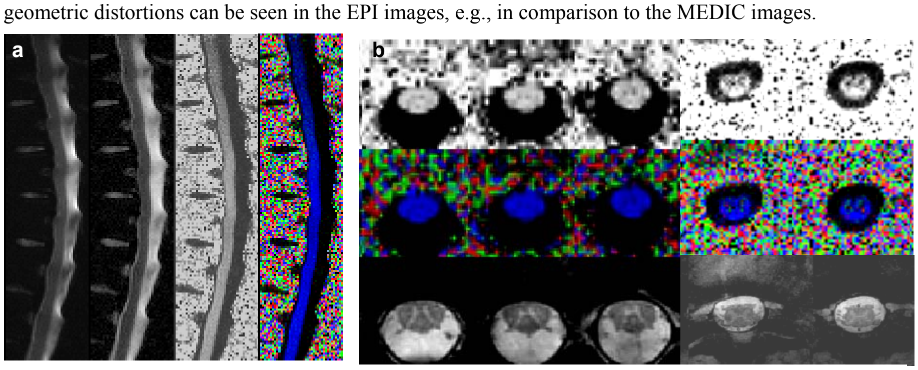

Fig. 4: (a) Sagittal and (b) transverse sections of the thoracic and lumbar spinal cord. (a) non-diffusion-weighted image, ADC map, grey-scale and color-coded FA maps (left → right), respectively, (b) gray-scale and color-coded FA maps and T2*-weighted (MEDIC) images (top → bottom), respectively.

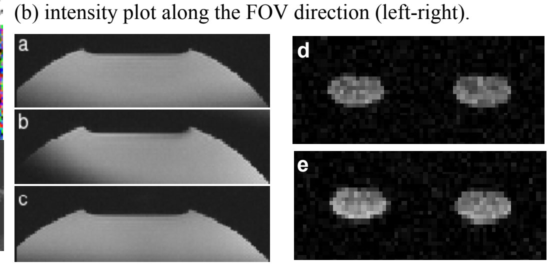

Fig. 3: Inner-FOV images acquired with (a,b,d) the standard and (c,e) the tilted geometry in a phantom (a) without and (b,c) with an offset of the shim current in x (left-right) and (d,e) the cervical spinal cord. The gray scaling in (d) and (e) is identical.

Discussion and Conclusion

For large objects and/or many slices, tilting the excitation plane can shorten the 2DRF pulse durations considerably. This shortens the echo time which increases the SNR correspondingly, but also reduces the 2DRF's sensitivity to off-resonance effects, e.g. profile distortions in the presence of magnetic field inhomogeneities. Furthermore, the optimal 2DRF excitation in such a setup is independent of the object size and the number of slices in the stack. Thus, the presented approach may help to improve and facilitate the applicability of inner-FOV EPI based on 2DRF excitations.

References

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[4] Saritas EU *et al.*, Magn. Reson. Med. **60**, 1099 (2008) [5] Finsterbusch J, J. Magn. Reson. Imaging **29**, 987 (2009)