T2*-Weighted Inner-Field-of-View Echo-Planar Imaging of the Spinal Cord

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Introduction

BOLD-based functional MRI of the spinal cord has gained increasing interest in the past few years (e.g. [1]). Most applications were based on conventional echo-planar imaging (EPI) [2] but this approach faces limitations regarding image quality and the spinal cord sections that can be investigated. First, for the high spatial resolution required to identify grey matter reliably, significant geometric distortions can occur due to the field inhomogeneities in and around the spinal cord. Second, while the cervical spinal cord can be acquired with a reasonable field-of-view (FOV), lower sections are diffcult to measure because a much larger FOV is required to avoid aliasing artifacts which prolongs the echo time, reduces the signal-to-noise ratio, and increases geometric distortions. In this study, the feasibility of inner-field-of-view EPI based on 2D-selective RF (2DRF) excitations [3,4] to acquire T2*-weighted images [5] of the human spinal cord in vivo is demonstrated. Compared to conventional, slice-selective EPI, geometric distortions are reduced because of the shorter echo train and lower sections of the spinal cord can be covered without the need for excessive FOVs. Like for conventional EPI, slice-specific z-shimming [6] can be used to minimize signal dropouts related to through-slice dephasing.

Methods

Figure 1 shows the basic pulse sequences used in the present study. The 2DRF excitation is based on a fly-back blipped-planar trajectory with the line and blip direction coinciding with the imaging frequency- and phase-encoding direction, respectively. The field-of-excitation (FOE) was chosen such that the side excitations could be positioned outside of the object in order to avoid unwanted signal contributions. A slice-specific "z-shim" gradient is applied in the slice direction to compensate for the dephasing effect of through-slice field inhomogeneities [6].

Measurements were performed on a 3T whole-body MR system (Magnetom TIM Trio, Siemens Healthcare) using a 12-channel head (phantom) and the standard neck and spine coils (volunteers). Healthy volunteers were investigated after their informed consent was obtained. The resolution of the 2DRF trajectory was 2.5×10 mm^2 (line×blip)with FOEs between 170 and 230 mm yielding 2DRF pulse durations between 22.8 and 31.0 ms. The 2DRF envelope was calculated for a rectangular excitation profile (5×30 mm^2) using the low-flip-angle approximation [4]. Echo-planar images were acquired with an in-plane resolution of 1.0×1.0 mm^2 yielding echo times between 30 and 34 ms. The optimum z-shim gradient moment for each slice was determined from a reference measurement stepping through different moments for each slice.

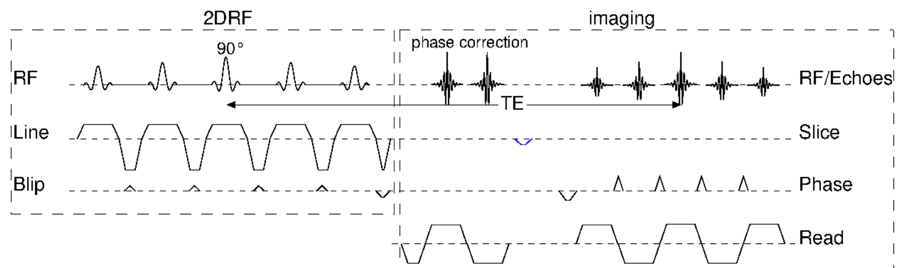

Fig. 1: Basic pulse sequence used in the present study. After the echoes acquired for phase correction, a slice-specific gradient pulse is applied in the slice direction (blue) that is adjusted for each slice to compensate through-slice dephasing effects caused by field inhomogeneities.

Results and Discussion

Figure 2 demonstrates that the distortions in a phantom and in vivo are reduced with inner-FOVs, e.g., the spinal cord sections appear less compressed. In Fig. 3, the feasibility to combine inner-FOVs using 2DRF excitations with a slice-specific z-shim gradient moment is shown. With z-shimming, through-slice signal dephasing in slices close to susceptibility differences can be minimized and significant signal intensities can be regained. Example images acquired in the cervical and thoracic spinal cord are presented in Fig. 4 and 5, respectively.

2DRF excitations usually are considerably longer than slice-selective RF excitations but for the small, inner FOVs that are accessible with them the echo train length is much shorter. For the neck region and the chosen in-plane resolution, very similar echo times can be achieved for inner FOVs with 2DRF excitations and conventional imaging (30 ms vs. 31 ms); however, for lower spinal cord sections, much larger FOVs are required for conventional imaging to avoid aliasing artefacts leading to echo times of about 50 ms or beyond. In contrast, only a slight increase of the echo time (4 ms) is needed for inner FOVs because the FOE of the 2DRF excitation must be increased accordingly. Thus, shorter echo times and a higher signal- and contrast-to-noise ratio are achievable with inner FOVs making it an interesting approach for BOLD-based functional imaging of the spinal cord.

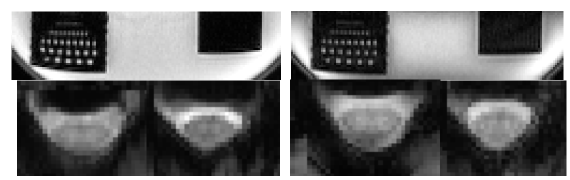

Fig. 2: Phantom (upper) and in vivo (lower) images with conventional slice-selective RF excitations (left) and inner-FOV imaging based on 2DRF excitations (right). Because of the shorter echo train, geometric distortions are less pronounced for the inner-FOVs, e.g., yielding less "compressed" spinal cord cross-sections.

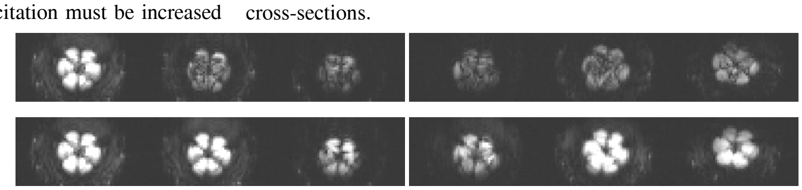

Fig. 3: Six slices of a cucumber cut into two pieces to obtain an air-filled gap in-between that simulates the effect of susceptibility differences and through-slice dephasing acquired without (upper) and with (lower) slice-specific z-shim gradient pulse. With z-shimming significant signal intensities can be regained.

References

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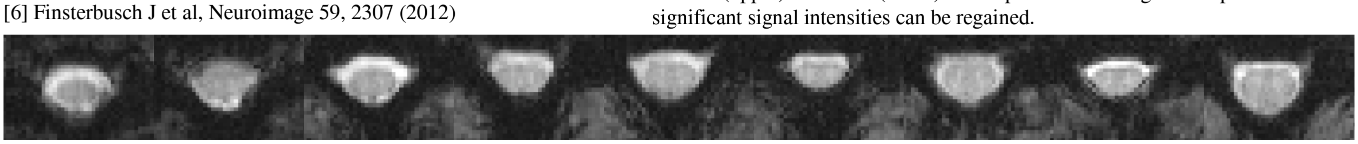

Fig. 4: Nine slices in the cervical spinal cord demonstrating the image quality that can be achieved with inner-FOV imaging using 2DRF excitations and slice-specific z-shim gradient pulses (1.0×1.0×5.0 mm^3, one average).

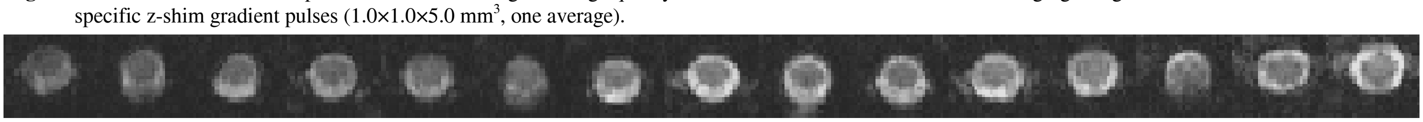

Fig. 5: 15 slices in the thoracic spinal cord (Th3-Th7) acquired with a resolution of 1.0×1.0×5.0 mm^3 (one average).