Multi-contrast inversion-recovery EPI (MI-EPI) functional MRI at 7 T

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TARGET AUDIENCE Researchers interested in the underpinnings of fMRI contrast mechanisms due to tissue composition or in physiological noise sources.

PURPOSE Conventional fMRI using echo planar imaging (EPI) observes the change of a combination of tissue MRI parameters in response to neurovascular fluctuations. While the BOLD contrast is primarily driven by T_2^* changes, T_1 effects also cause BOLD signal changes in the form of inflow effects, which are also driven in part by local neuronal activation. Furthermore, in the presence of tissue and/or head motion, voxels that sample two tissue classes with differing background signal intensities will exhibit signal fluctuations during the time series that depend on the local tissue contrast, which can be viewed as a dynamic partial volume effect. To be able to examine functional changes in different tissue contrast conditions and in different components of the BOLD response, here we introduce a fast multiple-contrast method for functional imaging based on inversion recovery (IR) EPI. Conventional IR EPI scans the imaging volume at a single inversion time (TI) at a time, thus scaling the data acquisition time linearly with the number of contrasts. Furthermore, conventional IR EPI scans each slice at a constant TI, which is very time consuming even when only one contrast is required (e.g., as in VASO) since only one slice or a small number of slices, with slice-selective inversion pulses, can be scanned at a single sequence repetition. To increase efficiency, a slab-selective inversion can be used with a slice ordering permutation scheme that allows all slices to achieve several TI times in a short period[1].

METHODS A conventional IR-enabled gradient echo (GRE) EPI sequence was modified in order to scan every slice several times after each inversion pulse, although at unequal TIs, and this sequence was tested at 7 Tesla. The inversion pulse from the source sequence was changed to an adiabatic frequency offset compensated inversion (FOCI) pulse[2] to gain transmit efficiency, FLEET autocalibration signal acquisition was adopted to improve GRAPPA quality[3], and finally, a novel turbo-IR method was used in which multiple readouts follow each inversion pulse. A low flip angle (α) selected for readouts, of which we here had twenty per slice, allowed for modeling the data using an exponential growth of longitudinal magnetization and getting a qualitatively correct T_1 map and a composite T_2^*/ proton density (ρ) weighted image. We also want to note that the short repetition time (TR = 3 s) we adopted for functional imaging does not enable complete IR, which is further hindered by the plurality of (low) α pulses. Both effects are being addressed in quantitative modeling of T_1 recovery. In addition to the calculated parameter maps and the arbitrarily weighted images that can be synthesized from these maps, the method also yields the collection of source images acquired throughout the IR with variable contrast properties.

Four volunteers were scanned on a 7-T whole body MRI scanner (Siemens Healthcare, Erlangen, Germany) using a 32-channel receive array with a birdcage head transmit coil after written informed consent. The visual stimulation presented to the subjects comprised a black-and-white spatial noise pattern counterphase flickering at 8 Hz presented in 15-s blocks alternating with 21-s control blocks showing a neutral gray background. Each measurement contained five stimulation blocks, the measurements started and ended in control blocks. Three runs were acquired for each subject. We acquired 3 slices of multi-contrast IR EPI data at (1.5 mm)^3 isotropic resolution with TR/TE/α/BW/matrix/R = 3 s/21 ms/20°/1628 s^{-1}/128 × 128/4 and 150-ms temporal sampling period for the IR data.

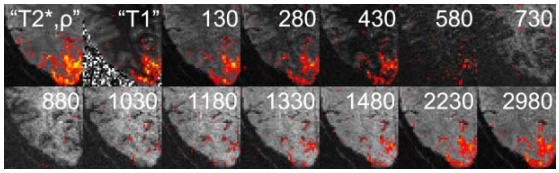

Fig.1. Multiple contrast images of one of 3 slices acquired during the IR fMRI scan in 3 s. From top-left, computed: T_2^*/ρ, T_1, measured: TI in ms as indicated.

RESULTS Fig. 1 illustrates the range of different images with thresholded statistical activation maps overlaid (double-Γ HRF, cluster $p < 0.05$, no smoothing; FSL/FEAT). It should be noted that the BOLD-like T_2^*/ρ-weighted image is effectively decomposed into different component images (or is actually composed of them), thus it is natural that the component images have only a fraction of the SNR of an image acquired in the absence of an inversion preparation. In Fig. 2, the time course at a single voxel predominantly sampling gray matter is shown for the contrast conditions of Fig. 1. It is evident that contrasts c6–c9 in Fig. 2 lack the stimulus response almost entirely.

DISCUSSION Despite the incomplete T_1 recovery, the sequence was able to capture the qualitative characteristics of the T_1 relaxation at a standard functional temporal sampling resolution of 3 s, including the nulling of specific tissue classes at certain TIs: white matter in c3, grey matter (GM) in c4–c5, and cerebrospinal fluid (CSF) in c6–c7. Interestingly, we found significant task-related increases in T_1 values, compatible with increases of blood volume in the parenchyma and/or residual inflow effects. The functional contrast changed remarkably at different contrast conditions, as exemplified by the single-voxel plots in Fig. 2, and did not just reflect the local signal or SNR levels. E.g., the baseline signal in Fig. 2(C) is similar for c3/c9, c4/c7, and c5/c6, yet only c3–c5 show clear activation signals. In addition to actual tissue properties, some of the contrast images show tissue boundaries as signal cancelation, specifically c5 for GM– CSF interface. The CSF-attenuated contrasts c6–c7 interestingly show poorer activation than the GM-suppressed contrasts. Dynamic partial volume effect could explain the functional contrast in c5 even in absence of any substantial GM signal. The lack of functional contrast in CSF suppressed contrasts where GM is adequately bright is currently under investigation, also see Fig. 2C. To increase the number of slices to achieve broader coverage, data may be sampled less frequently, since fewer data points sampling the IR curve would suffice to provide a stable fit. The sequence is also compatible with the Simultaneous Multi-Slice imaging.

CONCLUSION The multitude of acquired and modeled contrasts can help understanding the different contributions to the fMRI BOLD signal and some mechanical changes occurring in tissue during brain activation. The method has the inherent ability to null several tissue components within one acquisition, almost simultaneously, which can simplify experimental design for studies requiring such information. It is possible to choose the scan parameters carefully to select very specific tissues for suppressing, and use slice acquisition order jittering built into the sequence to e.g. null both intravascular blood and CSF in a single run, noting that the effective T_1 of CSF is not unique for the whole brain volume, therefore the ability to null several different T_1 values in one acquisition is a key feature to enable more effective nulling.

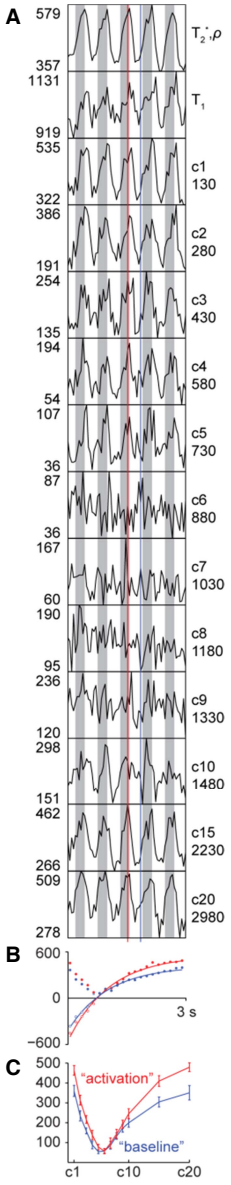

Fig. 2. A: Signal–time plots for several contrasts in a single voxel. Min, max, and TI values are shown, gray shading illustrates the stimulation periods. B: Data and IR model fits for two time points indicated by color. C: Baseline (3 time points preceding each simulation block) and activation (3 tp before control) signal for measured data, mean ± std.

REFERENCES 1. Clare S & Jezzard P. 2001 MRM;45:630–4. 2. Hurley AC, et al. 2010 MRM;63:51–58. 3. Polimeni JR, et al. 2013 ISMRM;21:2646.

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