Investigation into the benefits of 3D-EPI for high-resolution fMRI at 7T

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Introduction

Since the discovery of the BOLD effect, 2D gradient-echo echo planar imaging (EPI) has been the workhorse of functional MRI, largely because of its high sampling speed and excellent sensitivity to signal changes related to brain activation, at sufficient spatial resolution and brain coverage. However, physiological noise considerations make it beneficial to sample at a higher resolution than required, in order to profit from increased temporal SNR when smoothing, rather than directly acquiring at the lower desired resolution [1]. With high field scanners and multi-channel coil arrays becoming more readily available signal can traded off for resolution, and acquisition time can be shortened using parallel acquisition [3], which is more efficient at higher fields [2] and can be applied in any phase encoding direction. To fully profit from parallel imaging we propose a 2D-accelerated 3D EPI acquisition because the common 2D EPI scheme uses one phase encoding direction only.

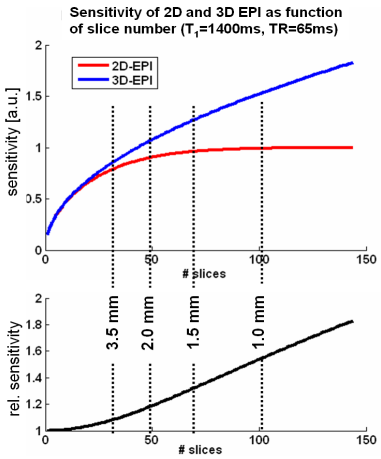

Fig 1: *Theoretical sensitivity 2D and 3D EPI as function of slice number (top), and their relative performance (bottom).*

7 Tesla 2.5mm isotr.	2D-EPI TR=2.26s	3D-EPI TR=1.38 s	Sensitivity relative to 2D
occ	92	74	103%
front	59	53	115%
cereb	39	50	164%
avg	**63**	**59**	**127%**

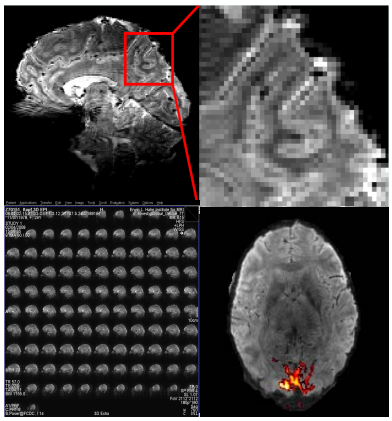

Fig 2 3D-EPI at 7T with 1mm^3. Top: with 8 channel coil, 2x3 acceleration, TR=4.2s. Bottom: 32 channel coil, 3x3 acceleration**, TR=3s.**

Theory

In 2D EPI a slice selective excitation is performed and the signal subsequently acquired under an oscillating read and blipped phase encode gradient, so as to acquire the data for a full 2D image in a single-shot. After repetition time $TR \cdot N_{slc}$ all N_{slc} slices have been acquired, and the process is repeated. In 3D EPI a thick slab of tissue is repeatedly excited, and a k_x-k_y plane of k-space acquired each time with a different k_z increment. Slice selection is hence replaced by a secondary phase encoding direction, and the complete 3D image is obtained after time $TR \cdot N_{slc}$. In principle, there are two important considerations in the choice of approach: **1)** The implications of TR and N_{slc} on achievable signal for a given T_1 relaxation time: The magnetization available for image formation is given by Eq [1] and maximal when α is the Ernst angle, Eq [2]. For full k-space sampling, $TR_{3D}=TR_{2D}/N_{slc}$ is and yields a lower steady state signal; however 3D has a factor $\sqrt{N_{slc}}$ sensitivity advantage as N_{slc} k-space points contribute to each image pixel. The potential benefit of 3D acquisition thus increases with slice number. This is simulated in Fig 1 which shows sensitivity as a function of slice number (using T_1=1400ms and TR=65ms and constant conditions). Dotted lines indicate the slice numbers required for different spatial resolutions and 100mm brain coverage in the slice direction. **2)** The opportunities for using parallel imaging with multiple receive coils, which permits k-space undersampling in the phase encode direction. In contrast to 2D schemes, 3D allows reducing the volume acquisition time by the acceleration factor in 3D encoding direction. When aiming for very high spatial resolution the volume acquisition time becomes unreasonably long for 2D EPI.

$$M = M_0 \sin\alpha \left(1 - \exp\left(-\frac{TR}{T_1}\right)\right) / \left(1 - \cos\alpha \cdot \exp\left(-\frac{TR}{T_1}\right)\right)$$

$$\alpha_{Ernst} = \arccos\left(\exp\left(-\frac{TR}{T_1}\right)\right)$$

Methods

A 3D EPI sequence with full partial Fourier and parallel imaging capability and flexible z-encoding order was implemented for the Siemens Magnetom scanner family (Siemens, Erlangen, Germany). Measurements were made at 7T in accordance with local ethics regulations. Temporal SNR was determined for several brain regions from 50 volumes of resting state data acquired with typical 2D and 3D protocols, using an 8-channel Tx/Rx head array (RAPID Biomedical, Germany): resolution 2.5mm isotropic, TE=23 ms, $TR_{volume\ 2D}$=2.26s, $TR_{volume\ 3D}$=1.38s. Additional data sets were acquired at 1mm resolution using the 8-channel coil (sagittal orientation) and using the MGH 32 channel coil [4] (axial orientation). A 3 minutes functional experiment using 20/20s on/off visual stimulation was performed (matrix 180x180x104, TE=23ms, 3x3acceleration, BW=2000Hz/ px, $TR_{volume\ 3D}$=3s).

Results

The table shows the sensitivity of 2D and 3D EPI at 7T with 2.5mm resolution, for ROIs in occipital / frontal cortex and cerebellum. On average, the sensitivity as calculated as SNR/√time is 27% higher with 3D EPI. At the same time, the much shorter TR (1.4s vs. 2.3s) permits more adequate sampling of the BOLD response. Fig 2 (top) shows one slice of 7T data acquired at 1mm^3 with the 8ch coil (a=2x3, TR=4.2s). Note the excellent image quality and gray-white matter contrast. The bottom left shows the 112-slice mosaic image from which the slice was taken In the bottom right panel an activation map (1mm spatial smoothing) of the 1mm^3 functional data, overlaid on the average image of one slice. The activation (maximum z-score=11.6) nicely follows the sulci of the occipital cortex.

Discussion and Conclusion

Using 3D EPI at 7T und 2D parallel acceleration along the z-direction allows high-resolution (1mm isotropic) protocols with a typical TR of 3 seconds. 3D EPI is inherently a multi-shot technique that is more prone to physiological noise (breathing, cardiac pulsation, motion) and at typical spatial resolution for fMRI, 2D sampling is preferable as here the detrimental effect is much weaker. However, the desire for and usefulness of high spatial resolution particularly at ≥ 7T severely limits 2D EPI: First, SAR safety limits are rapidly reached even with GE-EPI due to the large Ernst flip angles required for long TR, whereas 3D EPI requires very low flip angles. Second, the effective spatial resolution may be compromised by the imperfect slice profile in 2D; this is not an issue with 3D phase encoding, where only a few edge slices may have to be discarded. Third, the gradient amplitude required for thin slices (<2mm) reaches system limits and/or causes nerve stimulation unless the pulse duration is increased; also in this respect 3D EPI is less limited by technical and practical constraints than 2D EPI. It should hence be the method of choice for high field fMRI.

References

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