High resolution 3D fMRI of human visual cortex with elimination of large venous vessels

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Introduction

A continuing goal in fMRI is the attainment of higher spatial specificity. If a high enough SNR is present then this may be considerably better than the width of the hemodynamic point spread function. In standard gradient-echo fMRI the specificity may be compromised by the inability to eliminate signal contributions from larger venous vessels [1-3], and by the distortion inherent to the EPI technique, particularly at high field strength. In this abstract we advocate an alternative approach as follows: Parallel imaging is utilised to reduce the acquisition time for a 3D-FLASH acquisition to a duration compatible with a block design fMRI experiment thus eliminating distortion and permitting a very high spatial resolution; an 8 channel occipital coil is used to maximise sensitivity; large venous vessels are then removed post acquisition.

Methods

Five healthy subjects (2 female, age range 25-30 years) were studied after informed consent was given, according to institutional guidelines of the local ethics committee. All MR imaging experiments were performed on a 3 Tesla Siemens Magnetom Trio (Siemens, Erlangen, Germany) using a custom built eight channel occipital cortex array (Stark Contrast, MRI Coils, Erlangen, Germany). Functional data sets were obtained using a 3D flow-compensated gradient echo FLASH sequence as implemented in the susceptibility weighted imaging (SWI) package (work in progress package available from the manufacturer). The 3D slab was obtained using the following imaging parameters: repetition time (TR) = 35 ms, echo time (TE) = 28 ms, flip angle (α) = 15°, readout bandwidth (BW) = 100 Hz/pixel, rectangular field-of-view (FOV) = 144x108 mm2, slab thickness = 30 mm. GRAPPA [4] was used as the partial parallel imaging mode with an acceleration factor of 2 using 24 reference lines. Spatial resolutions of 1.13x1.13x0.94 mm3 (1.2 µl) and isotropic 0.75x0.75x0.75 mm3 (0.42 µl) were used, respectively. The corresponding geometric parameters were a 128 x 96 matrix and 32 slices of 0.94 mm thickness, and a 192x144 matrix and 40 slices of 0.75 mm thickness, respectively, resulting in an acquisition time of 67 and 118 seconds per 3D volume. Anatomical data sets were obtained using MP-RAGE with 0.75 mm resolution.

The stimulus paradigm consisted of a block showing a black screen with a fixation cross versus a block with a flickering checkerboard (5 Hz, i.e. 200 ms for the full cycle). The block duration was 67 s per block for the lower resolution and was repeated 14 times (7 fixation and 7 checkerboard blocks), and 118 s for the higher resolution per block (repeated 8 times, 4 fixation and 4 checkerboard blocks) resulting in a total acquisition time of 15 minutes 40 seconds per run.

The data were analyzed using a general linear model as implemented in FEAT (FSL, FMRIB, Oxford, UK) using no smoothing and modarate smoothing (1.0 mm and 1.5 mm), and MCFLIRT was used for motion correction. In a second step veins (including sagittal sinus) - easily visible as dark spots or lines (see Fig. 1a) - were segmented manually (MRIcro, Chris Rorden, www.sph.sc.edu/comd/rorden/mricro.html) on the reference volume used for motion correction. This ROI was used a venous mask (see Fig. 1b) to eliminate the respective voxels in the motion corrected functional data sets. These masked fMRI data sets were processed in the same way as the original ones. 3D statistical maps (unthresholded z-maps) were calculated and are displayed using the 3D tool of MRIcro.

Results

The maximum displacement due to motion found in all data sets was 0.28 mm over the whole run as estimated from the results of the motion correction algorithm. Averaging the maximum displacements per run over all subjects resulted in 0.17±0.08mm for the 1mm voxel size and 0.19±0.07mm for the 0.75 mm voxel size, respectively. In Fig. 2 unthresholded 3D z-maps before (a) and after application of venous mask (b) are shown. In (a) long tubular structures can clearly be seen, which are not present any more after application of the venous mask. Fig. 3 displays an overlay of a thresholded z-map on the corresponding anatomical slice.

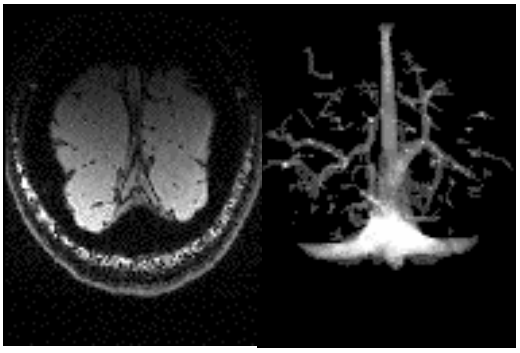

Fig.1: (a) Single slice out of the 3D data set with venous structures clearly visible. (b) Segmented veins including sagittal sinus used as venous mask.

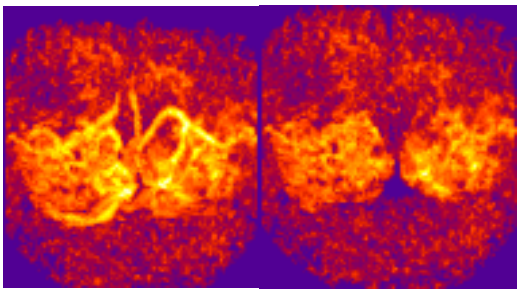

Fig.2: (a) unthresholded 3D z-map of visual activation. (b) unthresholded 3D z-map of visual activation after masking functional data with venous mask (both smoothed with 1.5 mm kernel).

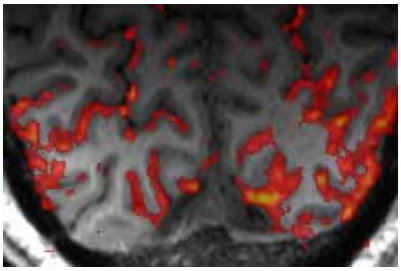

Fig.3: Thresholded (Z>1.7) and smoothed z-map of a single slice overlayed on a MP-RAGE image with gray/white matter contrast. Note the confinement of activation to gray matter areas.

Discussion and Conclusion

The proposed method delivers distortion free, very high spatial resolution isotropic 3D images enabling direct overlay on anatomical images. Our results clearly show that large vein effects can be a major source of "activation". While various indirect methods have been used to get rid of at least a part (e.g. intravascular effects by diffusion weighting) of large vessel effects, our results show that a direct identification and elimination of large veins is possible and clearly enhances the quality of activation maps. Compared to multi-slice measurements this true 3D method facilitates motion correction as no spin history effects have to be taken into account. This method might be used to identify small functional units, such as the ocular dominance columns where a comparison with anatomical information is necessary [5]. Currently, the drawback is the relatively long measurement time necessary, but a combination of higher field strengths and higher acceleration factors in parallel imaging should enable a considerable reduction in measurement time.

References

[1] Lai et al, MRM 1993 [2] Segebarth et al, Neuroreport 1994 [3] Hoogenraad et al, MRM 2001 [4] Griswold et al, MRM 2002 [5] Cheng et al, Neuron 2001

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