Special Considerations/ Technical Limitations of Blood-Oxygen-Level-Dependent Functional Magnetic Resonance Imaging

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KEYWORDS

• Blood-oxygen-level-dependent • Functional magnetic resonance imaging • Special considerations • Technical limitations

KEY POINTS

- Blood-oxygen-level-dependent (BOLD) functional magnetic resonance imaging (fMRI) has the potential to become a more universal standard of care in presurgical planning for localization of eloquent cortex at risk during surgical resection.
- BOLD imaging is affected by a series of technical issues limiting the widespread clinical use of BOLD fMRI.
- Extensive and standardized quality control tools need to be established for appropriate interpretation of both clinical and research fMRI studies.
- Newly developed methods can overcome current BOLD imaging issues and enhance future research and clinical application of BOLD fMRI.

INTRODUCTION

Over the last 20 years, blood-oxygen-level-dependent (BOLD) functional magnetic resonance imaging (fMRI) has been effectively used for clinical presurgical mapping.[1,2] However, there are important technical limitations and special considerations that one must be aware of to avoid pitfalls in both clinical and research applications of BOLD fMRI.

IMAGE ACQUISITION

Susceptibility Artifacts

Most clinical and research fMRI studies are performed by using a two-dimensional T2*-weighted gradient recalled echo (GRE) sequence with echo planar imaging (EPI) readout.[3] The rationale behind this choice is the high sensitivity of this pulse sequence to BOLD-related susceptibility changes and its ability to scan the whole brain with adequate spatial (2–3 mm) and temporal (2–3 seconds) resolution to monitor brain activation over time.[4] However, T2* GRE EPI shows high sensitivity as well to intravoxel dephasing caused by macroscopic magnetic field gradients generated by the difference in magnetic susceptibility of multiple tissues contained in 1 voxel.[5] The different magnetic fields experienced by the spins make them precess at different frequencies and,

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Neuroimag Clin N Am 24 (2014) 705–715
http://dx.doi.org/10.1016/j.nic.2014.07.006

over time, dephasing leads to signal loss. This effect is strong and results in signal loss in regions of the brain characterized by strong susceptibility differences at the junctions between air and tissues such as the orbitofrontal cortex (from the paranasal sinuses) or the medial temporal and the inferior temporal lobes (from the petrous apices and mastoid air complexes),[6] as shown in **Fig. 1**. These regions are important in visual and cognitive processing, including language and memory function.[7–9] The effect of this signal loss is a reduced sensitivity to brain activation in these regions, which may not be recognized when the statistical maps are overlaid on less distorted high-resolution T1-weighted anatomic images. Such susceptibility-related signal loss may result in regional false-negative activation on BOLD presurgical mapping studies. The amount of signal loss has been shown to be dependent on the image orientation, echo time (TE), and spatial resolution.[10] Because the magnetic field gradients are generated along the slice selection, phase encoding, and readout directions, in-plane dispersion is experienced as well as through-plane dispersion of the voxel magnetization.[11] Spatial resolution also counts, because the larger the voxel size, the larger the difference in Larmor frequencies among the spins contained in the voxel and, in turn, the faster the signal dispersion.[12] Reducing voxel size reduces the effects of susceptibility artifacts but at the cost of temporal resolution or reduced brain coverage. The spin dephasing increases along time; therefore, the strength of signal loss depends also on the TE.[13] In principle, one could reduce the TE, but this is at the expense of reducing BOLD sensitivity in other regions of the brain less affected by susceptibility artifacts. In current clinical studies, a tradeoff between spatial/temporal resolution and BOLD sensitivity needs to be achieved.

In clinical functional imaging, additional potential sources of susceptibility artifacts include vascular clips, stent grafts, or craniotomy hardware related to previous surgery. These devices can induce strong macroscopic field gradients and generate dramatic signal loss, as shown in **Fig. 2** for a patient who underwent presurgical

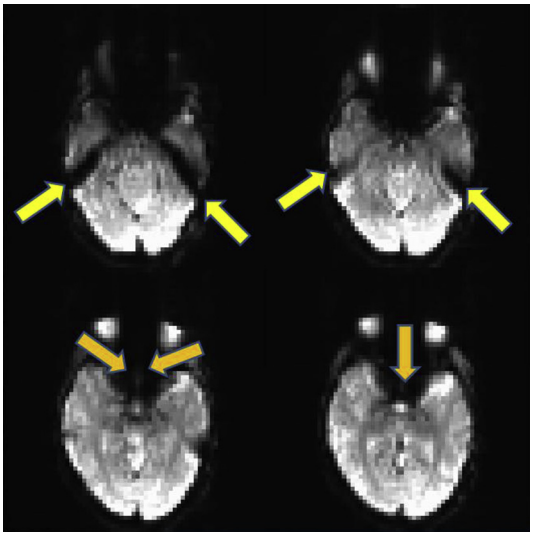

Fig. 1. Example of areas commonly affected by susceptibility artifacts in single-shot T2*-weighted EPI images (repetition time/echo time = 2000/30 milliseconds) acquired in a 37-year-old patient with grade II fibrillary astrocytoma referred for fMRI presurgical mapping at our institution. (*Top row*) Right and left inferior temporal lobes (*yellow arrows*) obscured by artifact from the petrous apices; (*bottom row*) susceptibility artifact from the sphenoid sinus, which may affect visualization of orbitofrontal activation (*orange arrows*).

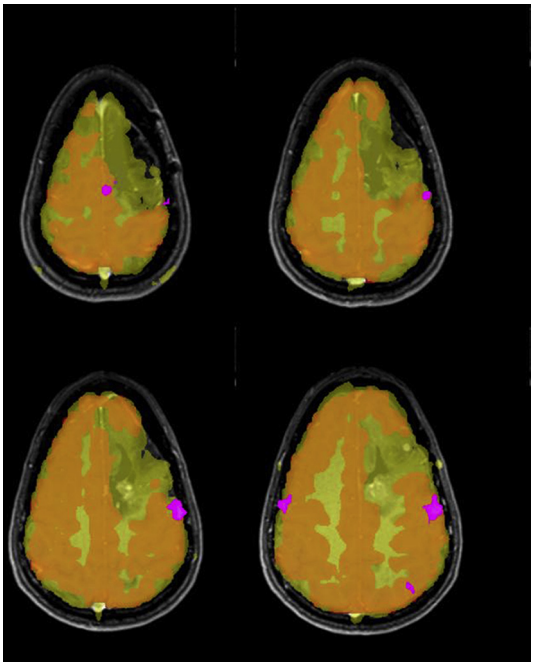

Fig. 2. Susceptibility artifact related to a left frontoparietal craniotomy that obscures activation in the dorsolateral prefrontal cortex/middle frontal gyrus on a silent word-generation verbal fluency task. Notice that the BH CVR map that has been overlaid on the raw EPI image shows loss of regional vascular reactivity, but this is caused by surgical hardware–related susceptibility artifact rather than neurovascular uncoupling (NVU). Note also that the degree of susceptibility-related anatomic distortion is greater on the EPI images than on the underlay of postcontrast T1 three-dimensional magnetization prepared rapid acquisition gradient echo images.