HIGH RESOLUTION VENOGRAPHY AT 3 TESLA

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Abstract: The purpose of this study was to evaluate Blood Oxygen Level Dependent (BOLD) MRI contrast imaging as a potential usable clinical tool enabling high resolution brain venography at 3 Tesla. BOLD MRI exploiting the intrinsic magnetic properties and the T2* effects of blood has been found capable of highlighting veins vasculature with respect to brain parenchyma and arteries. Furthermore this technique does not require injection of contrast agent given the endogenous nature of the needed contrast or utilization of specially designed radiofrequency pulse sequences in order to visualize the veins. The starting imaging data were obtained by means of T2* weighted sequence with optimized parameter acquisition settings. A minimum Intensity Projection (mIP) algorithm on the magnitude images (0,5 x 0,5 mm^2 in plane resolution) was applied obtaining a 2D image of the related venous structures. The results indicate that at 3T the contrast between veins and surrounding structures for T2* weighted images is sufficient to reconstruct with great detail the brain venous map. Also in terms of duration the results are promising, since the average total time requested for the scan and process was about 20 minutes, that sounds to be acceptable for future clinical use of this method.

Introduction

A good knolewdge of the geometrical parameters that characterize the cerebral vascular and microvascular system is a fundamental base for the study of brain functionality and to detect the tissues and the morphology of the brain itself. This is extremely important for clinical purposes, as it can be a very helpful tool for the characterization of vein pathologies and malformations in patients affected by hypertension, who could suffer from cerebral ischemia, thrombosis and so on. (e.g. [1]).

The brain vascular architecture is structured in a very complicated and intricate pattern [2]. The smaller vessel sizes range in diameter from several microns for capillaries to several tens of microns for venules, precapillary sphincters and arterioles to several 100 μm for small pial arteries and veins [3].

The mean flow velocity changes are inversely proportional to the cross sectional area. In pial vessels, the mean flow velocity ranges from several mm/s to a few cm/s, whereas in the capillary it is only on the order of 0.5 to 3 mm/s [4]. However, due to the usually slow flow of blood in small vessels, they are difficult to detect with conventional magnetic resonance methods, such as time-of-flight (TOF) [5] and phase contrast angiography (PCA) [6].

Both techniques require the presence of moving blood: the former is sensitive to rapidly inflowing spins, whereas the latter represents a proton velocity map. Time-of-flight (TOF) angiography is based on the phenomenon of flow-related enhancement of spins entering into an imaging slice. As a result of being unsaturated, these spins give more signal than surrounding stationary spins. With 2-D TOF, multiple thin imaging slices are acquired with a flow-compensated gradient-echo sequence. These images can be combined by using a reconstruction technique such as maximum intensity projection (MIP), to obtain a 3-D image of the vessels analogous to conventional angiography, since the vessels will appear bright compared with the surrounding tissue. Spins that are moving in the same direction as a magnetic field gradient develop a phase shift that is proportional to the velocity of the spins. This is the basis of phase-contrast angiography. In the simplest phase-contrast pulse sequence, bipolar gradients (two gradients with equal magnitude but opposite direction) are used to encode the spin velocity. Stationary spins undergo no net change in phase after the two gradients are applied. Moving spins will experience a different magnitude of the second gradient compared to the first because of its different spatial position. This results in a net phase shift. This information can be used directly to determine the velocity of the spins. Alternatively, the image can be subtracted from one acquired without the velocity encoding gradients to obtain an angiogram.

Most of recent advances have been made in the development of techniques assessing the arterial system; instead, high resolution magnetic resonance techniques to image the venous system are not as commonly available.

A new technique, BOLD MRI Venography has been developed in the recent years [7,8] which allows to visualize small veins with a diameter smaller than the pixel size. The contrast obtained in these images is directly related to the magnetic properties of the blood, which are very sensitive to its oxygenation state. Particularly the iron in the blood is in the hemoglobin, or more specifically in two states, oxyhemoglobin or deoxyhemoglobin. These two states differ in their magnetic properties. Oxyhemoglobin is diamagnetic

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with no unpaired electrons, while the deoxyhemoglobin is paramagnetic. Thus a change or a difference in the oxygenation state of arterial and venous blood manifests itself as a change or difference in the bulk magnetic susceptibility of the blood. This difference is a source of magnetic field variation in the human body generating in turn a variation of the relaxation time T2 and T2* between tissues having different magnetic susceptibility (e.g. fully oxygenated and fully deoxygenated red blood cells). This effect in a MR gradient echo imaging sequence manifests itself in dephasing of the signal coming from the venous blood vessel because of its longer T2 and T2* respect to arteries and parechyma. The amount of dephasing is given by [9]:

$$\varphi(\mathbf{r}, TE) = -\gamma \, \Delta B(\mathbf{r}) \, TE \qquad (1)$$

where $\Delta B(\mathbf{r})$ is the magnetic field variation between regions with different susceptibility, γ is the gyromagnetic ratio and TE is the sequence Time of Echo. The aim of this study was to investigate the visualization of venous vessels in the normal human brain at a field strenght of 3 Tesla, using high resolution MR imaging. The biological tissue signal-to-noise-ratio (SNR), infact, scales with the magnetic field strenght of the instrument magnet, to at least the first power. Higher SNR, in turn, translates into better temporal resolution or better spatial resolution or some combinations of two.

Materials and Methods

A healthy volunteer partecipated this study (age 25 years, male). Informed consent was obtained following the guidelines of the institutional review board. The images were acquired at the Signa GENERAL ELECTRIC scanner owned by the *Istituto Ricovero e Cura a Carattere Scientifico* "Casa Sollievo della Sofferenza" in S. Giovanni Rotondo (FG) Italy. The main magnet of this scanner produces a 3 Tesla axial magnetic field.

The pulse sequence used in this study was the 3D, spoiled gradient-echo (SPGR), flow compensated, T2* weighted single echo images. The sequence parameters were: TR = 30 ms, TE = 25 ms, FA = 25°, BW = 31,2. 60 slabs were acquired in the axial plane, 1 mm thick, with a field of view of 240 x 240 mm^2. The frequency and phase encoding steps were respectively 512 and 256. In the image reconstruction process the values along the phase encoding direction were interpolated giving a 512 x 512 image matrix size. The typical voxel size was, thus, 0,25 mm^3. An autoshimming procedure was setup before acquiring the data in order to avoid field inhomogeneity effects on the images.

The obtained images were displayed and processed by using the AFNI software packages [11]. In the processing the first step was the segmentation of intracranial regions for each image (Figure 1). This was performed by means of one automatic brain segmentation algorithm provided in the AFNI suite.

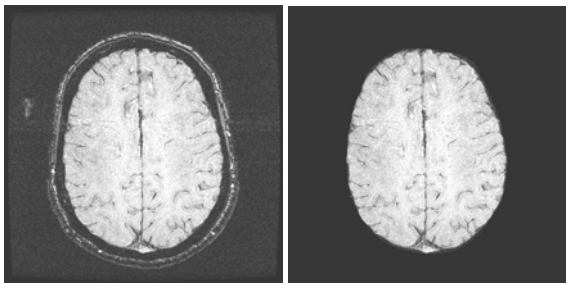

Figure 1: Image before (left) and after (right) brain segmentation

In this way the parts representing the braincase in the images were eliminated, the information left only described veins and parenchyma.

A mapping of the cerebral venous system was finally obtained applying a minimum Intensity Projection algorithm on a subset of the total number of acquired partitions (17 in our case). An mIP of these images means that a stack of images is viewed from the top down and the minimum signal intensity is recorded for a given pixel in the two dimensional plane of interest.

Results

As shown in Figure 2 a 2D map of the cerebral veins was obtained applying directly on the magnitude image of the central slide the mIP algorithm over 17 mm slab.

The result of this operation was a clearly defined venography of the human brain with a spatial resolution up to 0,5 x 0,5 mm^2 (the pixel size) without injection of any contrast agent.

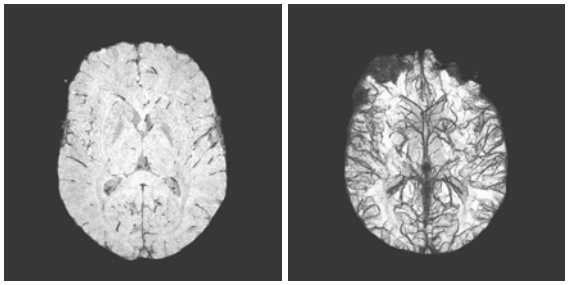

Figure 2: Axial slice (left) and application of mIP on 17 (right)

The mIP was performed on 17 slices (Figure 2) in order to get a good compromise between an acceptable display of the vessels connectivity and the image quality affected by some macroscopic field effects (anterior portion of the head at the air-tissues interfaces).

The results seem to be already valuable and ready to be exploited by physicians for clinical purposes. The total acquisition and processing time was only twenty minutes.

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Discussion

The possibility to reconstruct at high resolution the venous system in vivo may have important consequences for understanding brain function and vascular morphology. The conventional venography methods, such as 2D TOF and 3D PCA techniques do not allow the visualization of venous vessels, especially those with small calibers, without administration of contrast agent [11]. These techniques rely on the detection of blood flow and so their highest resolution is limited to the minimum detectable blood flow. The current value of this parameter is 2 cm/s [12], which limits the minimum vein size to the order of millimeters. Magnetic Resonance BOLD venography is able to provide information about the venous vascular network at the submillimeter scale in an absolutely non invasive way. The basic mechanism of this technique is due to the signal cancellation between venous blood and surrounding tissues. If a voxel contains a small vein with a certain blood volume fraction that corresponds to an MR signal fraction of λ and brain tissue with a signal fraction of 1- λ, the phase difference between the two materials caused by different local Larmor frequencies will increase with TE and will lead to a smaller resulting voxel signal (Figure 3).

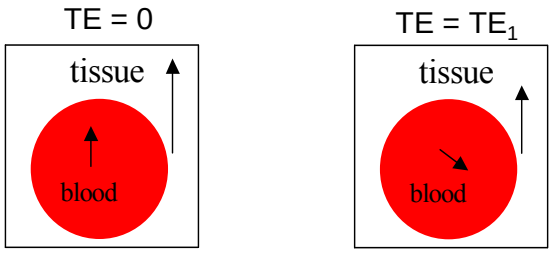

Figure 3: Basic mechanism of BOLD venography: the total signal is the net sum of the signal coming from tissue and blood belonging to the same voxel

At high field strength, imaging makes possible to achieve improved resolution due to higher SNR and higher contrast-to-noise ratio (CNR) [13]. The CNR increase has been demonstrated to be supralinear with the magnetic field strenght from 1.5 to 3 Tesla. This enhances at 3 Tesla the delineation of even smaller vessels compared with lower fields [14]. Performing BOLD MR venography at high fields it is possible to avoid susceptibility artefacts caused by the relative long TE and high field strenght. The signal cancellation between veins and parenchyma depends on the magnetic field variation between these two tisssues with different magnetic susceptibility. Such variation is a linear function of the main magnetic field strenght. It was demonstrated that high resolution venography at 3 T allows shorter echo times with respect to maximum signal cancellation (28 ms versus 50 ms) compared with 1.5 T. This makes it possible also to increase the spatial coverage, or shorten the acquisition time, which in turn, makes potential patient examinations more feasible in assessing venous vasculature in a normal or diseased state with high resolution. At low fields an important step to improve the venous visualization is to exploit the information coming from the phase images. This is usually performed [14] by constructing a phase mask filter image for each partition and multiplying these phase mask filters with the corresponding magnitude images of 3D slab prior the minimum intensity projection. The phase mask filter images area implemented by setting all the positive phase values to 1 and scaling linearly between 0 and 1 all negative phase values. In this way, in the magnitude images containing veins (negative phase values) pixel greyscale values are selectively diminished because they are multiplied with a numerical constant smaller than 1. This procedure is often difficult to implement because of the effects on the magnetic field inhomogenity on reconstructed phase images. Furthermore applying the phase mask filter the processing time could result in about one hour, too long time for clinical application.

When high resolution BOLD venography is performed at 3 Tesla or at higher fields it seems that phase mask filter procedures can be avoided without affecting the quality of reconstructed venograms and thus its clinical usefulness.

Conclusions

We can conclude that BOLD based venography is at present the only method that without the use of any exogenous contrast media is capable of depicting venous vasculature with a high spatial resolution being capable of detecting vessels having diameters smaller than 1 mm. Moreover, as it has been described in the previous paragraphs BOLD venography produces anatomical based images making straightforward and simultaneous the link between high resolution anatomical aspects and blood vasculature: other techniques previously mentioned such as TOF and PC venograms loose most of the references of the anatomic structures.
As retrievable from the literature further improvement could be obtained exploiting the information of the phase images, even if at 3 Tesla we obtain a great detail thanks to the magnitude images only. Future clinical use of this techniques should be supported also by the leading companies producing imaging systems in order to create the condition for a large use in the medical routine.

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