High resolution MRI of the brain at 4.7 Tesla using fast spin echo imaging

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Abstract. Over recent years, high field MR scanners (3 T and above) have become increasingly widespread due to potential advantages such as higher signal-to-noise ratio. However, few examples of high resolution images covering the whole brain in reasonable acquisition times have been published to date and none have used fast spin echo (FSE), a sequence commonly employed for the acquisition of $T_2$ weighted images at 1.5 T. This is mostly due to the increased technical challenges associated with uniform signal generation and the increasingly restrictive constraints of current safety guidelines at high field. We investigated 10 volunteers using an FSE sequence optimized to the 4.7 T environment. This sequence allows the acquisition of 17- and 34-slice data sets with an in-plane resolution of approximately 500 $\mu$m $\times$ 500 $\mu$m and a slice thickness of 2 mm, in 5 min 40 s and 11 min 20 s, respectively. The images appear $T_2$ weighted, although the contrast is due to the combined effects of chosen echo time, magnetization transfer, direct radio frequency saturation and diffusion as well as the $T_1$ and $T_2$ relaxation times of the tissue. The result is an excellent detailed visualization of anatomical structures, demonstrating the great potential of 4.7 T MRI for clinical applications. This paper shows that, with careful optimization of sequence parameters, FSE imaging can be used at high field to generate images with high spatial resolution and uniform contrast across the whole brain within the prescribed power deposition limits.

MRI systems utilizing high field magnets (3 T and above) have been available for a decade [1]. The advantages of higher magnetic field strength include increased MR signal-to-noise ratio (SNR) and increased sensitivity to blood oxygenation level dependent (BOLD) contrast, which is widely used for functional neuroimaging [2–4]. The increased spin-lattice relaxation time for blood makes the measurement of cerebral blood flow (CBF) by arterial spin labelling a more sensitive method at higher magnetic fields [5] and there are also attendant advantages to the performance of MR spectroscopy. The increased SNR, which varies approximately linearly with field strength, can theoretically be used to produce MR images with higher spatial resolution.

Conventional wisdom has suggested that the production of high-resolution images of the entire brain with uniform contrast behaviour may be confounded at high field by radiofrequency (RF) inhomogeneities. At magnetic field strengths of 3 T and higher the dielectric resonance effect and RF penetration issues [6] cause the flip angle to vary across the field of view (FoV) of the image. This produces spatial signal intensity variations and non-uniform contrast. For many MRI sequences, either the RF flip angle must be chosen as a compromise taking account of these variations, or specialised shaped RF pulses must be employed that compensate for excitation pulse inhomogeneities [7] or are less sensitive to them (adiabatic pulses).

Fast spin echo (FSE) imaging [8, 9] (also known as turbo spin echo (TSE) or multishot rapid acquisition with relaxation enhancement (RARE)) is a method commonly used for the acquisition of diagnostic MRIs at clinical field strengths [10]. The FSE approach produces high quality images with contrast similar to that of conventional spin echo images and with the benefit of a greatly reduced acquisition time. Clinical applications of FSE include $T_2$ relaxometry and assessment of neurological disorders [11–14]. While the application of FSE imaging at high field is desirable, concerns exist regarding power deposition and image uniformity. It is probably for these reasons that other methods have been more commonly employed to generate structural brain images at high field strength [15–17].

The purpose of this paper is to demonstrate that, with accurate sequence optimization, FSE can be used to acquire good quality high resolution structural images of the human brain within practical examination times at high field strength. FSE images from a 4.7 T scanner are presented that display excellent contrast, spatial resolution and SNR.

Methods

Between August and November 2002, 10 healthy subjects (6 males, 4 females, age range 26–48 years, median age 30 years) were scanned with FSE after informed consent and approval from the University College London Hospital Ethics Committee. All imaging
was performed using a 4.7 Tesla, 90 cm bore-diameter magnet (Magnex Scientific Ltd, Oxford, UK) controlled by a console supplied by Philips Medical Systems (Eindhoven, The Netherlands) based on a MR5000 design by SMIS Ltd (Guildford, UK). A shielded head gradient coil was used providing gradient fields of up to 36 mT m⁻¹. Images were acquired using a quadrature birdcage RF coil with an internal diameter of 28 cm. The FSE sequence had an echo train length of 8 echoes, with the first echo at 22 ms and an echo spacing of 22 ms. Sinc-shaped RF pulses were used for both signal excitation and refocusing, with relative amplitudes 1:1.8.

The repetition time (TR) was chosen either as 3.5 s for 17 slices or 7 s for 34 slices. Along the read axis, 512 points were acquired (sampling bandwidth 50 kHz) and 384 phase encoding (pe) steps were performed (with a factor of two oversampling) resulting in total acquisition times of 5 min 40 s (17 slices) or 11 min 20 s (34 slices). The slice thickness was 2 mm and the slice acquisition order was chosen to be either sequential with 2 mm gaps or interleaved with no gap between slices. In this way slices acquired consecutively in time were always at least 2 mm apart (to minimize slice interaction effects). The images presented here were acquired in axial and coronal orientations. For the axial images, a FoV of 240 mm × 180 mm (read axis × phase encoding axis) was chosen, resulting in an in-plane resolution of 469 μm × 469 μm. For the coronal images, a FoV of 200 mm × 200 mm was used (in-plane resolution: 391 μm × 521 μm). FSE imaging has also been performed with sagittal and oblique orientations (data not shown).

The specific absorption rate (SAR) for the FSE sequence with the above parameters was calculated to be below current safety limits of 4 W kg⁻¹ for the head (for short exposure times [18]).

The nominal echo time (TE) corresponds to the time of acquisition of the central area of k-space [19] and was chosen either as 22 ms, 44 ms or 66 ms (corresponding to the 1st, 2nd or 3rd echo, respectively).

The exact scheme of k-space sampling has been shown to play a significant role in final image quality [9, 20]. The k-space coverage strategy in the pe direction was designed to ensure that when the data was combined prior to Fourier transformation (FT), signal intensity differences between echoes produced a relatively smooth amplitude variation through k-space. Therefore, the corresponding point spread function [19] did not contain significant sidebands and good image resolution was preserved. The 512 × 768 data matrix was zero-filled to 512 × 1024 and a Hanning filter [21] was applied along both directions prior to image reconstruction (FT). Due to the 2x oversampling in the pe direction, only the central portion of the image (a 512 × 512 matrix) was selected and stored. The final images were then obtained after filtering with SharpView® [22], an image enhancement package currently installed on many clinical scanners.

## Results

All 10 volunteers were successfully scanned with FSE. We illustrate our findings on FSE images from three subjects. Figure 1 shows four transverse sections from a 34-slice data set obtained on a 29-year-old male healthy volunteer with TE=22 ms. The images appear to be predominantly T₂ weighted with good grey-to-white matter (GM/WM) contrast. Figure 2 shows the same images displayed with an inverted grey scale (with Figure 2d showing an expanded region from Figure 1d). In these images blood vessels appear bright and are more clearly visible against surrounding dark background. Some details in the white matter are also more easily visible in these images including Virchow-Robin spaces (bright in conventional contrast and dark in inverted contrast, for example see arrows in Figure 2b).

In the images of Figure 1b and 2b it is worth noting the appearance of the tail of the caudate nucleus (see arrows on Figure 2b). In Figures 1c and 2c the medullary laminae (arrow on Figure 2c) separating internal and external segments of the globus pallidus are also distinguishable. In Figures 1d and 2d the red nuclei and the substantia nigra appear hypo/hyperintense due to the effect of iron accumulation [23–25]. Fornices and mammillo thalamic tracts are also visible on the same images. Figures 3a and b show coronal sections from 17-slice data sets on two of the volunteers (male 30 years, TE=66 ms and female 34 years, TE=44 ms) with inverted grey scale. These images, as well as displaying excellent overall detail, allow easy differentiation of blood vessels and Virchow-Robin spaces. Figures 3a demonstrates good visualization of the internal capsule while in Figure 3b different layers of the hippocampus can be identified (alveus, cornu ammonis and gyrus dentatus, see arrows on Figure 3c). Despite the difference in TE the images display similar overall contrast. The GM/WM contrast is good and reasonably homogeneous in all the images shown.

## Discussion

We have shown that with the correct choice of sequence and optimization of experimental parameters, T₂ weighted imaging can be performed with FSE at 4.7 T and displays the gain in quality expected from a high field system. Images with submillimetre in-plane resolution, high SNR and good structural contrast covering large brain regions can be produced in clinically acceptable acquisition times. Here we discuss particular features of our images and how they arise from the combination of tissue properties and sequence characteristics.

FSE images have a similar overall appearance to conventional T₂ weighted images. Clinically, FSE has replaced conventional spin echo imaging as the method of choice for diagnostic T₂ weighted imaging and T₂ relaxometry [11, 12]. At high field, T₂ values of brain tissue become shorter and more convergent than at 1.5 Tesla (63 ± 6 ms for cortical grey matter and 50 ± 2 ms for white matter at 4 T [26] vs. 91 ± 6 ms and 69–76 ms at 1.5 T [27]) and T₂ differences do not suffice to explain the contrast between grey and white matter seen in our images. This contrast can be explained by several other factors [9] such as the T₁ weighting introduced by the presence of stimulated echoes and the magnetization transfer effects.

In a multiecho spin-echo sequence comprising a train of refocusing pulses, the signal intensity decreases throughout the echo train due to T₂ relaxation. This signal decrease also depends strongly on flip-angle and therefore, due to RF inhomogeneity, will vary spatially resulting in signal non-uniformity across the image. In FSE, stimulated echoes compensate for the signal loss of the spin echoes,
Figure 1. Fast spin echo high resolution transverse MRIs of a healthy volunteer (male, 29 years) displayed with conventional grey scale. Slice thickness 2 mm, in-plane resolution 469 µm x 469 µm, number of slices 34, total acquisition time 11 min 20 s. Figures (a) to (d) display four different sections from the top of the brain to the level of the substantia nigra. The white rectangle in Figure 1d indicates the region expanded in Figure 2d.
Figure 2. Same images as in Figure 1, presented with an inverted grey scale (with Figure 2d showing an expanded region from Figure 1d). Structures mentioned in the text are highlighted with grey arrows: tails of caudate nuclei and Virchow-Robin (VR) spaces (b); medullary lamina (c); fornix, mammillary tract, red nucleus and substantia nigra (d).
giving a degree of robustness to RF inhomogeneity, and the combined echo amplitude is dependent on $T_1$ as well as $T_2$ [28, 29]. At 4.7 T, the RF homogeneity within the brain is expected to be worse than at 1.5 T due to RF penetration and dielectric resonance effects [6]. However, the images shown here have relatively uniform signal intensity and contrast because of the aforementioned flip-angle insensitivity of FSE [30] in combination with the RF profile of a conventional birdcage coil.

Magnetization transfer (MT) effects [31], including both true MT and direct saturation, also contribute to the contrast in our images [16, 32]. These effects arise in the multislice FSE sequence because of the repetition of numerous refocusing pulses applied close to the resonance frequency of the imaged tissue during the acquisition of neighbouring planes [33]. Direct saturation affects grey and white matter by approximately equal amounts due to their similar $T_1/T_2$ ratios (~20 [26]). However MT increases contrast by preferentially attenuating myelinated tissue. The combination of $T_2$ weighting and MT may also account for the similar overall contrast observed for images acquired at different TE values [33].

Our images at 4.7 T were characterized by high contrast resolution as well as high spatial resolution resulting in excellent GM/WM contrast, good visualization of structures with high iron content and the identification of a high number of leptomeningeal blood vessels and Virchow-Robin spaces.

The GM/WM contrast appears to be good across all the slices. In addition a great deal of detail is visible in subcortical regions making the identification of anatomical structures, e.g. internal capsule, caudate nucleus, globus pallidus, medullary laminae, simpler than on equivalent images acquired at lower field strengths. Furthermore, it is interesting to note that the GM/WM contrast is enhanced when the volume coverage in the FSE sequence is increased. This is because the number of slices determines the minimum TR for a given resolution. For example the coronal 17-slice data sets shown in Figure 3 were acquired in 5 min 40 s (TR = 3.5 s), while the axial data (Figures 1 and 2) covered 34 slices in twice the time (TR = 7 s). The longer TR in the 34-slice sequence gives more time for longitudinal relaxation of the MR signal, resulting in a higher image SNR and better contrast.

Structures with high iron content such as the substantia nigra and red nuclei are also easily identifiable due to...
their low signal in the FSE images. The increased iron concentration causes a reduction in $T_2$ relaxation time and this effect is enhanced at high magnetic fields [24, 25]. This iron-mediated contrast mechanism could allow the use of FSE for the investigation of movement disorders, where it is known that localized iron accumulation is abnormally high [34, 35].

Our images also allow easy identification of blood vessels within the cerebrospinal fluid (CSF) space (leptomeningeal vessels). This is due to the strong appearance of CSF, which has a long $T_2$ value, combined with the low signal intensity of intravascular blood. The scrambling of the spin phase of the blood due to in-plane flow results in incomplete rephasing of its signal prior to application of each refocusing RF pulse. Through-plane flow also causes blood signal dropout if spins move between slices during the echo train [36].

In addition, we have already mentioned the great number of identifiable Virchow-Robin spaces in our images. This number is remarkable with respect to 1.5 T observations, especially considering the relatively young age of the subjects whose images have been presented. One reason for this is the good contrast between Virchow-Robin spaces (characterized by high signal due to their fluid content) and the surrounding white matter (lower signal). Probably more important, however, is the higher spatial resolution of our FSE images compared with conventional images obtained at lower field.

While it is true that the spatial resolution of the images presented here is better than that of a standard clinical structural image, e.g. an FSE image obtained at 1.5 T in comparable time, higher resolution MR images acquired at 3 T and above have been published [17, 37, 38]. However, these studies have all focused on specific areas of the brain, mostly using RF receive coils of limited coverage and these studies have all focused on specific areas of the brain, e.g. the hippocampus is feasible in temporal lobe epilepsy. Neurology 2002;58:257-64.


Conclusions

In summary, we have demonstrated that with appropriate optimization of sequence parameters, the FSE technique can be used efficaciously at field strengths higher than 3 T and can deliver increased spatial resolution with useful tissue contrast across the whole brain within current safety limits.

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References


