Implementation of high flip angle Fast Spin Echo imaging at 4.7T without exceeding safety limits: application to human brain imaging

D. L. Thomas[1], E. De Vita[1], S. Roberts[2], R. Turner[1,3], R. J. Ordidge[1]

[1]Wellcome Trust High Field MR Research Laboratory, Department of Medical Physics and Bioengineering, University College London, London, United Kingdom, [2]MR Research Systems, Guildford, United Kingdom, [3]Wellcome Department of Imaging Neuroscience, Institute of Neurology, University College London, London, United Kingdom

Introduction Although high field systems can be beneficial to the majority of MR imaging protocols, the acquisition of images with T_2 weighting at high field strength (≥3T) is only recently taking off. This is mainly due to the high power deposition (or specific absorption rate, SAR) associated with conventional Fast Spin Echo (FSE) imaging sequences (employing a train of 180° pulses) and the increased RF power required at high fields to obtain a given flip angle with respect to lower fields. Additionally, due to the reduced wavelength (in tissue) associated with an RF pulse of the appropriate Larmor frequency, the interactions between RF-field, sample and RF coil become increasingly important at high field [1] and make the B_1 distribution extremely inhomogeneous. As a consequence, the flip angle associated with an RF pulse becomes spatially variable and can produce image contrast and contrast-to-noise non-uniformity, especially in multi-echo spin-echo sequences. Recent approaches have explored the potential of FSE when a train of RF pulses (of generally low flip angles) with specially optimised amplitude (e.g. TRAPS [2], Hyperechoes [3] or quadratic phase modulation methods [4]). Here we show that the combination of a relatively low acquisition bandwidth and a long echo spacing, together with a simple modification to the k-space coverage scheme, allows T_2-weighted images with remarkable quality (signal to noise and anatomical detail) to be obtained at 4.7T without exceeding current SAR limits and minimal sequence modifications. Furthermore, the images show a reasonable level and uniformity of contrast across the brain.

Methods All imaging was performed on a 4.7 Tesla system (SMIS MR5000 provided by Philips). The FSE method was optimised for 4.7T [5] and comprised a train of 8 echoes with an inter-echo spacing of 22ms, an acquisition bandwidth (BW) of 50kHz and a repetition time of 3.5s (for 17 slices) or 7s (for 34 slices). The data matrix was 512(read) x 768 (phase, with 2x oversampling), yielding an acquisition time of 5min40s for 17 slices or 11min20s for 34 slices. The Field of View (FoV) was 240mm x 180mm (yielding 470x470μm^2 in-plane resolution). All RF pulses were standard unfiltered sinc pulses of 2.66ms duration (bandwidth 1500Hz) and the refocusing pulses were chosen to be 1.8 times the amplitude of the excitation pulse (based on numerical simulations to maximise signal intensity). The image slice thickness was 2mm, and a slice interleave scheme was implemented to reduce slice interference effects (slice ordering: 1,3,5,…2,4,6…). Images were acquired for several effective echo times (TE_{eff}) corresponding to the first three echoes (22ms, 44ms, 66ms) and SNR and grey-white matter contrast were measured. To cover k-space, the rotated centric-type phase encoding scheme [6] was used with a modification that we call *feathering*, demonstrated in Fig 1. Considering that the signal at each position in k-space along the phase encode (*pe*) direction has relaxed according to the echo time at which it was sampled, sudden steps in signal amplitude result (Fig 1a). These steps are larger for longer echo spacing, resulting in strong point spread function (PSF) sidebands (Fig 1b) and potentially causing 'ringing' artefacts in the images. The feathering replaces the sudden steps in k-space with an oscillation over a chosen distance in k-space (fig 1c) and displaces the odd sidebands of the PSF to the edges of the FoV in the *pe* direction (fig. 1d). If oversampling in the *pe* direction is employed, these sidebands (which cause low intensity ghost images) do not overlap with the main image, leaving it free of artefacts. Simulations show that for an exponential decay of the signal along the echo train, the optimal feathering fraction is 50% (*i.e.* ¼ of the echoes on both sides of the regions of constant amplitude in Fig 1a undergo *feathering*). Though the oversampling implies a doubling of the total acquisition time, its effect on SNR is equivalent to averaging. The SAR was estimated by considering the power output of the RF amplifier, measuring the power losses in the RF chain from the amplifier to the RF coil, and coil loading.

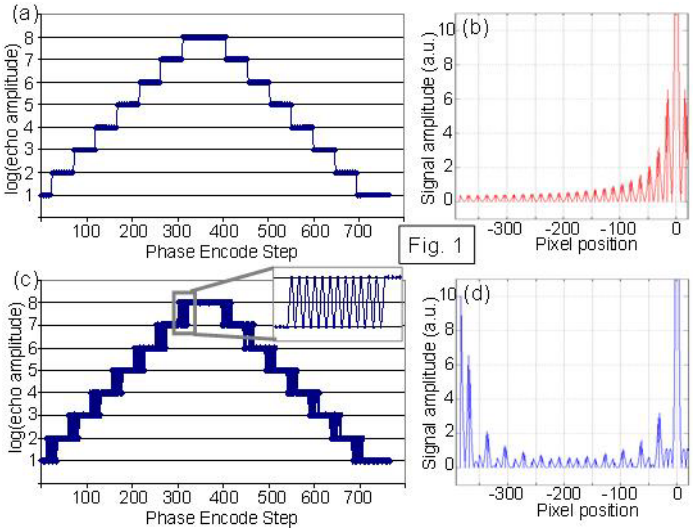

Fig. 1

Results Figure 2 shows representative axial FSE images acquired with TE_{eff}=22, 44, 66ms. Whilst the overall SNR increases for all tissues when TE_{eff} is reduced, good contrast is seen in all the images. This is due to the fact that MT effects compensate for the loss of proper T_2 contrast [5]. No artefacts are visible in the images. The SAR of the sequence was calculated to be approximately 3.3W/Kg.

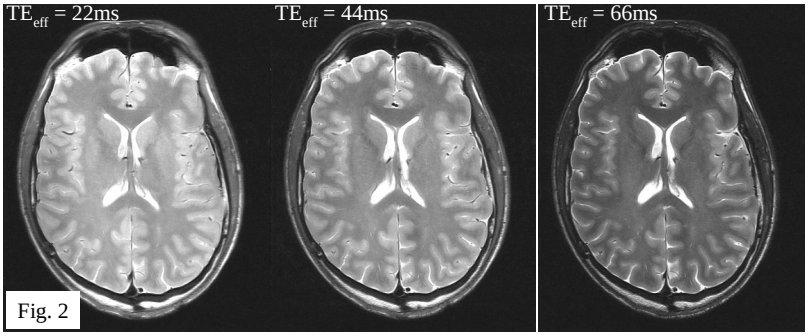

Fig. 2

Conclusions The images obtained with the optimised FSE sequence at 4.7 Tesla employing *feathering* for the phase encode ordering show good and relatively even contrast between tissues across whole brain sections. The low BW employed means that the RF duty cycle is kept low and high flip angle pulses can be used without exceeding SAR limits. The combination of low BW and 2x oversampling yield excellent SNR in acceptable acquisition times. The images display typical 'T_2-contrast', though the contrast is a combination of T_2-weighting, magnetisation transfer and T_1-weighting due to stimulated echo contributions. Our results show that optimised low flip angle techniques might not be a necessity at high field, though their use may well provide added flexibility to pulse sequence design and parameter choice and thus deliver further improvements to the FSE approach.

References [1] Ibrahim *et al.*, MRI 18:733-742 (2000). [2] Hennig *et al.*, MRM 49:527-535 (2003). [3] Hennig and Scheffler, MRM 46:6-12 (2001). [4] Le Roux, JMR 155:278-292 (2002). [5] De Vita *et al.*, BJR 76:631-637 (2003). [6] Constable *et al.*, MRM 28:9-24 (1992).

Acknowledgments We would like to thank the Wellcome Trust for support of this research.