Development of Ultra Short Echo (UTE) imaging for in vivo application at high field strength

A. A. Khrapitchev[1], N. R. Sibson[1], M. D. Robson[2], and A. M. Blamire[3]

[1]Physiology, Anatomy and Genetics, University of Oxford, Oxford, United Kingdom, [2]Oxford Centre for Clinical Magnetic Resonance, University of Oxford, United Kingdom, [3]Newcastle Magnetic Resonance Centre, University of Newcastle, United Kingdom

Introduction

Ultra short echo (UTE) imaging is a class of experiments which attempt to acquire data with effective echo times ranging from 50–150 μs (1). Such methods allow imaging of fast relaxing spins with transverse relaxation times above ~ 40 μs. Recently UTE sequences have been demonstrated on clinical MRI systems and have revealed interesting image contrast across a number of body areas (1, 2). Of particular interest to us is the suggestion that image contrast seen using UTE in a patient with multiple sclerosis is reporting on the demyelination process of this disease (1).

In order to determine the fundamental nature of the UTE signal seen *in vivo* requires direct correlation of the UTE signal with quantified levels of myelin in the tissue. This can be achieved by applying UTE sequences in small animal models with direct comparison to histological specimens. Here we describe the first step in this process, to develop UTE for high field, small animal work.

To achieve such short echo times requires self-refocussed RF pulses and centric radial K-space sampling (akin to back-projection techniques) to eliminate delays associated with slice refocussing, read-dephasing and phase encoding. Further, since significant relaxation will occur during data sampling itself, potentially limiting the achievable resolution, it is desirable to work with data sampling time of the order of T_2. This requires high readout gradient strength, large sampling bandwidth and complete compensation of B_o and gradient eddy currents. The manufacturing tolerances for small scale gradient sets are *proportionally* much larger than in clinical systems, and the gradient fields produced are increased, so gradient system imperfections (e.g. gradient cross terms, residual eddy current effects, etc) may have a more dominant effect on image formation than in clinical systems. In this paper we report the technical challenges associated with implementing UTE at high field on a small bore system and discuss measurements and correction schemes to achieve UTE data.

Methods

The UTE sequence (3) was implemented on a 7T Varian Inova spectrometer running vnmr6.1c software. The system is equipped with a Magnex/Varian self-shielded gradient system capable of 175 mT/m with a rise time of 80 μs, which was lined by a copper RF shield to prevent noise insertion via the gradient system.

Phantom imaging experiments were performed using an doped agarose filled (29 mm OD) Varian microimaging phantom [figure 1(a)] and a birdcage RF coil. Each radial FID consisted of 64 data points sampled at 100 kHz. Slice selection used a half-sinc RF pulse. *In vivo* imaging was performed in fresh, post-mortem rat brain with resolution of 390 μm and a 1 mm slice (TR = 1.0 s, $TE_{effective}$ = 50 μs).

Gradient performance was evaluated according to published methods (4) using a water-filled 4 mm i.d. NMR tube aligned coaxially with the magnet bore to approximate a point-source phantom in the transverse plane. Data were collected at isocentre to measure B_o eddy current effects and at several locations along the X and Y axes to measure the gradient system transfer function including the spatially dependent eddy current terms.

Image processing was performed using regridding of the radial K-space data using standard algorithms with corrections for gradient performance included.

Results

(i) Although our system routinely acquires high quality data using fast imaging sequences such as EPI, basic image quality assuming ideal hardware with the UTE sequence was poor [figure 1(b)].

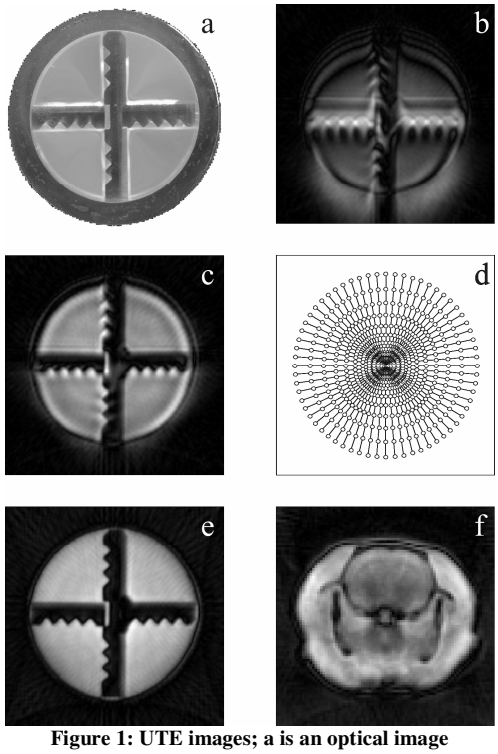

Figure 1: UTE images; a is an optical image (see text for details)

Characterisation of the gradient performance demonstrated a significant B_o eddy current term from pulsing the Y gradient, which was characterised by an initial slope due to the rising field gradient and an exponential decay with time constant (~25 μs). This B_o term causes phase roll across K-space, producing a smearing artefact [figure 1(b)], but which could be corrected by phase correction of each FID [figure 1(c)]. The source of these eddy currents was traced to the RF shield within the gradient coils. When this was removed the eddy-current effects were largely eliminated (data not shown).

(ii) The gradient waveforms on our system run at constant slew rate, which results in gradient rise time dependent on the required demand. Thus risetime varies with each radial projection with each individual K-space trajectory reaching a plateau at a different time within the acquisition window. This leads to disortion of K-space around the origin [figure 1(d)]. Correction of the K-space trajectory in post-processing led to a further improvement in image quality [figure 1(e)].

(iii) Figure 1(f) shows a transverse UTE image through the rat brain, demonstrating the first high field *in vivo* application of UTE imaging. Data were collected with the RF shield removed from the system.

Conclusion

The UTE sequence was implemented at 7 T on a small bore MRI system. System components which did not affect other sensitive fast imaging methods (e.g. EPI) were found to have a major effect on image quality. Full characterisation of gradient ramp-time effects and short time constant eddy currents were employed to trace and correct sources of artefacts. Good quality images were obtained in the rat brain at an effective echo time of 50 μs. These images will allow the source of the short T_2 components to be explored in small animal models and hence inform the use and interpretation of clinical UTE data.

References

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