A comparison of UTE versus SPRITE for Robust MR Imaging of short T_2 components

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Introduction

Conventional MRI performs well in biological tissue where the lifetime (T_2) of the free induction decay from the main tissue components of interest is relatively long. Short T_2 signal components (<< 500μs) do however exist in biological tissues and cannot be directly detected using traditional MRI. These components represent an, as yet, under-investigated component of the *in vivo* MR signal. Very short T_2 components also exist in other non-medical applications where non-invasive imaging may be important.

Single-Point Imaging (SPI) techniques are quite distinct from conventional MRI methods and are able to cope with these short lifetime limitations. Unlike conventional MRI, where the time evolution of magnetisation is measured by frequency encoding, SPI is a pure phase encoding technique which, as the name suggests, encodes only a single data-point per signal excitation. The most generally applicable form of the SPI MRI methods is the SPRITE (Single Point Ramped Imaging with T_1 enhancement) family of techniques [1], but these require 3D phase-encoding which has limited their area of application. Alternatively, the Ultra Short Echo (UTE) techniques [2] can image short T_2 components and do incorporate either slice selection or 3D acquisition. UTE samples multiple data-points per signal excitation linking the point spread function in the image to the species T_2*.

In this abstract we implement and compare the performance of SPRITE and UTE for high field *in vivo* imaging in small animal models.

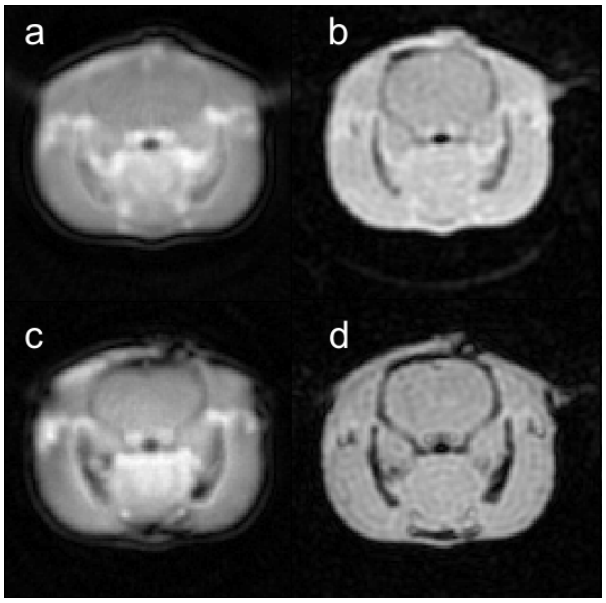

Figure 1: UTE (a, c); SPRITE (b, d) (see text for details)

Methods

SPRITE and UTE sequences were implemented on a 7T Varian Inova spectrometer running vnmr6.1c software (Varian Inc, Palo Alto, USA). The system is equipped with a gradient system capable of 175 mT/m in a rise time of 80 μs (Magnex Scientific, Yarnton, UK). A circularly polarised RF volume coil was used for all data acquisition.

SPRITE: The 3D SPRITE sequence was implemented acquiring K-space data as a series of interleaved trajectories [3]. Signal excitation used a 6 degree, 5 μs hard pulse followed by the requisite phase accumulation delay to achieve the required "echo" time (TE). Each of the trajectories acquired k-space data-points over a 340 ms/3.4 s total duration for 100 μs (Fig. 1b) and 1.0 ms (Fig. 1d) encoding times respectively which was followed by a 1.0 s delay (to allow for relaxation and due to gradient duty cycle constraints). FOV was set to 40 mm by 40 mm for a 64*64 matrix. All sampled k-space data-points were positioned on a rectilinear grid and resulting data were processed using standard 3D Fourier transformation.

UTE: To allow direct comparison of data collected with the 2 sequences, a 3D encoded UTE sequence was used. This consisted of a 30 degree, 25 μs non-selective excitation pulse followed by sampling of the FID during the rising edge of the radial readout gradient [2]. This sequence allows a true knowledge of the delay time between signal excitation and sampling of the k-space origin. Effective "echo" time was varied matching the SPRITE acquisition (100 μs, 1.0 ms) and repetition time was 0.1 s. (Field of view 40 mm, 128 projections, 400 kHz sweep width, 64 rotations). Image reconstruction of UTE data used regridding from radial to rectilinear followed by conventional 3D Fourier transformation (fig 1a – 100 μs; fig 1c – 1.0 ms).

Sequences were tested and compared in short T_2 phantoms and then *in vivo* to image fresh post-mortem specimens focussing on the brain.

Results and discussion

We have compared for the first time the performance of the UTE and SPRITE sequences applied to *in vivo* imaging. At very short TE (which is the preferred regime for SPRITE acquisition), image quality was found to be very comparable between methods. For very short T_2 component signals the short encoding time used in SPRITE allows truly quantitative images, since signal amplitude and image point spread function is no longer a complex function of the signal lifetimes. Typical *in vivo* images collected using the two sequences are shown in figure 1 (left UTE; right SPRITE). The images show identical slices from consecutive scans in the same animal at a short and long echo time. The SPRITE image is theoretically expected to show some T_1 contrast but otherwise to yield true proton density information which is consistent with the uniform nature of the intensity within the soft tissue. The UTE image shows generally the same uniform intensity although greater T_1 contrast (associated with the higher flip angle used to maximise signal to noise ratio) can be seen. In areas of susceptibility gradient near the top of the skull, the UTE image shows reduced signal due to the longer readout sampling of multiple data-points, while the equivalent SPRITE image is largely unaffected. However, the main drawback of the SPRITE implementation is the prolonged experimental time (55 min/115 min for 100 μs/1.0 ms encoding respectively) by comparison to UTE (40 min). This is partially due to the gradient duty cycle and the sequence loading time, which can be overcome in the future. Therefore SPRITE is preferred for imaging application where resolution is crucial or gradient eddy-currents are a problem; on another hand UTE is faster, has less the gradient duty cycle requirements and allows a range of T_1 weighting.

References

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