Application of Magnetic Resonance Fingerprinting to measure brain oedema

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Abstract

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1. Introduction

Magnetic resonance imaging (MRI) has become an established tool for diagnosis and disease monitoring in clinical environments. As with the phenomenon of nuclear magnetic resonance (NMR), MRI makes use of the behaviour of nuclear spins in a magnetic field. However MRI has the crucial difference that it allows the acquistion of images from the behaviour of nucleic spins. When a strong, static, magnetic field B_0 is applied to a sample, there is a net alignment of the spins to B_0, producing net magnetisation along the longitudinal axis of B_0. Radio frequency (RF) excitation pulses can be applied to tip the net magnetisation of a particular species of nuclei so that it is no longer aligned with B_0. This produces components of magnetisation that are orthogonal to the static field (transverse magnetisation) and results in reduced longitudinal magnetisation. With the removal of B_1, the tipped spins will precess around the axis of the static field. Over time and during the precession, the transverse and longitudinal components decay and recover, respectively. The evolution of these two components occurs according to exponential terms, as described below. Ideally, the transverse component decays towards zero, according to Eq. (1), where t is the time elapsed, $M_xy(t)$ is the transverse magnetisation at t and

[1] Since 1880.

T_2 controls the rate of decay. In this case, the decay is due to the dephasing of spins, due to interaction between different spins. However, in practice, the constant T_2^* determines the rate of decay, because of spatial inhomogeneities in the static field.

Equation (2) describes the influence that the relaxation due to B_0 inhomogenties has on the total transverse relaxation, where the effect of the inhomogenities is represented by $T_2^{'}$. The inhomogeneities cause variations in the strength of the static field experienced by different spins. The frequency of precession is proportional to the static field strength and therefore static field inhomogeneities will cause spins in different locations will precess at slightly different rates.

$$M_{xy}(t) = M_{xy}(0) \exp^{-\frac{t}{T_2}} \tag{1}$$

$$\frac{1}{T_2^*} = \frac{1}{T_2} + \frac{1}{T_2^{'}} \tag{2}$$

Equation (3) describes the behaviour of the longitudinal magnetisiation, where t is the elapsed time, $M_z(0)$ is the initial longitudinal magnetisation and $M_{z,eq}$ is the equilibrium magnetisation that would occur, given B_0, but in the absence of excitation pulses. The constant T_1

$$M_z(t) = M_{z,eq}(1 - e^{-\frac{t}{T_1}}) + M_z(0)e^{-\frac{t}{T_1}} \tag{3}$$

The precessing magnetisation induces current in receiver radio frequency coils, which decays as the magnetisation realigns with the static field. By observing the induced current, the relaxation parameters $T1_1$ and T_2 can be measured on a pixel-wise basis. Field gradients are applied to produce controlled differences in spin frequency and phase, which can be used to enable the spatial origin of a signal to be determined [1].

A large portion of MRI research has been directed towards acheiving accurate, efficient and reliable images of the human brain. The magnetisation

properties of different tissue types can makes it possible to acheive contrast between areas of different tissue types.

Commonly used approaches for measuring T1 and T2, such as inversion recovery (IR) and spin echo (SE) require a significant period of time. * mention the fastest times?*

Within the setting of emergency medicine, fast response and diagnosis is cruicial for giving the patient the best chance of a positive outcome. Specifically, patients suffering from acute stroke benefit will benefit from fast assessment and treatment. A stroke is the loss of blood supply to a section of the brain. Commonly, strokes are the result of blockages in the surrounding blood vessels (ischemia), but bleeding, in or around the brain tissue, can alsombe a cause. Temporary strokes can occur when surronding blood vessels are temporarily blocked (transient ischemia). There is a danger that the affected region of the brain will experiecne damage and cell death, due to the lack of oxygen and nutrients. This damage is often exhibited as extracellular water build up (Oedema) or a change in pH. In order to guide subsequent treatment, it is important to be able to measure these the markers accurately and precisly. There is also a need for a fast assessment, in order to allow a patient to be treated soon after the onset of the stroke.

Recently, a new approach known as magnetic resonance fingerprinting (MRF) has been developed and applied, which has been used to acquire multiple parameters, such as T1 and T2, with a reduction in scan time [2] . During a fingerprinting experiment, a a pesudo-random seqeunce is used to manipulate the spins in the sample. This can be achieved by varying factors such the time in between subsequent RF pulses and any flip angles used.

Once a series of images have been acquired via the fingerprinting sequence, the pixel-wise signal variation over the experiment is compared to a previously built dictionary of time courses, in order to find the closest match. The dictionary is built by simulating the signal expected from the experiment, for variety of difficult combinations of parameters (such as T1 and T2).

2. Materials and Method

2.1. Image aqcuistion

The image data for this report were acquired at the acute vascular imaging centre (AVIC) at the John Radcliffe Hospital in Oxford, using a 3T strength Siemens Verio magnet and a 12 channel Siemens head coil. A spin echo sequence was used, to acquire 48 images, via 48 repetitions. The TE, TR, and flip angles for each repetition were varyied, using a pre-written, pesudo-random, list.

Using MATLAB (The MathWorks, Natick, MA), a dictionary was composed for *mention T1 and T2 values used*. In order to account for some variations in the B_1 field, deviations in the flip angle of $\pm 30\%$ ()were also used in the creation of the dictionary.

In order to test the performance of the method with a sample containing various properties we created a phantom with a range of T1 and T2 values.

3. Results and Discussion

4. Conclusions

References

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