Application of Magnetic Resonance Fingerprinting to measure brain oedema

Jack Allen, Supervisors: Peter Jezzard & James Kennedy

University of Oxford, Oxford, UK.

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. The angle that the magnetisation is tipped by is known as the flip angle (FA). The process of tipping 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, eventually realigning with B_0. During the process of 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 T_2 determines the rate of decay. In this case, the decay is due to interactions between spins, which cause differences in the precession frequencies. If spins precess at different frequencies, they will become out of phase with each other, resulting in a reduction in the net transverse magnetisation. However, in practice, the constant T_2^* determines the rate of decay, because spatial inhomogeneities in the static field also play a role in produce differences in precession frequency.

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 $T_2^{'}$. Inhomogeneities cause variations in the strength of the static field experienced by different spins. Therefore, because the frequency of precession is proportional to the static field strength, static field inhomogeneities will cause spins in different locations will precess at slightly different rates and become dephased.

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

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

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 determines the rate at which the longitudinal magnetisation chnages over time.

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

The precessing transverse magnetisation induces current in receiver coils, which is interpreted as signal. By altering experimental settings and observing the changes in the observed signal, the relaxation parameters $T1_1$ and T_2 can be measured.

[1] Since 1880.

Field gradients are applied to produce controlled differences in spin frequency and phase across a sample, which can be used to enable the signal measurements to be mapped and images produced [1].

A large portion of MRI research has been directed towards acheiving accurate, efficient and reliable images of the human brain. Many imaging experiments make use of the differences in properties such as T1 and T2, to acheive contrast between tissue types such as grey and white matter.

Commonly used approaches for measuring T1 and T2 rely on altering the magnetisation with radiofrequency pulses and measuring the signal at different time points afterwards. For example, spin echo (SE) experiments involve two RF pulses before the acquistion of an image. The pulses allow recovery of the signal that is lost due to the dephasing effect from inhomogeneities in B_0. The second pulse causes rephasing (an 'echo') at time TE after the first pulse. Parameter measurements usually requiring a significant amount of time.

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 also be 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 by a change in the pH value of the affected tissue. 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 simultanously measure multiple parameters, such as T1 and T2, with a reduction in scan time [2] . During a fingerprinting experiment, a pesudo-random seqeunce is used to manipulate the spins in the sample. This can be achieved by varying factors in the seqence, such the time in between subsequent RF pulses and the size of any rotations in the magnetisation that may be induced by the RF pulses.

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.

2.2. Dictionary creation

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.

2.3. Phantom production

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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