# A Rapid Two-Shot T1-measurement and its Applicability to Oxygen Partial Pressure Measurements in Fluid

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Oxygen partial pressure ($pO_2$) in fluids such as CSF, amniotic fluid and urine provides a means to detect hypoxia in tissues producing and in contact with the fluid (1). All previous measurements of body fluid $pO_2$ have required invasive means, which are prone to sampling error and contamination with room air. It has been shown that the T1 of water varies with $pO_2$, allowing $pO_2$ to be detected non-invasively with MRI (2-5). Current T1 measurement techniques, however, can be very time consuming since TR times need to be long to distinguish T1 differences in long-T1 fluids. Here we present a method to acquire a high resolution T1 map in as little as 4 seconds per slice using a pair of single-shot fast spin echo (SSFSE) acquisitions.

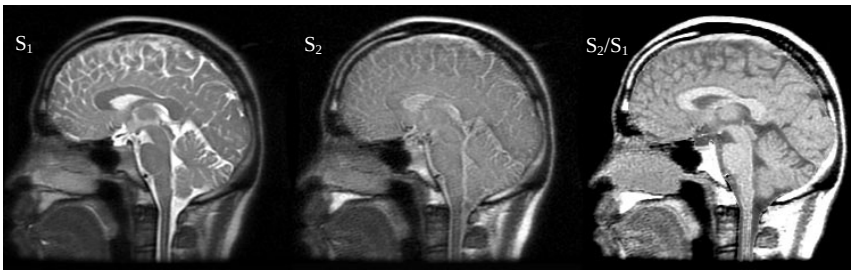


Figure 1: Acquired $S_1$ (T2w) and $S_2$ (mixed T1,T2w) images, and calculated $S_2/S_1$ (T1w) image.

***Methods***

Two single-shot fast-spin-echo (SSFSE) images are acquired in succession. If magnetization is fully recovered prior to the first acquisition, but saturated by the long train of refocusing pulses and only partially recovered prior to the second acquisition, then signal (image intensity) of the two acquisitions may be written as

$$S_1 = C\,M_0\exp(-TE/T2) \quad (1)$$

$$S_2 = C\,M_0\left(1-\exp(-T_{SR2}/T1)\right)\exp(-TE/T2) \quad (2)$$

where $M_0$ is the initial magnetization, $T_{SR2}$ is the saturation recovery period prior to the second acquisition, and C is a function that takes into account coil sensitivity and all other aspects of the measurement process that convert transverse magnetization into signal. The ratio of these two measurements depends only on T1:

$$\frac{S_2}{S_1} = 1-\exp(-T_{SR2}/T1) \quad (3)$$

Performing this operation on a pixel-by-pixel basis may be done to generate a T1-weighted image, as shown in Fig. 1. The T1 may also be directly estimated by solving Eq. 3 for T1:

$$T1_{est} = \frac{-T_{SR2}}{\ln\left(1-\frac{S_2}{S_1}\right)} \quad (4)$$

Non-selective refocusing may be used to eliminate variation in saturation due to inflow. In this case, when multiple slices are acquired finite TR leads to incomplete longitudinal recovery before the first acquisition, and Eq 4 will under-estimate T1. However if the saturation recovery time prior to the first acquisition ($T_{SR1}$) is known, the effect of partial saturation on $S_1$ may be iteratively estimated (based on previous estimation of T1) and corrected.

$$T1_{est}' = \frac{-T_{SR2}}{\ln\left(1-\frac{S_2}{S_1/\left(1-\exp(-T_{SR1}/T1_{est})\right)}\right)} \quad (5)$$

For example, if $T_{SR1}$ is twice the T1 of the material in question, then the initial estimate of T1 will be a function of $T_{SR2}$, shown in blue in Fig 2. Iteration causes convergence to the true T1, as shown in the figure.

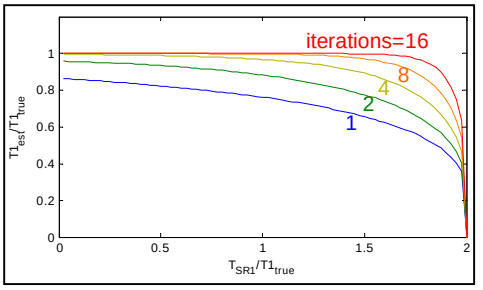


Figure 2: Iterative determination of T1

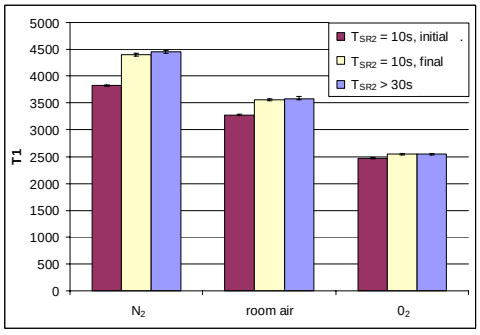


Figure 3: T1 measurements in vitro

Uncertainty in T1 ($\sigma_{T1}$) is related to the uncertainty (noise) in the source images, $\sigma_S$, as well as the parameters $T_{SR1}$ and $T_{SR2}$. Propagation of error and Monte Carlo methods were used to determine settings for $T_{SR1}$ and $T_{SR2}$ that produce the lowest uncertainty in T1 in the presence of noise. It was found that best results are obtained for $T_{SR2} \approx 0.6\ T1_{max}$ and $T_{SR1} \geq 2.5\ T1_{max}$, where $T1_{max}$ is the longest T1 expected to encounter. Given these settings, uncertainty in T1 in the range $0.4 - 1.0 \times T1_{max}$ should be below $4.5 \times$ the relative uncertainty of $S_1$.

Experiments were performed to validate the measurement technique *in vitro*. Distilled water with various $pO_2$ levels was prepared anaerobically by bubbling $N_2$, $O_2$, or room air through it. These samples were placed in a 37° C water bath and imaged at 1.5T with a GE Twinspeed EXCITE whole body scanner. Twelve pairs of SSFSE images were acquired in which $T_{SR2}$=2s and $T_{SR1}$=10s or >30s. ROI's were drawn to determine mean and standard deviation values for each sample, providing values of $S_1$, $S_2$ and $\sigma_S$. From these values, T1 and $\sigma_{T1}$ were estimated.

Volunteers were imaged to generate T1 maps of CSF *in vivo*. Various sections of the brain were examined. TE = 750ms was used to suppress signal from soft tissue and thus avoid partial-volume contamination.

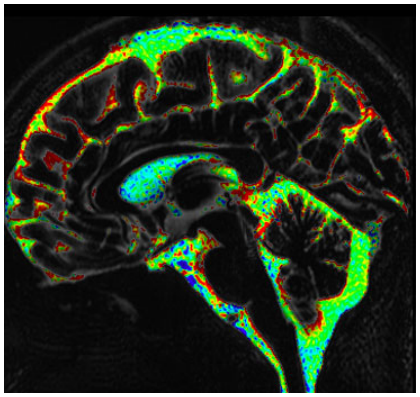

Figure 4: T1 mapping in vivo

***Results & Discussion***

Figure 3 demonstrates that oxygenation level of water strongly affects T1 relaxation time. It also shows that when the first acquisition is partially saturated, the initial T1 estimation is low, but it may be fully corrected by the iterative method presented, as predicted. Figure 4 shows a T1 map of CSF with blue representing T1=4.7 s ($pO_2$=0mmHg) and red representing T1=3.85 s ($pO_2$=194 mmHg), and the orange-yellow-green spectrum lying in between. Regional CSF oxygenation has never been measured before *in vivo*, but the mean values obtained are consistent with invasive measurements.

We have demonstrated a rapid method for measuring T1 by acquiring a pair of SSFSE images separated by a given saturation recovery period. By measuring T1 in fluids, $pO_2$ levels in these fluids may be estimated, providing information regarding arterial and tissue oxygenation, which may be altered in disease or during supplemental oxygen administration.

***References***

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