Quantification of in vivo T1 relaxation times in rat spinal cord at 17.6 Tesla

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Introduction

In vivo MR imaging methods provide the potential to improve information obtained by histopathology. Since MRI is limited by the signal-to-noise ratio (SNR) increasingly higher magnetic fields are used for rat spinal cord imaging. There is a clear signal gain with increasing field through an increase in the equilibrium magnetization, but this gain is moderated by changes in relaxation rates at higher field strengths. It has been shown that longitudinal tissue relaxation times – especially at fields below 1 T – depend strongly on the field strength [1]. But also beyond that range changes in relaxation times occur [2,3]. In this study, we have determined longitudinal relaxation times of rat spinal cord tissue at a much higher field strength and compared them to previously published values.

Objects and Methods

Animal preparation

Healthy adult female Fischer rats were anesthetized with 2% isoflurane with oxygen as respiratory gas and kept at body temperature while in the magnet using gradient temperature unit. Life functions of the rat were continuously observed during anesthesia. In order to reduce motion artefacts, the back of the rat was placed directly next to the surface coil to channel a large part of the breathing motion in the ventral direction, not affecting the spinal cord data.

MR imaging

All MR imaging experiments were conducted on a Bruker 750 MHz wide bore system (Bruker Biospin, Rheinstetten, Germany) at 17.6 Tesla with an animal gradient system with 200 mT/m and a 57 mm inner diameter. A homebuilt probehead and a linear surface coil were used to offer maximum space for the animal. For T1 quantification, an IR Snapshot FLASH technique [4] was used. Due to the inhomogeneous field of the surface coil we used an adiabatic pulse for inversion. Breath triggering was applied to start the inversion directly after a breathing period. The α angle of the Flash was adjusted so that the zero-crossing occurs before the next breathing motion starts.

Images were acquired in segments of 16 k-space lines per image per inversion, giving a total of 64 images with the largest inversion time slightly above 5 seconds. The in-plane resolution was 156x196 µm² and the slice thickness 1.5 mm. The repetition time between inversions was 15 s and the TE/TR for the snapshot FLASH sequence were 2.1/5.0 ms.

Fit algorithm

T1 maps were calculated using least squares optimization. In order to both improve the calculation time and get rid of noisy and therefore less useful data points, the fit routine was only performed on pixels above a certain signal threshold. After this fit procedure, all data points which had T1 values either higher than the maximum inversion time or below the inversion time of the 5th image were discarded as well, since the sequence parameters used are not optimized for T1 values outside these limits. To draw a line at the 5th image is arbitrary, but simulations showed that reliability of the fitting algorithm increases greatly with a higher number of points before the zero-crossing.

Simulation

In order to be able to evaluate the significance of the result in terms of SNR, a simple simulation was performed to check the signal change for a typical gradient echo sequence taking into account exclusively the change in longitudinal relaxation rates. For a broad range of repetition times from 4 ms up to 500 ms the Ernst Angle was calculated for each tissue at each relaxation rate and used to get the relative signal

$$S = \frac{(1 - e^{-TR/T_1}) \sin(\alpha_E)}{1 - \cos(\alpha_E) e^{-TR/T_1}}$$

Results

Because gray and white matter are usually not clearly distuingishable on T1 maps, a T2* weighted reference image with identical geometry was used to draw the regions of interest. There was no statistically significant difference between longitudinal relaxation times in gray and white matter. T1 values were 1664 ms and 1727 ms for white and gray matter, respectively. This is a substantial increase compared to previously published data at 2 Tesla. The fact that there is no significant difference between WM and GM longitudinal relaxation times was also observed at 2 Tesla.

Looking at the simulated values of the relative signal amplitude with respect only to longitudinal relaxation times (see Table 1), it can be seen that the signal gained from 2 to 17.6 T is reduced again by a little more than 20% due to increased T1, both in gray and white matter. Still, the larger part of the signal loss – espcecially for gray matter – occurs between 2 and 7 T and comparatively little between 7 and 17.6 T. The signal change was almost identical for all repetition times between 4 und 500 ms and therefore only one number per tissue and field strength was given.

	WM	GM	Ref.
2 T	1089 (0.97)	1021 (1)	[3]
7 T	1450 (0.84)	1650 (0.79)	[2]
17.6 T	1664 (0.78)	1727 (0.77)	

Table 1: Longitudinal relaxation times at different field strengths (simulated values for the relative signal amplitude in a gradient echo sequence are in brackets)

Discussion

Longitudinal relaxation times of the spinal cord of rats were obtained at 17.6 Tesla. T1 values are – as expected from theory – larger compared to published values at 2 and 7 T. As can be seen from Table 1, there is a strong increase in T1 values compared to those measured at 2 Tesla, but only a slight increase from 7 to 17.6 Tesla, which means that comparatively little signal is lost between 7 and 17.6 T due to increased T1 times.

Nevertheless, this provides only a rough estimate of the changes occuring, since T1 values depend on a variety of factors – beyond field strength – , such as temperature, respiratory gas, and the rat species. T1 values from [2,3] were determined on Sprague-Dawley rats, while Fischer rats were used in this study. Even though spinal cord structure is fairly similar across rat species, T1 values still might vary to some degree from one species to the other. Adult Sprague-Dawley rats could not be used in our study, because the maximum weight for a rat to fit inside the probehead is around 200 g.

As mentioned in the Methods section, spinal cord imaging is also affected by breathing motion. Up to now, the motion compensation schemes applied were breath triggering to assure that the inversion starts at the beginning of the resting state within the breathing cycle, and that the rat is properly arranged in the animal handling system so that the larger part of the breathing motion occurs in ventral direction. The precision of the measurements could probably still be improved by recording both the motion signal and the times when inversions are performed in order to filter all motion-corrupted images during post-processing.

References

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