# *In Vivo* Rat Spinal Cord Relaxation Times Measured at 4.7 T and 11.1 T

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## Introduction

The magnetic-field-strength dependence of NMR provides an important motivation for the development of high field magnets, since the increased signal strength can provide greater temporal, spatial, and/or spectral resolution. In addition, the frequency dispersion of relaxation times provides important information about dynamic processes in biological tissue. However, the field-strength dependence of NMR parameters (e.g. relaxation times) must be taken into account and may limit improvements in MR signal acquisition[1]. This is essential in studies of the vertebrate central nervous system, since MR provides a tool of critical importance for understanding health and disease, and is particularly significant in the spinal cord, since other methods are limited by penetration depth though bone or lack good soft tissue contrast. Therefore, the purpose of this work was to obtain data, at two additional field strengths, for *in vivo* rat spinal cord white matter (WM) and gray matter (GM) $T_1$ and $T_2$ relaxation times, and relative proton density values (PD), and explore the effects of field strength on the relaxation times.

## Methods

Rat spinal cord relaxation time measurements were performed on a 4.7 T horizontal bore magnet system and an 11.1 T horizontal bore magnet system. At both field strengths, a custom quadrature birdcage volume coil (diameter = 8.3 cm) was used for homogenous excitation of the sample and a custom quadrature surface coil was used to enhance the sensitivity of signal detection near the rat spinal cord. A total of 12 MR measurements were performed on 10 female Sprague-Dawley rats using approved experimental protocols and procedures. The spinous process of the T13 vertebral body was palpated and placed in the center of the surface coil with the rat in the supine position. $T_1$ and $T_2$ measurements were acquired with a spin echo sequence using axially-oriented slices with a 2 cm × 2 cm FOV, matrix size of 128 × 128 and 3 averages (4.7 T: slice thickness = 2mm, 11.1 T: slice thickness = 1mm). $T_1$ progressive saturation data was acquired with TR = 5000, 2000, 1000, 500 and 250 ms and TE = 15 ms. $T_2$-weighted images were acquired with TE = 75, 60, 45, 30 and 15 ms and TR = 2000 ms using a minimally-diffusion-weighted, spin echo sequence to eliminate (or reduce) image-gradient diffusion effects on $T_2$ measurements by limiting read gradient pulse separation. Regions of interest (ROI) were drawn to completely enclose the WM and GM regions (Fig 1.a & d) using a region growing algorithm and $T_1$ and $T_2$ were calculated using a saturation-recovery model and spin-echo decay model, respectively. A student's t-test for population differences with heteroscedastic variance was then used to compare the average $T_1$ and $T_2$ values for white matter and gray matter using R (The R Foundation for Statistical Computing).

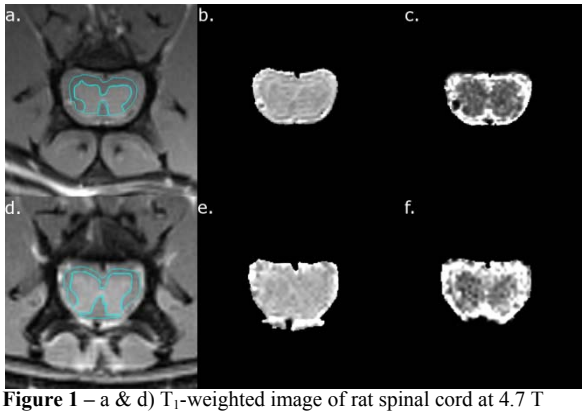


**Figure 1 –** a & d) $T_1$-weighted image of rat spinal cord at 4.7 T and 11.1 T, respectively (outlines indicate WM and GM ROI). b & e) $T_1$ maps at 4.7 T and 11.1 T. c & f) $T_2$ maps at 4.7 T and 11.1 T

## Results

$T_1$, $T_2$ and PD values calculated in the WM and GM structures at 4.7 T and 11.1 T are reported in Table 1. A two-tailed student's t-test revealed no statistically significant differences in WM and GM $T_1$ values at 4.7 T ($p = 0.218$) or 11.1 T ($p = 0.205$); however, statistically significant differences were seen at both field strengths when comparing $T_2$ ($p < 0.001$) and PD ($p < 0.001$) values. As shown in Figure 1 (b & e), *in vivo* $T_1$ maps, constructed by calculating $T_1$ on a pixel-by-pixel basis at both field strengths, show little contrast difference further suggesting that $T_1$ values in the WM and GM are similar. Alternatively, the calculated $T_2$ map shows clear differences in WM and GM at both field strengths (Fig 1.c & f).

**Table 1 –** Reported $T_1$, $T_2$ and relative PD in the rat spinal cord *in vivo*

| | | | $T_1$ (ms) | | $T_2$ (ms) | | PD (%) | |
|---|---|---|---|---|---|---|---|---|
| | Field | Level | WM | GM | WM | GM | WM | GM |
| (2) | 2 T | T7 | 1089 ± 126 | 1021 ± 93 | 79 ± 6.9 | 64 ± 3.4 | | |
| This Study | 4.7 T | T12-L4 | 1481 ± 115 | 1514 ± 99 | 57.3 ± 4.2 | 51.9 ± 1.3 | 40.0 ± 2.1 | 60.0 ± 2.3 |
| (3) | 7 T | C3 | | | 57.0 ± 1.6 | 43.2 ± 1.0 | 46.9 ± 1.5 | 53.1 ± 0.42 |
| This Study | 11.1 T | T12-L4 | 1581 ± 168 | 1620 ± 176 | 41.4 ± 6.3 | 34.5 ± 3.8 | 43.0 ± 2.0 | 57.0 ± 2.4 |

## Conclusions

At both 4.7 T and 11.1 T, differences in WM and GM $T_1$ relaxation times were not statistically significant; however, differences in $T_2$ relaxation times and relative PD values were found to be statistically significant at both field strengths. In addition, the frequency dispersion of $T_1$ relaxation values shows an increase in $T_1$ values with field strength, as expected for dipole-dipole relaxation[4]. However, the frequency dispersion of $T_2$ relaxation values decreasees with field strength. In order to compare these results with previously published data, $T_2$ values were measured with Hahn echoes, rather than a multiple-echo pulse sequence, so dynamic dephasing due to water diffusion in microscopic tissue susceptibility gradients contributes to a greater effective $T_2$ relaxation. These background susceptibility gradients would be expected to increase with field strength leading to shorter $T_2$ with increasing field strength.

## References

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