High resolution myelin water imaging in rat spinal cord in vivo with actively decoupled implanted RF coil

P. Kozlowski[1], J. Liu[2], J. C. Tso[1], N. Wolfe[1], and A. C. Yung[1]
[1]UBC MRI Research Centre, Vancouver, BC, Canada, [2]ICORD, Vancouver, BC, Canada

Introduction

Myelin water imaging has been successfully used to assess myelin content in human and animal CNS tissue [1,2]. It has been shown that myelin water fraction (MWF) [3] correlates well with the amount of myelin in a rat model of spinal cord injury (SCI) [2] and can therefore be an important marker of the cord pathology. Quantitative T_2 measurements used in myelin water imaging require very high SNR and homogeneity of the B_1 field, making it challenging to apply this technique in rat spinal cord in vivo. Although the use of a small surface coil implanted surgically next to the cord significantly improves SNR [4], the B_1 field inhomogeneity remains a problem. We have designed and built an RF implanted coil, which is actively decoupled from a volume coil. Here we present preliminary results of high spatial resolution myelin water imaging using this system.

Methods

Two Sprague-Dawley rats with rectangular copper coils (11x19 mm size) implanted over the T9/T10 spine (as described in [4]) were imaged in the prone position in a 7T animal scanner (Bruker, Germany). A Bruker quadrature birdcage coil was used for transmission, while a 3.4cm diameter pickup surface coil with diode-activated decoupling LC-tank circuit was used in combination with the implant for reception (schematic is shown in Figure 1). The mutual inductance between implant and pickup produces two "overcoupled" resonances, with the lower peak tuned to the Larmor frequency during reception. During transmission, the pickup coil is deactivated with a PIN diode, which converts the implant's frequency response from two overcoupled peaks to a single resonant peak far away from the Larmor frequency, in order to minimize coupling with the active birdcage coil.

Myelin water measurements were carried out using a single slice, multi-echo CPMG sequence [5] (256x256 matrix, TR/TE=1500/6.673ms, NA=6, 32 echoes). A 1.5mm axial slice was selected through the thoracic spine (T10) and an FOV of 3cm was used resulting in in-plane resolution of 117µm. T_2 distributions were calculated from the multi-echo data using non-negative least squares analysis [3]. MWF maps were generated by integrating the 7.75-20 ms range and dividing by the total integral of the T_2 distribution in each pixel.

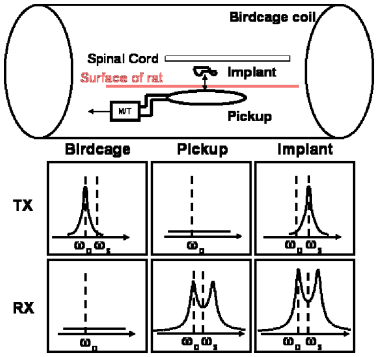

Fig. 1: System schematic

Results and Discussion

Figure 2 shows the first echoes of the CPMG data (left) and the reconstructed MWF maps (right) for the two rats. Figure 3 shows T_2 distributions from the ROIs encompassing WM (top) and GM (bottom) calculated from CPMG data acquired from rat #1 (red) and rat #2 (blue). Myelin water fraction was 0.22 and 0.25 in WM and 0 and 0.04 in GM for rat #1 and rat #2 respectively.

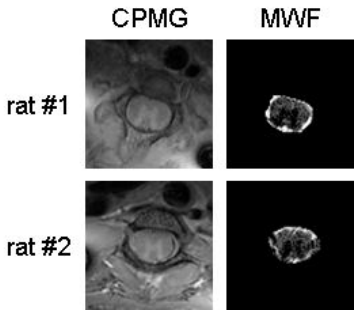

Fig. 2. First echo of the CPMG data (left) and the reconstructed MWF maps (right) acquired from two rat spinal cords in vivo. The in-plane resolution is 117 µm and the slice thickness is 1.5 mm.

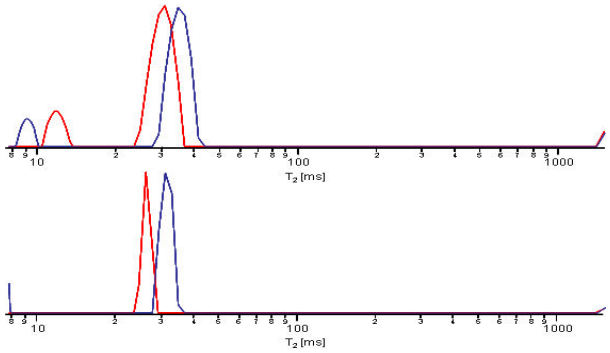

Fig. 3. T_2 distributions in ROIs encompassing WM (top) and GM (bottom) calculated from CPMG data acquired from rat #1 (red) and rat #2 (blue). Myelin water fraction was 0.22 and 0.25 in WM and 0 and 0.04 in GM for rat #1 and rat #2 respectively.

The reconstructed MWF maps from the rat spinal cords show excellent WM/GM contrast corresponding to the differences in myelin content. The MWF values in WM and GM correspond well with previously reported data and expected amounts of myelin within the rat spinal cord [6]. The T_2 distributions shown in Figure 3 demonstrate good consistency of the data acquired from two rats, although a study involving larger number of animals is needed to confirm this. The differences in T_2 values of the short and medium T2 components between the two data sets are not surprising, since Whittall et al. describe computer simulations that indicate that MWF is usually centered on the true value for noisy CPMG decay curves, whereas the T_2 times are more difficult to estimate by the NNLS algorithm [3].

Conclusions

In this pilot study we have shown that high resolution myelin water mapping in rat spinal cord in vivo is feasible. The MWF maps show details of the cord morphology, and the average MWF values in WM and GM correspond well with previously published results and the expected amounts of myelin within the cord.

Acknowledgments

This study has been supported by the Canadian Institutes of Health Research and the Natural Sciences and Engineering Research Council of Canada.

References: [1] Laule C, et al. Mult Scler, 2006, **12,** 747; [2] Kozlowski P, et al. J Neurotrauma, 2008, **25**, 653; [3] Whittall KP, et al. Magn Reson Med, 1997, **37**, 34; [4] Yung AC, et al. Magn Reson Imaging, 2007, **25**, 1215; [5] Poon CS, et al. J Magn Reson Imaging, 1992, **2**, 541; [6] Kozlowski P, et al. Magn Reson Med, 2008, **59**, 796.