Myelin water fraction using multiple-echo 2D and 3D GRE at 3T with whole-brain coverage

Ana-Maria Oros-Peusquens[1], Sandra M Meyers[2], Alex L MacKay[2], and N. Jon Shah[1,3]
[1]*INM-4, Research Centre Jülich, Jülich, Germany,* [2]*Department of Physics and Astronomy, University of British Columbia, Vancouver, B.C., Canada,* [3]*Faculty of Medicine, JARA, RWTH Aachen University, Aachen, Germany*

Introduction It is increasingly recognised that myelin is ubiquitous in determining MR contrast in the living brain, especially at high fields [1,2]. Investigating its distribution is not only useful in the study of several neurodegenerative diseases but can also be used for the *in vivo* parcellation of the brain. Obtaining quantitative information about myelin content *in vivo* is, however, a difficult and not fully solved task. The method which comes closest to a 'gold standard' for quantification of myelin content *in vivo* is based on the detection and quantification of the properties of the myelin water pool, well-established in conjunction with an NNLS-based decomposition of multi-time point T_2 decay curves [3]. Due to SAR constraints, which limit the applicability of RF-intensive sequences, and also due to the richness of the T_2^* contrast, the imaging method of choice at high fields is based on gradient echo (GRE). Combining T_2^* decay information from multiple-echo GRE scans with the analysis method established on T_2 decay curves appears thus ideally suited to the study of myelin water content at high fields. A few initial results using GRE have been reported: T_2^* information obtained from a multiple-echo 3D GRE acquisition was investigated with NNLS at 1.5T [4] and 2D GRE acquisitions at 3T were combined with either a 3-exponental fit [5] or with a bi-exponential decomposition including phase offsets [6] (also applied at 7T). We report here on a 12-volunteer study of whole-brain myelin water, where T_2^* decay curves were obtained with both 2D and 3D multiple-echo GRE (meGRE) sequences and analysed with NNLS.

Materials and methods All volunteers were scanned following approval by the local ethics committee and after giving informed consent. Results using the 2D meGRE were obtained from twelve healthy volunteers (6 females and 6 males, mean age 30y, from 19 to 44) scanned on a 3T TIM-Trio Siemens scanner equipped with gradient of 40mT/m, body coil transmit and a 32-element phased-array receive coil. The parameters of the experimental 2D meGRE protocol included: FOV=192x162 mm^2, 25 2.5mm thick slices with 1.25mm gap, matrix size=192x162, TR=2200ms, flip angle=90deg, BW=898 Hz/px, 32 echoes, TE1=3.24ms, ΔTE=1.54ms, 2 or 4 averages (acquisition time = 11 or 22 min). In addition, results using the 3D meGRE were obtained from 2 female volunteers (25 and 27y). The 3D meGRE protocol used: FOV 220x165x202 mm^3, matrix 196x128x144, TR=61ms, flip angle = 28deg, BW=797 Hz/px, 30 echoes, TE1=1ms, ΔTE=2ms, 4 acquisitions (11 min/acquisition). The data were evaluated using Matlab (The Mathworks Inc) according to the procedure described in [7], with no a priori assumptions about the number of contributing T_2^* components. No image filters were used.

Results and Discussion Example T_2^* distributions are shown in Fig.1. Two components can be readily recognised, one centred at ~60ms, the other one at ~10ms. For comparison, a monoexponential fit to the (either 2D or 3D) data leads to a distribution of T_2^* values centred around 50ms. The peaks were attributed to myelin water (short T_2^*) and tissue water (long T_2^*). The decay curves were not sampled to TE values long enough to allow the identification of a "CSF" peak.

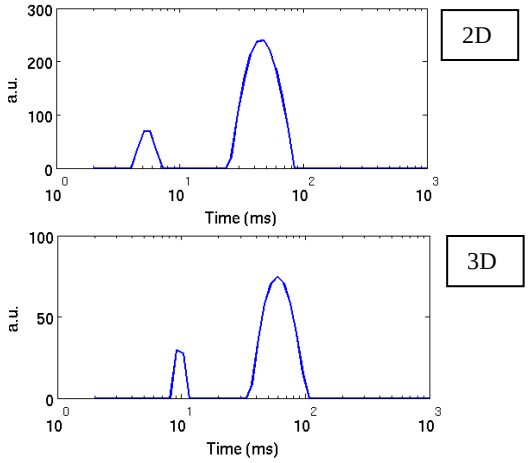

Figure 1. Example T_2^* distributions for 2D (top) and 3D (bottom) data sets

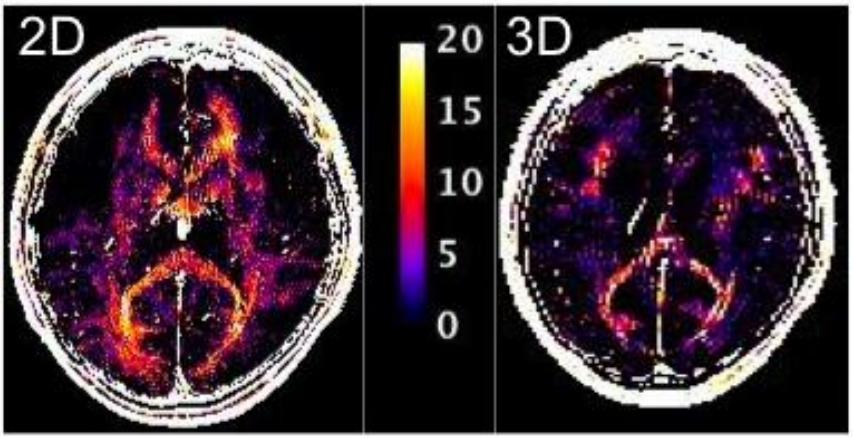

Figure 2. Myelin water fraction maps from 2D multi-slice (left) or 3D (right) acquisitions.

MWF maps are shown in Fig. 2. Results obtained with the 2D method are compared to results obtained for the same volunteer and a similar slice with the 3D method. The average value of white matter MWF obtained with the 2D method over all 12 volunteers amounts to 6.4(0.9)% (only non-zero voxels were included). The values are comparable to those reported with 3D GRE at 1.5T [4] but lower than the values obtained with the spin-echo method [3]. The anterior part of the brain (Fig. 2) shows lower MWF values on average, most likely due to the effect of field inhomogeneities on the decay curves. For the moment, these effects were not accounted for, but work is in progress to improve the processing method. For a test case, correction of the decay curves for first-order field inhomogeneities (sinc modulation in the 2D acquisition [8]) substantially improved the quality of the maps in the frontal regions (data not shown). In the 3D case the correction is more difficult to describe analytically and can be circumvented by limiting the number of echoes included in the fit upon consideration of the measured values of background field gradients. A stable NNLS analysis of the decay curves requires very high initial SNR. The high read-out efficiency of the GRE method, combined with a moderately high field strength, is capable of delivering adequate SNR with full brain coverage in a clinically acceptable measurement time (e.g. 11 min for the 2D method). If image filtering is used to reduce the effect of noise, a 2-average or even a single-scan acquisition for the 3D data becomes acceptable (22 min or 11 min, respectively). Several methods for noise suppression are being currently investigated. We estimate that the method could be used at very high fields (7T and above) if high-resolution imaging and advanced shimming are employed, both reducing the effect of field inhomogeneities on the decay curves.

References [1] Li TQ et al., Magn Reson Med., 62 (2009); [2] S. Geyer et al., Front. Hum. Neurosci., 5 (2011); [3] A. MacKay et al., Magn. Reson. Med., 31 (1994); [4] C. Lenz et al, Magn. Reson. Med., 68 (2011); [5] D. Hwang et al. NeuroImage, 52 (2010); [6] P. van Gelderen et al., Magn Reson Med., 67 (2012); [7] K.P. Whittall and A.L. MacKay. J. Magn. Reson., 84 (1989); [8] D.A. Yablonskiy, Magn. Reson. Med. 39 (1998); [9] C. Labadie et al., proc. ISMRM 2009, p. 3210.