Optimisation of 3 T and 7 T T_2^*-weighted MRI for the detection of small parenchymal veins in MS lesions

J. E. Dixon[1], M. J. Brookes[1], E. C. Tallantyre[2], N. Evangelou[2], and P. G. Morris[1]
[1]Sir Peter Mansfield Magnetic Resonance Centre, University of Nottingham, Nottingham, Nottinghamshire, United Kingdom, [2]Department of Clinical Neurology, Nottingham University Hospital NHS Trust, Nottingham, Nottinghamshire, United Kingdom

INTRODUCTION White-matter (WM) MS lesions are visible on MRI scans at conventional field strength and so MR is often used to diagnose and monitor patient progression. However, the poor correlation between MR findings and patient symptoms highlights the need to advance the study of these lesions *in vivo*. In particular, there is considerable interest in the heterogeneity of WM lesions. Post-mortem studies of WM lesions have shown a close spatial relationship to parenchymal blood vessels[1] and, in previous work, we have shown that this may be a distinguishing factor between MS and ischaemic lesions[2]. However, the study of this *in vivo* is limited at conventional field strength by the difficulty in demonstrating both lesions and small vessels on one MR image. Recently, we have shown that the decrease in T_2^* at ultra-high field (7T) allows us to acquire T_2^*-weighted images with enhanced susceptibility contrast and high SNR compared to conventional field strength, aiding the detection of these small blood vessels[2]. In order to obtain an accurate picture of the relationship between vessels and lesions, it is essential to determine and implement optimum scanning parameters for the detection of these vessels. It is also important to know the minimum size of a vessel that can be detected using T_2^*-weighted imaging. Because the magnitude of the susceptibility effect around veins is orientation dependent, there is inherent bias towards detection of veins in certain planes. In this work, the susceptibility effect is simulated in order to predict the optimal TE and voxel shapes (by changing slice thickness and in-plane resolution, keeping voxel volume constant) at 3 T and 7 T, to increase the sensitivity of vessel detection and to reduce this bias.

THEORY Modelling a vessel as a cylinder of infinite length with radius r_v, the change in resonant frequency for a spin inside a vessel oriented at θ^c to the B_0 field can be calculated by: $d\omega = (dX/2)\omega_0(\cos^2\theta - 1/3)$. For a spin outside the vessel, at a position r,Φ from the centre of the vessel, the change in frequency will be given by: $d\omega = (dX/2)\omega_0(r_v/r)^2\sin^2\theta\cos(2\varphi)$. The total signal within a voxel containing a vessel can then be calculated by summing the contributions from all individual spins: $S_v = (S_0/N)\Sigma\exp(id\omega TE)\exp(-TE/T_2^*)$ whereas the signal from a voxel that does not contain a vessel can be simply calculated by: $S_n = S_0\exp(-TE/T_{2\,WM}^*)$. For a vessel to be considered detectable, we have assumed that the difference in signal caused by the vessel must be greater than twice the standard deviation of the noise. This gives a threshold of $(S_n - S_v)/S_n > 2/SNR$.

METHODS 3T images were acquired using a Philips Acheiva 3.0 T MR system with a whole-body gradient set, whole-body transmit coil and 8-channel SENSE rf receive coil. Scanning at 7T was performed using a Philips Acheiva 7.0 T system with whole-body gradient set, head-only transmit coil and NovaMedical 16-channel SENSE rf receive coil. T_2^* images were acquired using a 3D gradient-echo acquisition, with flip angle 14° over a 192x164 mm^2 FOV in 4 stacks, each with 50 slices at 7T and 32 slices at 3T, using a 150-ms TR. SNR was first measured at both field strengths using 1-mm isotropic voxels, and then using 0.8-mm isotropic at 3T and 0.5-mm isotropic at 7T, as these were the minimum size to give reasonable SNR in acceptable imaging time. The predicted signal was calculated over the voxel volume using the theory outlined above. SNR in WM was measured at both 3T and 7T by running the sequence with a second dynamic with no RF excitation in order to obtain a noise-only image. The signal in a selected ROI was then divided by the standard deviation over the corresponding ROI in the noise image. The standard deviation of the noise in the phase images was estimated to be the inverse of the SNR in the magnitude image[3]. Susceptibility-weighted images were created and simulated by applying a high-pass filter to the phase information which was then raised to the power 4 before being multiplied by the magnitude image[3]. The contrast-to-noise ratio (CNR) between a voxel containing a vessel and a voxel containing WM only was calculated according to Haacke et al[3] and the vessel was deemed to be detectable if the CNR was greater than 2.

RESULTS The SNR in images acquired with 1-mm isotropic voxels was between 1.5 and 3 times higher at 7 T than 3 T, over the range of TEs tested. The predicted smallest vessel detectable in 3 T and 7 T images for a range of TEs and slice thicknesses is shown in Figure 1, as an average of the prediction across a range of vessel orientations (0 – 90° to B_o, in steps of 10°) along with the standard deviation across orientations.

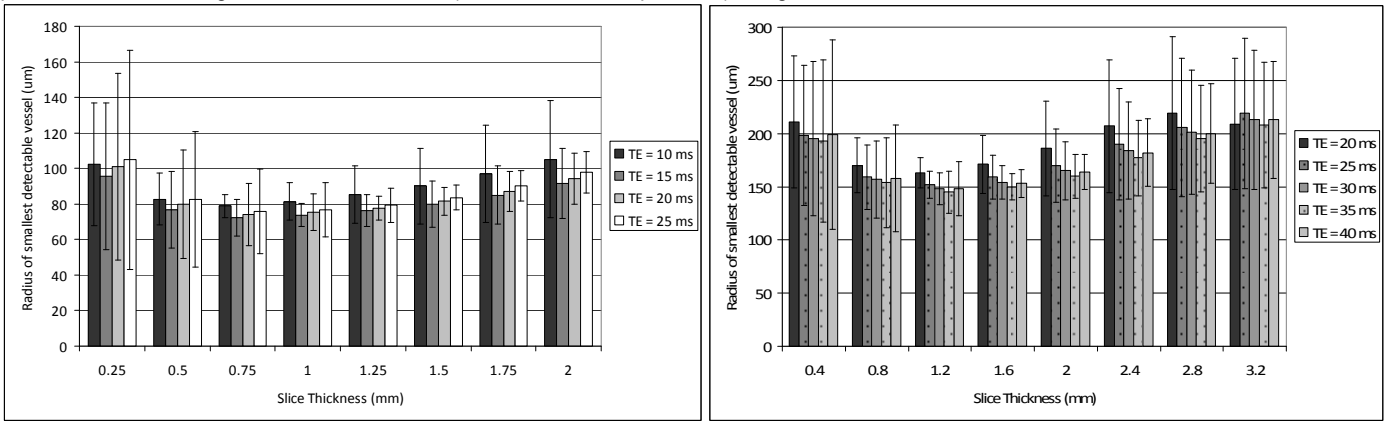

***Figure 1** – Radius of smallest detectable vessel in 7 T (left) and 3 T (right) T_2^*-weighted magnitude images, over a range of slice thicknesses and echo times [note the different scale].*

DISCUSSION Simulations showed that, using a 25-ms TE, veins could be detected at 7 T with areas as small as 0.007 mm^2; however, this is highly orientation dependent, as vessels perpendicular to these will be detected only if they have areas larger than 0.062 mm^2. The smallest standard deviation in vessel radius over the full range of vessel orientations is achieved using a 15-ms TE with 0.35 x 0.35 x 1.00 mm^3 voxels. This allows detection of veins with 0.017-mm^2 cross-sectional areas (average of predictions across angles). At 3 T, the sensitivity to veins was highest using a 35-ms TE. The slice thickness giving highest sensitivity was 1.2-mm, though dependence on vessel orientation was reduced using 1.6-mm slices, giving a cross-sectional area of the smallest detectable vessel of 0.071 mm^2. This is more than 4 times larger than the veins detectable at 7 T.

CONCLUSION T_2^*-weighted imaging allows detection of small parenchymal veins; simulations predict that veins with cross-sectional areas as small as 0.071 mm^2 can be detected at 3 T, with little dependence on vessel orientation. At 7 T, this can be reduced to 0.017 mm^2, suggesting that ultra-high-field MRI may offer significant benefit in the further study of multiple sclerosis.

REFERENCES 1. Lucchinetti et al, Ann Neurol 2000; 47: 707-17. 2. Tallantyre et al, Neurology 2008; 39: 1682-92.

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