Optimised Acquisition of Magnetic Field Correlation Mapping for Improved Precision

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Introduction

Several different MR measures have been proposed to quantify brain iron such as transverse relaxation rates, phase, magnetic susceptibility and recently magnetic field correlation mapping (MFC) [1]. MFC is defined as a temporal correlation function $MFC(t) = \gamma^2 <B(t)B(0)>$, where γ is the gyromagnetic ratio, B(t) is the difference between the actual magnetic field experienced by a water molecule and B_0 at time, t [2]. MFC is believed to have a more direct relationship with magnetic field inhomogeneities (MFIs) than R_2' [1], and may therefore be more sensitive to microscopic MFIs produced by iron deposits. MFC is measured by acquiring data using an asymmetric spin-echo with echo shifts of the order of several ms, using a fixed read out window, and fitting measured signal intensities to the equation: $S(t_s)=S(0)exp(-2t_s^2MFC)$ [2], where t_s is the time-shift of the 180° pulse and S(0) is the signal intensity measured at t_s=0ms (i.e. a symmetric spin-echo). Reported protocols to date typically collect data at equally spaced time points, t_i where i=1 to n different time shifts, with the number of repeat acquisitions at each time shift, N_i equal for all i. Based on optimisation of other quantitative techniques [3][4], we investigated if by varying parameters t_i and n (for a fixed value of ΣN_i) the precision of MFC could be improved, using simulations, and tested the results *in vivo*.

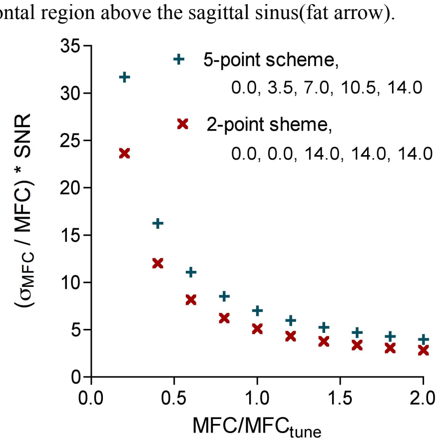
Methods Simulations: The variance in MFC, σ^2_{MFC}, was estimated using a similar method to that proposed by Fleysher et al [4]. A "design matrix" [5]A was constructed, with columns equal to the partial derivatives of the function wrt. to the fitted parameters (S(0) and MFC), and rows 1 to n:

$$A_{i,j} = \frac{1}{\sigma_0}\begin{pmatrix} e^{-2t_1^2 MFC} & -2t_1^2 S(0) e^{-2t_1^2 MFC} \\ \vdots & \vdots \\ e^{-2t_n^2 MFC} & -2t_n^2 S(0) e^{-2t_n^2 MFC} \end{pmatrix}$$

where σ_0 is the noise in the image data, and was estimated from acquired data. σ^2_{MFC} can then computed from the covariance matrix, $(A^+ A)^{-1}$. The "optimum" acquisition is defined as the set of timing parameters that minimises, σ^2_{MFC} for a particular value of the parameter of interest (MFC) for which the acquisition is 'tuned' [4]; here a value of 500s^{-2} (typical in iron rich regions of interest) was used for "MFC_{tune}". The originally published method used n=5 and N_i=10 for all n, with t_s equally spaced between 0 and ~16ms [1]. For simulations, we investigated n≤5 (so the maximum n matched previous reports) and ΣN_i=5 (since changes to this simply scale the overall error), and all possible values of t_2 to t_5 between 0 and 14ms (t_1=0ms for all configurations, and t_s=14ms is the maximum possible shift on our scanner given constraints discussed below), in step sizes of 0.5ms, resulting in over 700x10^3 different timing configurations. After the minimum value of σ^2_{MFC} was found for MFC_{tune}, σ^2_{MFC} was than calculated for a set of MFC values ranging from 0.2 to 2.0MFC_{tune}. Imaging: A symmetric and asymmetric spin-echo sequence was implemented on our Signa HDx 3T MRI (General Electric, Milwaukee, WI) scanner. Data was collected with time-shifts approximating those reported of t_s=0,-3.5,-7.0,-10.5 and -14.0ms with N=10 for each shift, and a second scan with the "optimum" shifts and repeats as determined by simulation. All other acquisitions parameters were the same for both scans, including: TR/TE=1500/48.7ms, FoV=25.6x25.6cm, matrix=128x128, #shots=4, rBw = 1421 Hz/pixel, slice thickness/gap=2/2mm, #slices=15. Processing: All volumes were registered to the mean non-shifted (t_s=0ms) image using a 3D rigid body registration. All repeats were used to fit the signal equation above, using the Levenberg-Marquardt algorithm [6] to produce S(0) and MFC maps. "Error" maps were generated from the square root of the variance found from the covariance matrix [6].

Results

The minimum value of σ^2_{MFC} from all possible timing configurations simulated was found to be 0, 0, 14, 14, 14ms, i.e. n=2 with N_1=2 and N_2=3. Figure 1 shows the normalised coefficient of variation ((σ_{MFC}/MFC)*SNR) [4] for both the "published", 5-point scheme and for the optimised, 2-point method. For all values of MFC/MFC_{tune} (equivalent to MFC=100 to 1000 s^{-2}) there is a decreased coefficient of variation (improved precision) for the 2-point scheme compared to the 5-point scheme (fig.1). Figure 2 shows S(0), MFC and error maps for the 5-point, and 2-point acquisitions. S(0) and MFC values measured are very similar between both acquisition schemes. The error is reduced for the two-point scheme, most noticeably in the regions of high MFC such as the globus pallidus (thin arrow) and orbital-medial-frontal region above the sagittal sinus(fat arrow).



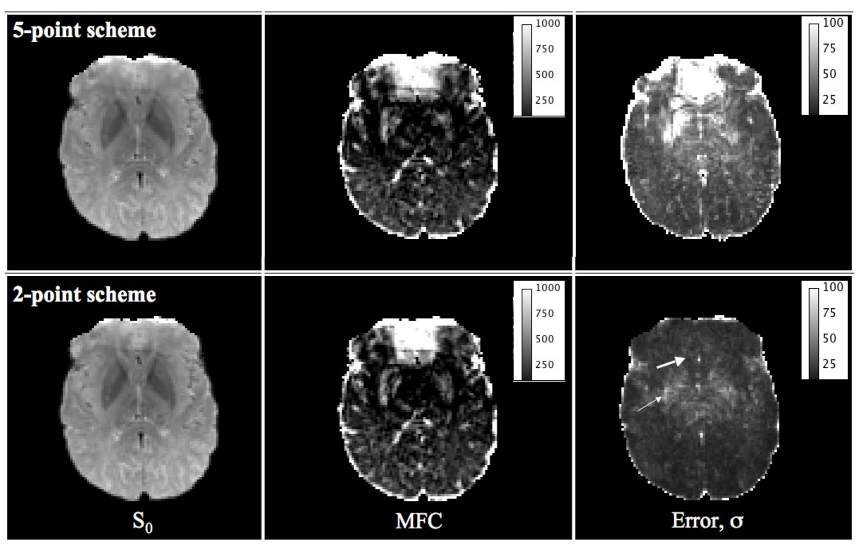


Figure 1 (above) Simulation results of coefficient of variation for the 2-point and 5-point scheme over a range of MFC values. Figure 2 (right). Maps of fitted parameters of S(0), MFC and MFC error term.

Discussion and Conclusion

Other simulations with a longer maximum shift (20ms) resulted in further reduction of the variance, confirming that MFC is more precise at longer shifts [2]. However the maximum time shift, whilst keeping the centre of the shifted echo within the acquisition window, is dictated by receiver bandwidth, and echo train length (ETL). The maximum bandwidth was dictated by the echo time required to match other published protocols (since MFC varies with TE [1]) and the ETL dictated by the number of shots to reduce distortion artefacts sufficiently. Implemented on our scanner, the maximum time shift was 14ms, for parameters above. This work indicates that for a given maximum time shift and number of repeats, acquiring MFC data with the "two-point" method improves the precision of the MFC estimate. Another advantage of the two-point scheme is that an analytical solution can be found, allowing faster processing times.

Acknowledgements: PhD Funding for C. Mallik from NIHR Biomedical Research Centre at SLAM and IoP KCL.

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