Measurement of Field Inhomogeneity and Susceptibility Effects for Liver Iron Quantification in Patients with Iron Overload

B. A. Taylor[1], R. B. Loeffler[1], R. Song[1], R. J. Stafford[2], B. McCarville[1], J. S. Hankins[3], and C. M. Hillenbrand[1]
[1]Radiological Sciences, St. Jude Children's Research Hospital, Memphis, TN, United States, [2]Imaging Physics, The University of Texas M. D. Anderson Cancer Center, Houston, TX, United States, [3]Hematology, St. Jude Children's Research Hospital, Memphis, TN, United States

Introduction: The toxicity associated with iron overload requires monitoring and management of iron content in visceral organs of patients receiving chronic blood transfusions [1]. Noninvasive MR techniques are available that calibrate the T_2* (or R_2*) to liver iron content (LIC). This is typically done using a monoexponential fit of data from a multi-gradient echo acquisition [2-4]. However quantification in these studies could potentially be compromised by local susceptibility changes and lipid contamination. Recently, an autoregressive moving average (ARMA) model has been introduced for rapid calculation of T_2* and chemical shift from a limited number of sparsely sampled echoes [5, 6]. ARMA has the potential of providing robust liver iron quantification in the presence of field inhomogeneities and off-resonances. In this work, we compared traditional T_2* mapping of a biopsy calibrated patient cohort to ARMA modeling for T_2* calculation.

Methods: Fifty patients with iron overload were scanned in accordance to two Institutional Review Board-approved protocols. All patients were scanned on 1.5T whole body scanners (Siemens, Malvern, PA). Using a phased-array body coil, a single breath-hold technique was used to obtain magnitude multi gradient echo axial images at the level of the main portal vein with a repetition time of 200 ms. Echo times ranged from 1.1 ms to 17.3 ms at 0.8 ms increments (echo train length=20). Other imaging parameters were: flip angle 25º, slice thickness 10 mm, in-plane resolution 3.125 mm^2, matrix size 128 × (96–128) and field of view 380 mm × (285 mm–380 mm). Total acquisition time for the 20 images ranged from 18 s to 25 s. The T_2* maps were calculated in two ways: (a) Truncated monoexponential fitting using a least-square method [3]; (b) ARMA modeling, which can extract T_2* values of multiple chemical species [5,6]. In four patients of this cohort magnitude and phase images were acquired, thus providing input for ARMA to also calculate the chemical shift by assuming a linear combination of complex exponentials with noise. This algorithm characterizes the exponential signal as a rational polynomial in the *z*-domain via the z-transform. The poles of the polynomial correspond to the field and T2* of each chemical specimen (i.e. water, lipid) in the signal. This was used to create T_2* maps that took field inhomogeneities and susceptibility into account. Identical ROIs from an area of the liver void of vessels were used to calculate mean T_2* values in all patients for both techniques.

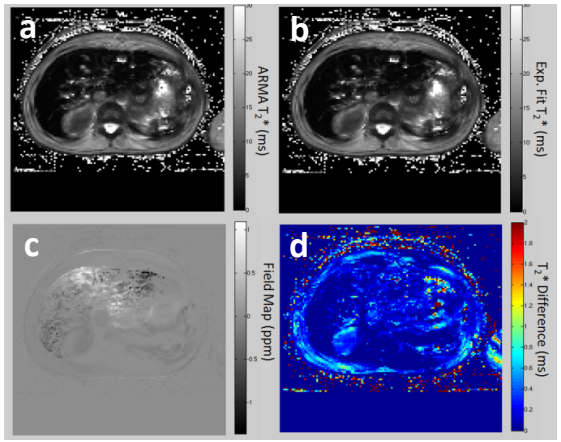

Figure 1: T_2* maps from the ARMA model (a) and monoexponential fitting (b). The field map from the ARMA model (c) shows increased off-resonance in the liver. Even in areas with high off-resonance the difference in T_2* between the two methods was relatively small (d).

Results: In the four patients, where field inhomogeneity was also calculated, the difference in T_2* between the monoexponential fit and the ARMA model was 0.033 ± 0.043 ms. Figure 1 shows the patient with the largest susceptibility difference in the liver (~0.65 ppm), i.e. worst case in our patient study. Even with this large susceptibility the difference in T_2* values (Fig. 1d) from the ARMA model (Fig. 1a) and traditional fitting (Fig. 1b) was less than 0.6 ms in the liver. Figure 2 shows the correlation between each method for T_2* measurements in the large set of patients (n=46) that contained only magnitude information. Without the phase images, the field inhomogeneity could not be calculated with high accuracy. The slope of the regression line was 1.05 with an intercept of -0.25 (R^2=0.9982), indicating very high correlation. These same patients showed high correlation to liver biopsies obtained as part of the study [3].

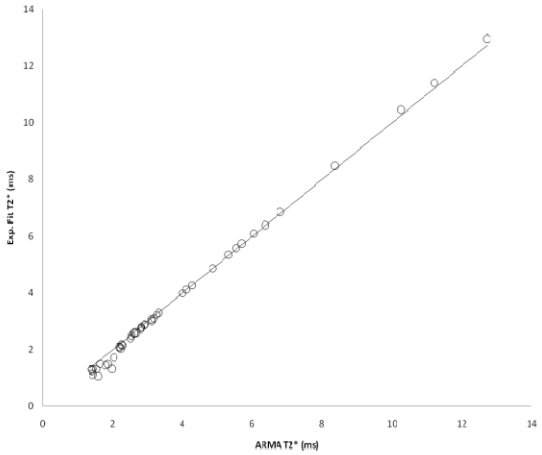

Figure 2: Correlation between T_2* measurements from the ARMA model and T_2* values used in calibration for LIC. The slope is near unity at 1.05 with an intercept of -0.25 (R^2=0.9982).

Discussion: Very similar T_2* estimates were obtained with 2 independent processing approaches (Fig. 2), which increases the trust in both methods. The low T_2* values in tissue with high iron content will also reduce SNR and, thereby, the precision of the spectral estimates in ARMA. This can partly explain the increased noise in the field map at the liver in figure 1. Lipid in the liver was not detected in this patient population consisting mainly of children with sickle cell disease, but the ARMA technique can also take lipid into account to calculate separate water and lipid T_2* values [5,6]. This will make the ARMA technique very useful if patients do present with fatty liver disease. Overall, our data suggest that field inhomogenities do not appear to negatively impact iron quantification in the liver. Although we could not calibrate the ARMA T_2* values directly to liver biopsies with field inhomogeneity estimation in all 46 cases, we are confident that it can be cross-calibrated with existing calibration measurements given the very high correlation to monoexponential T_2* measurements.

References: [1] Taher, A.T., K.M. Musallam, and A. Inati., Hemoglobin, 2009. 33 Suppl 1: p. S46-57. [2] Anderson, L.J., et al., Eur Heart J, 2001. 22(23): p. 2171-9. [3] Hankins, J.S., et al., Blood, 2009. 113(20): p. 4853-5. [4] Wood, J.C., et al., Blood, 2005. 106(4): p. 1460-5. [5] Taylor, B.A., et al., Med Phys, 2008. 35(2): p. 793-803. [6] Taylor, B.A., et al., Med Phys, 2009. 36(3): p. 753-764.