Reference Phantom Validation of T2-mapping: Maximum Likelihood Estimation of T2 from Magnitude Phased-Array Multi-Echo Data

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Introduction: Quantitative T2/T2* measurements are being studied for a number of applications, e.g. cartilage degeneration, iron-load in liver, tumor vascularization, and SPIO-labeled cell tracking. In a typical T2-mapping experiment, multi-echo data is acquired and T2 is then obtained by a pixel-wise fit of an exponential model to the magnitude of signal as a function of echo time. For low SNR input data, T2 fits are biased upward [1,2] because rectified noise on later echoes mimics signal that persists with increasing TE, and hence long T2. Explicitly including noise statistics in a maximum likelihood (ML) estimation of T2 potentially mitigates dependence of T2 values on SNR [3]. Purpose of this work is to validate the use of ML estimation of T2 from multi-echo data using a reference phantom. Specifically, this paper reports on accuracy and precision of T2-fitting to data acquired using phased-array coils and using parallel imaging.

Theory: For single channel coils, noise distribution in magnitude images is Rician and its variance can be estimated from a signal-less region in the image background, this then allows reduction of the noise bias in the T2 fit, by subtracting the noise power [1]. In phased-array coils, element combination with sum of squares and its variants leads to non-central chi noise statistics that may be inhomogeneous over the image [4] and hence harder to compensate for in the T2 fit. The SENSE algorithm preserves the Rician noise distribution, which is inhomogeneous noise levels over the image [2,5]. For phased array coils, potential noise bias can be countered by reducing the weight of the later echoes in the fit: weighted least squares (WLSQ).

ML methods allow for accurate and precise estimation of relaxation times [6]. Given a correct modeling of the signal evolution and of the noise statistics, ML estimators are asymptotically unbiased and reach the minimal variance level (Cramer-Rao lower bound). In ML estimation, the model parameters that maximize the likelihood of the data given the measurement noise statistics are computed. For T2 mapping from a series of N spin echo magnitude images y_k acquired at echo times TE_k, this amounts to solve pixel-wise the optimization problem [3]:

$$\min_{(\rho_0,T_2)} \prod_{k\le N} \frac{2y_k}{\sigma^2}\exp\left(-\frac{y_k^2+s_k^2(\rho_0,T_2)}{\sigma^2}\right) I_0\left(\frac{y_k s_k(\rho_0,T_2)}{\sigma}\right) \text{ (1) with } s_k(\rho_0,T_2)=\rho_0\, e^{-TE_k/T_2} \text{ (2),}$$

where ρ_0 is the spin echo signal at TE=0, σ is the local noise standard deviation, and I_0 is the modified Bessel function of the first kind with order 0. This non-linear minimization problem can be efficiently solved by means of Newton type algorithms such as Levenberg-Marquardt by applying a quadratic approximation to Eq. (1) as proposed in [6]. Correction of noise bias by means of the ML Eq. (1) requires accurate knowledge of σ. In this work, it was estimated on the basis of noise calibration data acquired in k-space and propagation with the linear operators involved in the image reconstruction process.

Methods: Experiments were performed using a reference phantom with 12 vials with various T1/T2 values, (Eurospin TO5, Diagnostic Sonar, Livingston, Scotland). Multi-slice multi-echo data were acquired on a 1.5-T clinical scanner using an 8-channel head coil, with the coil in quadrature mode or in phased-array mode, respectively. SENSE acceleration factors of 2, and 3 were tested. Scan parameters were: field-of-view 220 mm, 0.5 mm pixel resolution and 3 mm slices. Eight or 32 echoes were acquired at 10 ms intervals with TR = 1500 ms. T2 was estimated using ML and WLSQ. For each vial, mean T2 and coefficient of variation were calculated using 180mm^2 ROIs. Temperature MR room was 296 K. Accuracy was expressed as a relative error: 100% * (ROI mean T2 – Reference T2)/Reference T2.

Results: For all samples, accuracy of T2 estimation using ML was well within 5% of the reference value for quadrature reception, SENSE reconstruction, and SENSE with an acceleration factor of 2 (Fig. 1a). WLSQ mean T2 was significantly biased upward, especially for the long T2 values. The higher SNR of phased array signal reception leads to a consistently lower variance on the T2 (Fig. 1b). ML shows a significant reduction in variance going from 8 to 32 echoes, especially for the long T2 samples. This reduction is absent for WLSQ. This disparity demonstrates how the additional information of the later echoes is wasted by the weighting that serves to reduce the effect of noise in WLSQ. Finally, ML variance was consistently lower than WLSQ variance, consistent with ML approaching the Cramer-Rao bound.

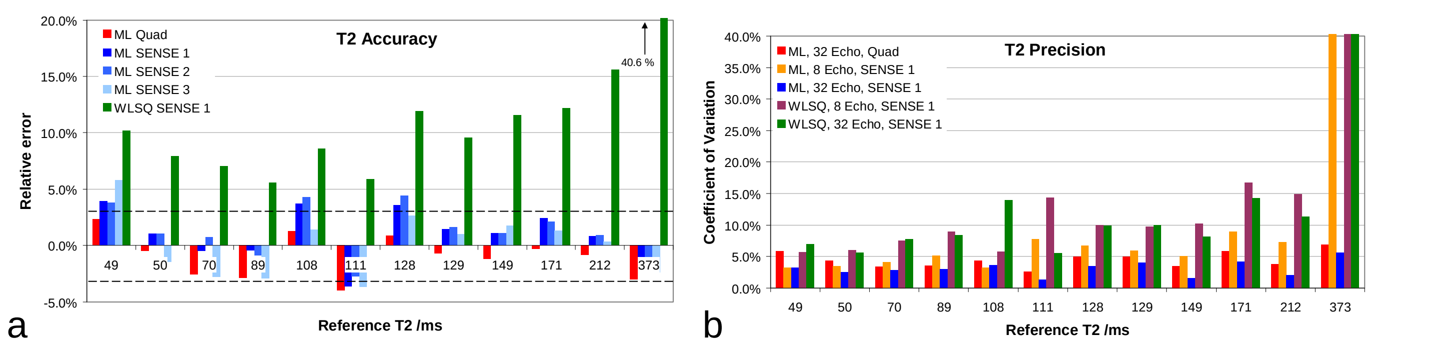

Figure 1: Accuracy of T2 measurement for ML estimation with quadrature reception or SENSE reconstruction and WLSQ estimation (a). Dashed lines indicate 3% tolerances on the refer ence phantom T2. Comparison of variance for ML with quadrature reception, and ML and WLSQ estimation with 8- and 32-echo acquisition (b).

Discussion and conclusion: ML estimation of T2 from phased-array magnitude data including explicit noise statistics was validated on a reference phantom and was found to give accurate T2 figures. Previous approaches to deal with noise bias depend on a preliminary T2 range estimate and reject the information of later echoes, e.g. in WLSQ. ML here has the potential to extend the dynamic range of an experiment to shorter and longer T2s. The noise power subtraction method by Miller et al is limited to use with single channel coils [1]; phased-arrays, however, are currently the mainstay of imaging coils. Having a means to deal with the noise characteristics of phased-array coils can benefit T2 mapping by improving the SNR of the multi-echo data and thus reducing the variance of the T2 maps. With ML estimation, T2 values remained accurate when employing parallel imaging, with the potential of reducing the sometimes prohibitively long examination times for these methods, such as the 11 minute T2-mapping sequence in the OsteoArthritis Initiative protocol for knee cartilage. Although the estimation problem is similar for T2* imaging, this was not part of the current work, for the lack of a recognized reference phantom.

In conclusion, ML estimation has utility for accurate and precise pixel-wise calculation of T2.

References: [1] Miller et al, MRI, 11:1051-1056 (1993). [2] Graves et al, JMRI; 28:278-281 (2008). [3] Sénégas et al, ISMRM, # 1782 (2007). [4] Roemer et al, MRM, 16:192-225 (1990). [5] Prüsmann et al, MRM; 42:952–962 (1999). [6] Karlsen et al, MRM, 41: 614-623 (2006).