Precision in T1-mapping and estimation of quality maps

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Introduction: T1-mapping in the myocardium may be used to detect both focal and diffuse disease processes that result in an elevation of T1. The sensitivity for detecting abnormal elevation of T1 is limited by the precision of T1 estimates which is a function of the number and timing of measurements along the T1-recovery curve, the signal-to-noise ratio (SNR), the tissue T1, and the method of fitting. Quantifying the statistical variation of measured T1 could improve confidence in detecting the significance of results. It is proposed to produce a map calibrated in T1 units that represented the quality of the T1 estimate, by transforming the standard deviation of residual fitting error into the standard deviation of the estimate parameters. Estimate of the T1 parameter error based on the fit residuals is derived analytically, and validated on phantom measurements. In-vivo examples demonstrate the potential utility.

Methods: T1-mapping using inversion recovery approaches such as MOLLI are based on performing a 3-parameter fit at each pixel to measurements of the inversion recovery at multiple inversion times. The 3-parameter model $y_i = A - B \exp(-t_i/T1^*) = A - B \exp(-t_i (B/A-1)/T1)$, where the apparent time constant $T1^* \approx T1/(B/A - 1)$ represents the Look-Locker correction for the influence of the readout. The chi-square (or LSE) cost function may be written as:

$$\chi^2(\vec{\alpha}) = \sum_{i=0}^{N-1}\left(\frac{y_i - y(x_i|\vec{\alpha})}{\sigma_i}\right)^2$$

where $y_i = y(t_i) = 0,\ldots,N-1$ are the measurements (image values for i-th inversion time at each pixel), x_i are the independent variables, σ_i is the standard error when measuring the data point y_i assumed to be normal for the PSIR reconstruction [1], and $\vec{\alpha} = [\alpha_0, \alpha_1,\ldots\alpha_{M-1}]$ are the parameters to be estimated (i.e., A, B, T1). The covariance matrix **C** of the estimated parameters may be approximated [2] as $\mathbf{D}^{-1}$ with elements

$$D(k,l) = \left[\sum_{i=0}^{N-1}\frac{1}{\sigma_i^2}\left[\frac{\partial y(x_i|\vec{\alpha})}{\partial \alpha_k}\frac{\partial y(x_i|\vec{\alpha})}{\partial \alpha_l}\right]\right]_{kl}$$

where the partial derivatives are derived analytically and evaluated at each inversion time. Second order derivatives have been dropped in this formulation. As a result of the PSIR reconstruction, the σ_i for each measurement may be assumed to be equal and are estimated at each pixel from the fit residuals using a median absolute deviation approach. At each pixel the estimated standard deviation in the T1 is calculated producing a confidence map.

A Monte-Carlo simulation using N=65,536 trials was used to compute the standard deviation in T1 as a function of SNR, and T1 for a specific MOLLI protocol (5-3 sampling, TImin=105 ms, TI shift = 80 ms), and was compared with the estimate of standard deviation based on the proposed approach using the fit residuals. Experimental validation was performed by repeated measurements of a set of T1-measurements comparing estimated and calculated standard deviations. T1-maps and corresponding estimated standard deviation (SD) maps are provided for in-vivo examples. SNR maps [3] are calculated from the signal intensities of the longest inversion time image which has achieved steady state. The estimated image SNR and T1 SD may be compared with the Monte-Carlo simulation (Fig 1).

Results: The measured and calculated estimates for T1 standard deviation are graphed (Fig. 1) for T1 in the range 400-1600 ms and SNR = 20, 30, and 40. An example in-vivo T1-map for normal subject (Fig. 2) illustrates the variation in T1 estimation error with surface coil intensity variation, and is in close correspondence to predicted performance (Fig. 1). The anteroseptal ROI has SNR=32.1, T1=1012ms, SD=25.0ms, whereas the lateral wall ROI has lower SNR=20.9, T1=1026ms, SD=41.8ms (increased). An example T1-map for a subject with myocarditis (Fig. 3) has a focal elevation of 103 ms (1098 vs 995) and SD= 41 ms, i.e., focal elevation $\approx$ 2.5 SD indicating statistical significance (P<0.02).
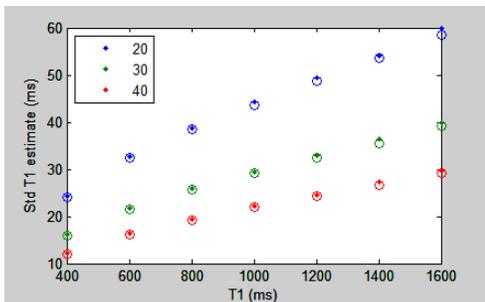
Figure 1. Measured (dots) vs calculated (circles) standard deviation vs T1 for various image SNRs from Monte-Carlo simulation with 65,536 trials.
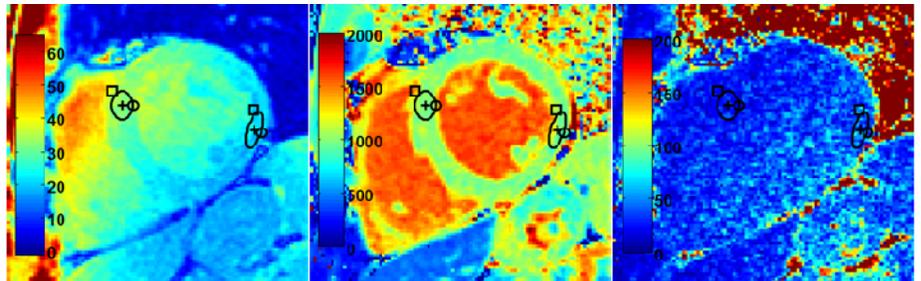
Figure 2. SNR map (left), T1-map (center), and T1-std map (right). Note the increased T1 standard deviation of T1 at the lateral wall corresponds to decreased SNR resulting from drop off of surface coil sensitivity.

Discussion and Conclusions: The formulation for scaling the fit residuals into the desired parameter error (namely T1 standard deviation) may be used to provide confidence maps for gauging the statistical significance of abnormally elevated T1. This formulation is validated for PSIR reconstructed images although it is not readily translated to magnitude IR T1-mapping due to multi-fitting as well as the Rician noise distribution in magnitude signal. This formulation may also be used to calculate the standard deviation of R1=1/T1 and by extension be used to generate SD maps for extracellular volume (ECV) fraction maps calculated as the change in R1 with contrast calibrated by measurement of hematocrit.

References: [1] Xue, H et al. Magn Reson Med, 2012 Jun 26. doi: 10.1002/mrm.24385. [2] Alper JS, et al, J Phy Chem 1990. 94, 4747-51. [3] Kellman P, et al. Magn Reson Med. 2005 Dec; 54:1439–1447.
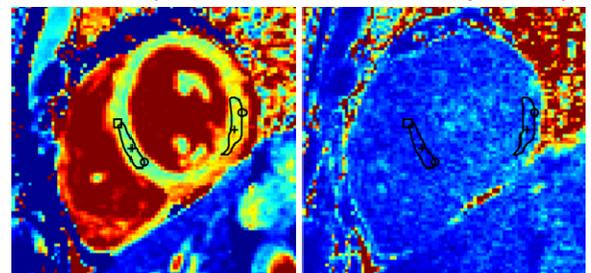
Figure 3. Example T1-map and SD map for subject with myocarditis and focal abnormality corresponding to T1 elevation (103 ms) between lateral wall and septum of 2.5 SD (lateral wall SD = 43 ms).