An estimation method for improved reconstruction of MR signal parameters in unilateral scanner

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Introduction

Unilateral NMR devices are a valuable tool in various applications such as non-destructive testing or well logging, but are not applied routinely for biomedical imaging. A low cost, unilateral and portable MRI can introduce MRI into clinical applications previously non-feasible, exploiting the advantages of the open architecture, the portability and affordable costs of these scanners. However, the inhomogeneity of B_0 in these scanners [1] results in a poor signal to noise ratio (SNR), and forces the use of the relatively slow scan-schemes and long averaging times. Improving the SNR of scans with Unilateral NMR is therefore a key factor as it can reduce the total scan time (by reducing the need in averaging repetitions). This may allow the implementation of additional methods to accelerate imaging.

We suggest a novel post-processing method to improve the SNR of the acquired signal in unilateral NMR scanners. We estimate the signal parameters from the noisy data with the weighted least square (LS) approach, and exploit more efficiently the inherently known characteristics of the NMR signal. The method was first developed and tested for T2 measurements with a CPMG-like sequence applied in a unilateral scanner. Then, using a similar concept we further developed this method to improve the SNR of lateral slice–selective imaging scans specific to the unilateral scanner.

Methods

Assume a CPMG measurement with N sampled points per echo and with M generated echoes. The acquired data from the real and the imaginary channel can be described as follows:

$$\begin{bmatrix} Y_R \\ Y_I \end{bmatrix}_{2N\times M} \approx \begin{bmatrix} X_r(\underline{\theta},\underline{a}) \\ X_i(\underline{\theta},\underline{a}) \end{bmatrix}_{2N\times M} + \begin{bmatrix} v_R \\ v_I \end{bmatrix}_{2N\times M}$$

where $\underline{\theta}$ is the set of parameters defining the echo shape, $\underline{a}$ is the vector of maximum echoes amplitudes, and v_R, v_I are the noise from the real and imaginary channels, respectively. The parameters that are estimated from the signal are derived by minimization of the following expression:

$$J(\underline{\theta},\underline{a}) = [Y - \underline{h}\cdot\underline{a}]^T C^{-1}[Y - \underline{h}\cdot\underline{a}], \quad \left(Y = \begin{bmatrix} Y_R \\ Y_I \end{bmatrix}_{2N\times M}, \ \underline{h} = \begin{bmatrix} \underline{h_r}(\underline{\theta},\underline{t}) = -\cos(\phi(t_n - t_0))\cdot\exp(-(t_n - t_0)^2/2\sigma^2), \quad n = 1,2,..,N \\ \underline{h_i}(\underline{\theta},\underline{t}) = \sin(\phi(t_n - t_0))\cdot\exp(-(t_n - t_0)^2/2\sigma^2) \end{bmatrix}_{2N\times 1} \right)$$

The echo shape is fitted to a Gaussian model (dictated by the RF coil reception properties) which is appropriate for structures that are approximately uniform in the longitudinal axis. C is the covariance matrix of v_R and v_I, which is extracted from noise measurements that are acquired in between excitations periods. By solving the above non-linear weighted LS problem the signal parameters can be estimated yielding an improved reconstruction of the echoes.

In slice–selective imaging scans, spatial encoding is performed by phase-encoding gradient in the lateral directions (x,y) [2]. However, since a constant gradient in the longitudinal direction (z) is present, the acquired echo is also frequency encoded during acquisition. Due to the very thin thickness of the excited slices (a few hundreds of microns) we assume that images of biological tissues can be approximated as having a depth variance that is independent of the planar variation, i.e. we can model the signal from the tissue with two separate components: $I(x,y,z) = I(x,y)\cdot I(z)$, resulting in:

$$S(k_x,k_y,k_z(n)) = \sum_{x=0}^{M-1}\sum_{y=0}^{L-1} I(x,y)e^{-2\pi i(k_x x/M + k_y y/L)} \sum_{z=0}^{N-1} I(z)Ke^{-2\pi i(k_z(n)z/N)} = S(k_x,k_y)\cdot S(k_z(n)) \quad n = 1,2,..,N$$

The vector $S(k_z)$ therefore remains constant for every phase encoding step, and can be evaluated independently in a single short measurement. The complex number $S(k_x,k_y) = S_r + iS_i$ can then be estimated for every phase encoding step, by minimizing the following expression:

$$J(S,\underline{a}) = [Y - H\cdot\begin{pmatrix} S_r \\ S_i \end{pmatrix}\cdot\underline{a}]^T C^{-1}[Y - H\cdot\begin{pmatrix} S_r \\ S_i \end{pmatrix}\cdot\underline{a}], \quad \left(Y = \begin{bmatrix} Y_R \\ Y_I \end{bmatrix}_{2N\times M}, \ H = \begin{pmatrix} \underline{h_r} & -\underline{h_i} \\ \underline{h_i} & \underline{h_r} \end{pmatrix}_{2N\times 2} \right)$$

Results & Conclusions

We use the NMR-MOUSE (ACT, Germany) unilateral scanner. The LS algorithm was applied to estimate the signal parameters from samples of Glycerol and Paraffin excited with CPMG-like sequence. Both the estimated and the straightforward-calculated decaying curves were compared to gold standard measurements (1000 averages) with the MSE criteria (Fig. 1, Table 1). To evaluate the potential of the lateral imaging LS estimation method, a simulation was performed with different white noise levels; the estimation method showed a significantly smaller MSE (with respect to the original image) compared to the conventional image reconstruction (about 23% in the example shown in Fig.2). Results indicate the potential of post-processing estimation for SNR improvement in Unilateral NMR, aiding the use of such devices in bio-medical applications.

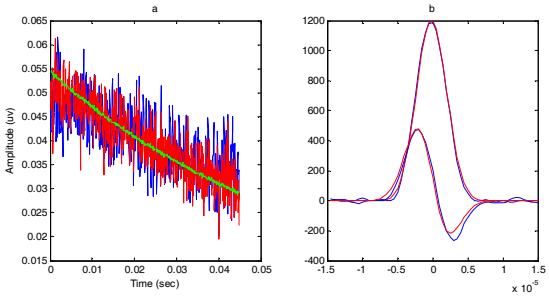

Fig. 1: **(a)** The estimated decay curve (red) from a CPMG measurement is compared to a not-estimated (blue) and gold standard (green) curves. **(b)** The estimated echo shape (red) versus the average of all echoes (blue).

Sample	Averages	Error Improvement
Glycerol	None	46%
Glycerol	4	43.2%
Glycerol	8	39.3%
Paraffin	8	42.5%
Paraffin	16	55%

$$Error\ Improvement = 100 - \frac{\text{MSE(estimated)}}{\text{MSE(not estimated)}}\cdot 100$$

Table 1: Comparison of the MSE of the estimated and the not-estimated (raw) decay curves from CPMG measurements. Errors are relative to the gold-standard measurements.

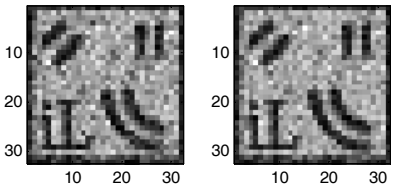

Fig. 2: Comparison of image reconstruction, conventional (left) versus LS algorithm (right) reconstruction. The MSE of the LS reconstructed image is lower by 23%.

References [1] Blumich, B. *et al*. Prog. Nuc. Mag. Res. Spec.2008. [2] Perlo, J. *et al*. JMR 2004.