Estimation of Coil Sensitivities in Myocardial First-Pass Perfusion Imaging Using a Model-Based T1 Mapping Technique

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Target Audience: Researchers and clinicians working in the field of quantitative myocardial perfusion imaging and cardiac MRI.

Purpose: To correct for the influence of coil sensitivities in quantitative myocardial first-pass perfusion imaging without any additional scans.

Introduction: Quantification of myocardial perfusion using dynamic contrast-enhanced MRI requires the concentration of the contrast agent (CA) in the arterial input function (AIF) as well as the myocardial tissue enhancement curves to be known for each temporal frame [1]. This is typically achieved by an estimation of the T_1 relaxation time for each time frame. As an exact determination of T_1 for each heartbeat is not feasible using available T_1 mapping techniques, T_1 is typically derived from the time-signal intensity curve $S(t)$, which, for short recovery times or low CA concentrations, can be assumed to be linearly proportional to the relaxivity $R_1 = 1/T_1$. A major difficulty is that the acquired signal $S(t)$ typically consists of the MR signal $M(t)$ superimposed by the sensitivity ξ of the receiver coil, introducing errors in the estimation of the CA concentration and therefore also in the calculation of the perfusion values. The influence of the coil sensitivities can be corrected based on additionally acquired proton density images or pre- and post-contrast T_1 maps [2]. In this work, a recently proposed model-based approach for T_1 mapping will be used to estimate a coil sensitivity map ξ and correct signal variations due the coil sensitivities without additional scans.

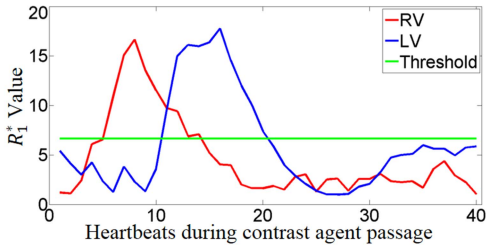

Fig. 1. ROIs of R_1^* in RV (red) and LV (blue), threshold corresponding to 150ms (green).

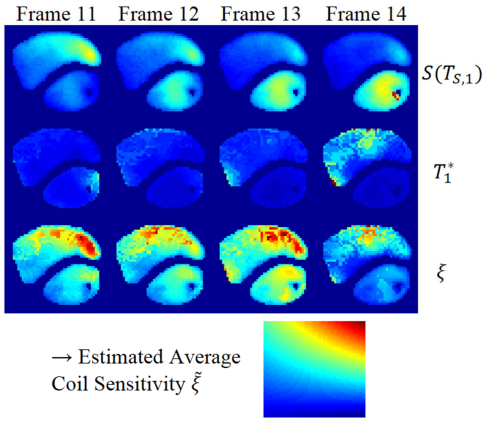

Fig. 2. Steps of the coil sensitivity estimation.

Theory: As described in [3], the MAP algorithm enables the model-based reconstruction of saturation recovery (SR) prepared FLASH acquisitions, which can be modeled by [4]:

$$S(t)\begin{cases} = \rho\xi \cdot \left(1 - e^{-t/T_1^*}\right) & \text{for } t \geq T_1^* \\ \approx t \cdot \rho\xi / T_1^* & \text{for } t \ll T_1^* \end{cases} . \quad [1]$$

Here, ρ is the spin density, ξ the coil sensitivity and T_1^* the longitudinal relaxation parameter T_1 modified by the continuous RF excitation of the FLASH acquisition [5]. For $t \ll T_1^*$, a Taylor expansion leads to a linear relation (bottom of Eq. 1) and a distinction between $S_{0,\xi}$ and T_1^* is not possible. If the recovery time T_S is, however, in the range of the relaxation time T_1^*, the MAP reconstruction yields T_1^* as well as the product $\rho\xi$ of spin density and coil sensitivity. With T_1^* known, the average coil sensitivity can be estimated as:

$$\xi \propto S(T_S)/\left(1 - e^{-T_S/T_1^*}\right) . \quad [2]$$

Methods: An ECG-gated first-pass perfusion experiment was performed on a 3T whole-body scanner (MAGNETOM Trio, Siemens AG, Sector Healthcare, Germany). A contrast agent (Gadovist®, Bayer HealthCare AG, Germany) bolus of 4ml, followed by a saline flush of 20ml was intravenously injected in a healthy volunteer at a flow rate of 4 ml/s. Subsequently, radial SR FLASH acquisitions (FOV: 230×230×10mm^3, T_E: 1.54ms, T_R: 3.49ms, α: 12°, projections: 60, readout points per projection: 128, breath-held) were applied over 40 consecutive heartbeats. After data collection, a MAP reconstruction (100 iterations) was applied to obtain T_1^* in each voxel. Next, the time course of T_1^* was observed in 3×3 ROIs in the left (LV) and the right ventricle (RV) to select a set of heartbeats where the condition $T_1^* \leq 150$ms was fulfilled (green dashed line in Fig. 1). Using T_1^* and $S(T_{S,1})$ for the selected heartbeats in Eq. 2, ξ was estimated, and areas with $T_1^* \leq 150$ms were masked. Last, a polynomial with 2 variables and degree 2 was fitted to each of these ξ maps [6]. After normalizing each of the resulting surfaces to a maximum value of 1, their average $\bar{\xi}$ was used for the correction of the coil sensitivity according to $\tilde{S}(t) = S(t)/\bar{\xi}$.

Results & Discussion: Fig. 1 shows the evolution of R_1^* in regions of interest in the LV and RV. The values only fall below the T_1^* threshold of 150ms during the passage of the CA through either of the ventricles. Below the threshold, the temporal coverage of the relaxation curve is insufficiently short compared to the order of T_1^* and only the product $\rho\xi/T_1^*$ can be determined, leading to an unreliable T_1^*. Fig. 2 shows the steps involved in the estimation of the average coil sensitivity $\bar{\xi}$. Fig. 3 depicts the average contrast of heartbeat 11 before and after the ξ correction (a), a cross section (b, indicated by the green line in Fig. 3a) and time-signal curves of 3 myocardial ROIs (c, indicated by different symbols in Fig. 3a). Especially the levelling of the LV signal and the ROIs further away from the coil array (circles & pluses) underline the effective reduction of coil sensitivity effects by the correction.

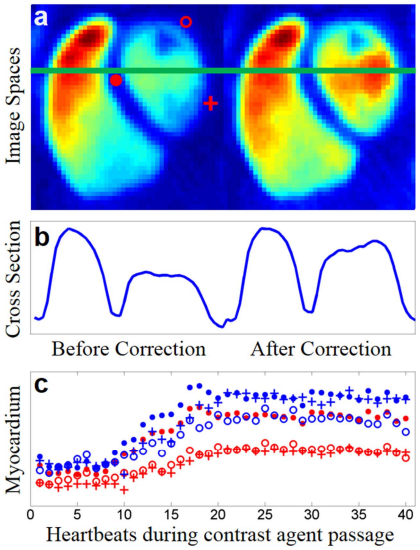

Fig. 3. a: Image space of heartbeat 11. b: Cross section (green line in a). c: Signal-time curves in myocardial ROIs before (red) and after (blue) the correction.

Conclusion: This study shows that in areas of high contrast agent dose such as LV and RV in a myocardial first-pass perfusion measurement, the longitudinal relaxation parameter T_1^* can be determined even for the extremely short acquisition time of about 200ms. With T_1^* known, the average coil sensitivity $\bar{\xi}$ can be estimated without any additional scans, and coil sensitivity effects in the perfusion image series can be largely reduced.

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References: [1] Gerber et al., J Cardiov Magn Reson 10:18 (2008), [2] Cernicanu et al., Acad Radiol 13:686-693 (2006), [3] Tran-Gia et al., Magn Recon Med 70:1524-1534 (2013), [4] Kholmovski et al., Magn Reson Med 57:821-827 (2007), [5] Deichmann et al., J Magn Reson 96:608-612 (1992), [6] http://www.weizmann.ac.il/home/eofek/matlab/.