Single-shot Look Locker T1 mapping with golden angle radial sampling for free-breathing liver DCE-MRI in rats at 1.5 T

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Introduction: In dynamic contrast enhanced (DCE) MRI, the temporal behavior of the MR signal following contrast agent (CA) injection, provides insight into a wide range of physiologic parameters in tumor diagnosis and therapy monitoring. E.g. the blood flow or perfusion, endothelial permeability and extra-vascular extra-cellular space (EES) can be estimated by means of dedicated pharmacokinetic modelling of the CA concentration over time. The latter can be estimated from the temporal behaviour of the longitudinal relaxation rate R1 by applying the fast exchange limit regime ($R_1 = R_{10} + r_1 \cdot C_{Gd}$) [1]. Currently used techniques that estimate R1 from a series of T1-weighted images, acquired with 3D gradient echo sequences, are prone to displacement errors, motion and flow artefacts as well as to inhomogeneous excitation ($B1^+$) fields [2]. An alternative approach to reduce these effects is to dynamically quantify the R_1 relaxation rates directly. For this purpose, a 2D single-shot Look-Locker T_1 mapping technique (LLRGC) was proposed that applies golden angle radial sampling in combination with a contrast enhancing k-space filter [3]. In the present study, we evaluate this technique for non-triggered and free breathing DCE-MRI experiments in rats.

Methods: All phantom and DCE-MRI experiments were performed on a 1.5T clinical MRI scanner (1.5T Philips Achieva, the Netherlands). To evaluate the accuracy of the LLRGC technique, a gel phantom, containing samples of different T1 (40-1900ms), was imaged with a circular surface coil (receive only) and different flip angles (FA: 4°,6°,8°,10°). The results were compared to reference values obtained with an IR-TSE sequence (TR/TE=2000/6.7s, inversion times T_i=50-1600 ms). The T1 variability, FA dependency, and motion sensitivity of the proposed technique were further analyzed by quantitative abdominal imaging in four healthy rats. Moreover, DCE-MRI measurements using a 4-channel wrist array coil were performed in the same animals. All experiments were approved by the local ethics committee. A double-dose (0.2*mmol/kg*) of Gd-DTPA was bolus-injected (≤ 1s) and image acquisition was performed during free breathing. The imaging slice was positioned in the axial plane covering liver and back muscle. T_1 maps were acquired every 6s during bolus arrival and first-pass (3min) and every 24s during washout (12min) as follows: after non-selective adiabatic inversion, 225 radial profiles were acquired during the first 2.6s of the recovery cycle (TR/TE=11.6/5.2 *ms*, FA=10°, matrix =100x100, FOV=80x80mm^2). From these data, 22 T_1-weighted images were reconstructed by k-space filtering, from which the T_1^* map were computed applying a three parameter, non-linear least-squares fit using $A+B \cdot \exp(-T_R/T_1^*)$. True T_1 values were obtained by $T_1=T_1^* \cdot (A/B-1)$ [4]. The T_1-time curves were converted to C_{Gd}-time curves using a Gd-DTPA relaxivity of r_1=4.1mM^{-1}. To prove the modelling capability of C_{Gd} curves, Ktrans and v_e were obtained by one-compartment modelling based on the Kety-Schmidt equation [5]. A population-based arterial input function (AIF), derived from DCE-CT data of 11 rats was used. The summed relative squared residuals (SRSR) were used to provide goodness of fit values.

Results: The Phantom T1 values deviate from the IR-TSE reference values by a maximum of less than 10% (for T_1 range of 300-1800 ms) with a negligible FA dependency (see table). Fig. 1 depicts representative results of the FA dependency of the mean T_1 values in different tissue ROIs of one animal. As indicated by the error bars, the T1 variability significantly improves with increasing flip angle ($p<0.05$) while the mean T_1 remains stable. Fig. 2 shows the magnitude image (T_i = 1.8s) and the color-coded T_1 map 2min post Gd-DTPA injection. Beside some typical radial streak artefacts and blurring, no impairing breathing or motion artefacts are observed in the magnitude image and the T1 map. Fig. 3 depicts the C_{Gd}-time curves for ROIs in the liver (black) and muscle (grey) from all 4 animals. All curves were successfully fitted with high accuracy (SRSR, muscle: 5±1% and liver: 15±11%) yielding the listed K^{trans} [min^{-1}] and v_e values. Muscle values agree with the very sparse data available in literature [6].

Mean (stddev) phantom T1 values [ms]				
IR-TSE	LLRGC 10°	LLRGC 8°	LLRGC 6°	LLRGC 4°
172	148 (25)	142 (32)	133 (41)	128 (51)
387	394 (20)	390 (28)	387 (36)	388 (46)
638	574 (49)	579 (48)	579 (47)	585 (55)
960	890 (49)	903 (50)	925 (55)	936 (73)
1344	1311 (94)	1326 (96)	1345 (105)	1355 (152)
1835	1694 (177)	1714 (188)	1732 (198)	1727 (242)

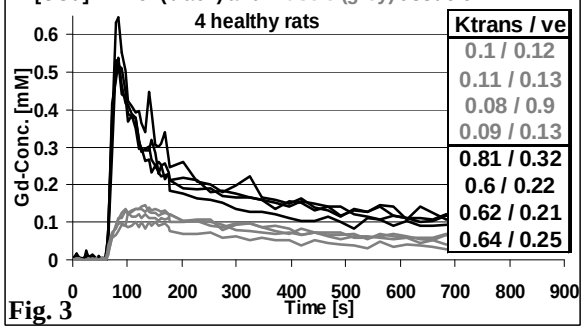


Fig. 3

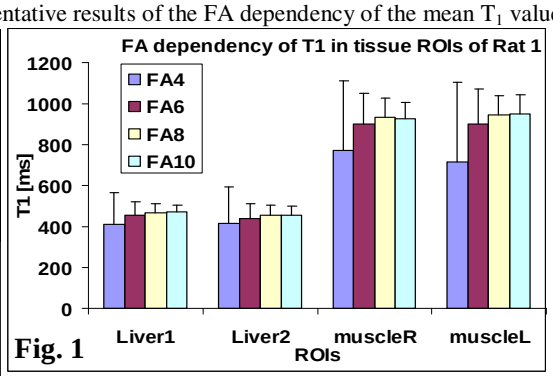
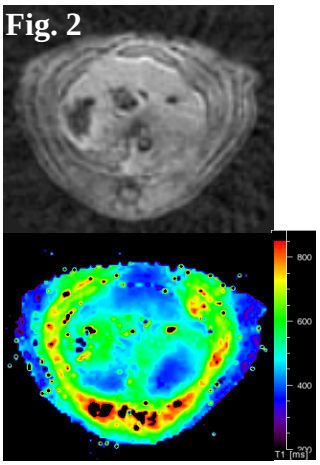


Fig. 1



Discussion: The low variability of the C_{Gd}-time curves enabled good quality fits for one-compartment modelling. The agreement of the individual K^{trans} and v_e values of liver and muscle underline the reproducibility of the method. No reference data for rat liver is available in literature for comparison. A flip angle of 10° provides sufficient SNR for rat *in vivo* imaging at 1.5T while still allowing for accurate Look-Locker correction. As expected, the radial sampling strategy showed minor sensitivity to motion artefacts however increased blurring at tissue interfaces. The golden angle radial spacing and k-space filtering technique enabled single-shot Look-Locker T1 mapping for free breathing DCE-MRI experiments in the rat abdomen.

References: [1] Landis CS et al. MRM, 44, 2000; [2] Cheng HM et al. MRM, 55, 2006 ; [3] Winkelmann S. et al. IEEE, 26(1), 2007; [4] Deichmann R & Haase A. JMR, 96, 1992; [5] Tofts et al. JMRI, 7, 1997 ; [6] Yankeelov TE et al. MRM, 57, 2007