Simultaneous estimation of T_1 and T_2* with a 2D radial multi gradient echo sequence: proof of concept in a phantom and in tumor bearing mice in the presence of respiratory motion

J. Vautier[1,2], M. Heilmann[1,2], C. Walczak[1,2], J. M. Pérez-Sánchez[3], and A. Volk[1,2]

[1]Institut Curie, Orsay, France, [2]INSERM U759, Orsay, France, [3]Instituto de Estudios Biofuncionales, Universidad Complutense de Madrid, Madrid, Spain

Purpose: Simultaneous dynamic T_1 and T_2* assessment has been shown to be of interest to characterize vasculature of tumors at high magnetic field [1]. Radial sequences are already used in DCE-MRI [2] and T_2* quantification has been described recently [3]. The purpose of this study was to implement and evaluate, on a phantom and *in vivo*, a radial multi gradient echo technique for simultaneous dynamic T_1 and T_2* estimation at high temporal resolution in experimental tumors.

Materials and Methods: A 2D-radial RF-spoiled multi gradient echo (2D RAD-MGE) technique was developed on a 4.7T small animal scanner (Biospec, Bruker Biospin, Ettligen, Germany). A home-built quadrature birdcage coil was used (∅=52mm, length=70mm). 6 echoes were acquired TE = (3.1; 4.5; 5.6; 7.0; 8.1; 9.4)ms. TR was 20ms (minimal TR allowed by our system). Flip angle was 12° on the phantom and 8° on the mouse (1ms sinc3 pulse). 64 projections and 64 complex readout points were acquired with a total acquisition time of t_{acq}=1.3s. Images were reconstructed using a home-written standard regridding algorithm (Kaiser-Bessel kernel, W=3, β=13.9, regridding matrix=128x128) [4] yielding, after FT and resizing, 64x64 matrix images (FOV=30x30mm^2, slice thickness=2mm). In order to optimize image reconstruction and to correct for residual eddy currents, k-space trajectory was measured [5] before acquisition and a density compensation function (DCF) was calculated and applied using the Voronoï diagram [6] (Matlab, Mathworks, Natick, USA). Dummy scans (mouse: 100x, phantom: 200x) were performed prior to acquisition to achieve steady state. T_2* and S_{TE0} were estimated pixel by pixel by mono-exponential fitting (IDL, ITT Visual Information Solutions, Boulder, USA): $S(TE) = S_{TE0} e^{-TE/T_2^*}$. S_{TE0} was given by the signal equation from spoiled gradient echo steady state: $S_{TE0} = \rho_0 \frac{(1-E_1)\sin\vartheta}{(1-E_1\cos\vartheta)}$. With $E_1 = e^{-TR/T_1}$, T_1 was estimated according to $T_1 = -\frac{TR}{\ln(\frac{\rho_0 \sin\vartheta - S_{TE0}}{\rho_0 \sin\vartheta - S_{TE0}\cos\vartheta})}$. Proton density ρ_0 was calculated using a prior calibration by a variable flip angle measurement. K-space trajectory was measured in both, phantom and mouse, latter with and without respiratory trigger. To evaluate the respiration effect, a set of images was acquired with and without respiratory trigger with TR=1.25s (similar to respiration delay), all other parameters being identical. Dynamic acquisition (500 repetitions of the sequence at highest temporal resolution) has been tested on a multi-compartment phantom and *in vivo* on a nude mouse bearing a colorectal tumor subcutaneously implanted at the abdominal level (TC302, Institut Curie, France). Phantom compartments (4 vials) contained Endorem® (Guerbet, France) solutions at different concentrations yielding a range of T_1 and T_2* values similar to *in vivo*. T_1 and T_2* estimations in the phantom were compared respectively to inversion-recovery (IR) T_1 measurement (MTX=128x128, FOV=30x30mm^2, TR=15s, TE=9ms, t_{acq}=8h) and to cartesian multi gradient echo (MGE) T_2* measurement (MTX=64x64, FOV=30x30mm^2, TR=15s, ΔTE=2ms, 100 echoes, t_{acq}=15min). T_1 and T_2* were measured (mean±SD) on parameter maps in ROIs covering the phantom compartments or the entire tumor.

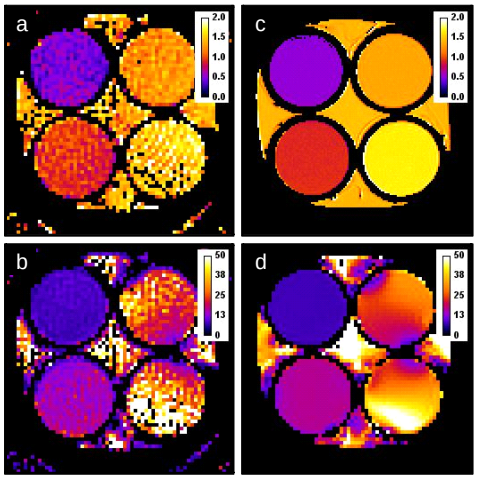

Fig. 1: a, b: T_1 (s) and T_2* (ms) maps of a phantom generated from a fast (1.3s) 2D RAD-MGE acquisition; **c:** T_1 (s) measured by IR; **d:** T_2* (ms) measured by cartesian MGE (susceptibility gradients are present in the right hand vials). Measurement times were 8h (c) and 15min (d).

Results and discussion: Phantom study: T_2* measured by MGE (7.2±0.1; 16.1±0.2; 24.7±3.0; 34.4±7.3)ms and by 2D RAD-MGE (7.8±0.7; 14.2±2.2; 26.6±9.8; 39.8±19.0)ms were in good agreement. T_1 measured by IR (1.64±0.01; 1.33±0.01; 0.89±0.02; 0.50±0.01)s, and estimated by 2D RAD-MGE (1.53±0.27; 1.26±0.13; 0.94±0.12; 0.45±0.08)s were similar too (Fig. 1). Mouse study: K-space trajectory needs to be measured with respiratory trigger in order to avoid artifacts. However, alternatively, the trajectory measured in a phantom under the same conditions allowed also to reconstruct correctly the *in vivo* images. With radial acquisition, images free of "classical" motion artifacts (but with some blur) were obtained without respiratory triggering (Fig. 2). This is of high importance for dynamic acquisitions as TR can be kept constant in order to stay in steady state. During the dynamic experiment, T_1 (1.87±0.06)s and T_2* (37.5±3.5)ms measured in tumor ROIs were reproducible (mean±SD from 500 repetitions).

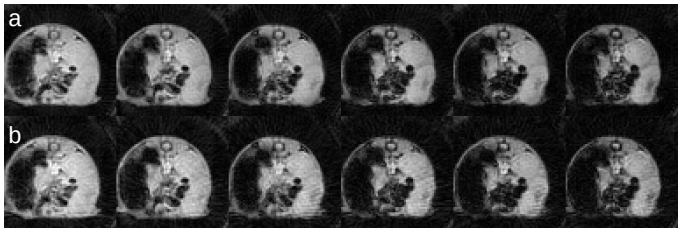

Fig. 2: Mouse images with increasing TE from left to right: with (a) and without (b) respiratory trigger. Image quality is very similar.

Conclusion: In this study, an optimized protocol for simultaneous dynamic T_1 and T_2* estimation by a fast 2D radial MGE technique was developed. Relaxation time estimations were validated on a phantom with respect to standard cartesian MRI sequences, and feasibility of high resolution dynamic acquisitions in the presence of respiratory motion was demonstrated on tumor bearing mice. With this approach, it should be possible to combine the advantages of multiparametric, T_1 and T_2* based, dynamic MRI for characterization of tumor vasculature with those of radial acquisition techniques (low motion sensitivity, pseudo high temporal resolution with sliding window reconstruction, multi spatial resolution acquisition).

References: [1]: Heilmann M. *et al* MAGMA 2007 (in press, DOI: 10.1007/s10334-007-0082-2); [2]: Song H.K. *et al* MRM 2001, 46(3): 503-9; [3]: Winkelmann S. *et al* JMRI 2006, 24(4): 939-44; [4]: Jackson J.I. *et al* IEEE Trans Med Imaging 1991, 10(3): 473-8, [5]: Beaumont M. *et al* MRM 2007 58(1): 200-5, [6]: Rasche V. *et al* IEEE Trans Med Imaging 1999, 18(5): 385-92

Acknowledgements: The authors would like to acknowledge the Cancéropôle Ile-de-France for support.