Rapid Simultaneous Data acquisition of T1 and T2 Mapping, using Multishot EPI and Automated Variations of TR and TE at 3T

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INTRODUCTION The change of the apparent relaxation rate ($R_1=1/T_1$) ΔR_1 has been accepted as a quantity, which linearly reflects the concentration of a contrast agent in DCE MR imaging using Gd-DTPA.[1] Pharmacokinetic information can be obtained from the time-variation of $\Delta R_1(\mathbf{r})$ at position $\mathbf{r}$. Conventional spin-echo imaging pulse sequence with either inversion-recovery (IR) or saturation-recovery (SR) has been used as the most accurate acquisition methods. However, its usage is limited the stationary tissues because of the long imaging time. 2D ss-EPI and its variant sequences (Look-Locker EPI, IR-EPI) may be used for very high temporal resolution, but its spatial resolution is limited by susceptibility-induce artifact. Multishot spin-echo EPI acquires relatively good temporal and spatial resolutions [2]. The change of T_2 due to the change of the concentration of the Gd^{3+} based contrast agent might introduce T_2 decay effect into the T_1 map, although the change in T_2 may be negligible at low concentration. The angiogenesis of the tumor physiology typically induces the large blood supply, which may require the correction of T_2 decay effect from the dynamic MR signal to improve the accuracy of the quantization of the pharmacokinetics. In this report, a rapid multishot EPI (ms-EPI) is presented to simultaneously acquire images to measure T_1 and T_2 values, and use T_2 remove the T_2 decay effect from dynamic T_1 map.

METHODS Using IDEA (MR Pulse sequence development environment) on a clinical 3T MRI system (Siemens Medical Solution, Erlangen, Germany), a segmented spin-echo EPI sequence was modified to add 180° RF and acquisition of an additional EPI echotrain, and to vary TR and TE automatically to eliminate the additional delay time between different series of imaging. The signals from both echoes in each segment are described by the standard saturation recovery spin-echo equation [3]

$$S(\vec{r}:TR,TE)=S(\vec{r}:\infty,TE)\cdot\left(1-e^{-TR/T_1(\vec{r})}\right)\cdot e^{-TE/T_2(\vec{r})} \quad \text{eq. (1)}$$

T_1 measurement was accomplished by using first echoes in multishot segments with different TRs and fixed TE (TE_0). The calculation of precontrast T_1 for the acquired data was performed pixel-by-pixel, using equation (1). T_2 was calculated by using second echoes with different TRs and TEs (TE_1 and $TE_n=TE_1+(n-1)\Delta TE$). At each imaging pixel, two echo signals of the largest TR which had echo times TE_0 and TE_n, were used for the smallest and the largest TEs, and the second echoes of other repetition times were used for those in between. For example, the signal for the echotime TE_a was calculated from the second echo of TR_a multiplied by a factor S_{n0}/S_{a0}, which is the ratio of the first echo signals of TR_n and TR_a where TR_n (the largest TR),

$$S_{na}=S_{n0}/S_{a0}\cdot S_{aa}=(S_0\cdot(1-e^{-TR_n/T_1})\cdot e^{-TE_0/T_2})/(S_0\cdot(1-e^{-TR_a/T_1})\cdot e^{-TE_0/T_2})\cdot S_0\cdot(1-e^{-TR_n/T_1})\cdot e^{-TE_n/T_2}=S_0\cdot(1-e^{-TR_n/T_1})\cdot e^{-TE_n/T_2} \quad \text{eq. (2)}$$

Then, the resultant signal for the second echo TE_a has the same T_1 recovery term $1-e^{-TR_n/T_1}$ as those echoes of TE_0 and TE_n. These signals were fitted to a line (semi-log) to calculate T_2 value. The calculation of dynamic T_1 is similar to that of precontrast T_1 except that we use the calculated value for the fully recovered signal intensity,

$$S_0(\vec{r};C(t))=S_0(\vec{r})e^{-TE_0/T_2'(\vec{r};C(t))}=S_0(\vec{r})e^{-TE_0/T_2(\vec{r};C=0)}\cdot e^{-TE_0(1/T_2'(\vec{r};C(t))-1/T_2(\vec{r};C=0))}$$

as an additional data in dynamic imaging, where $S_0(\vec{r})e^{-TE_0/T_2(\vec{r};C=0)}$ is the fully recovered signal intensity with $T_2(\vec{r};C=0)$ and T_2 decay calculated from precontrast imaging, and $e^{-TE_0(1/T_2'(\vec{r};C(t))-1/T_2(\vec{r};C=0))}$ shows how we correct T_2 changing effect by using both precontrast and dynamic T_2 maps.

$MnCl_2$ solution phantom was used to acquire T_1 and T_2 using both ms-EPI and IR-SE to measure and compare the stationary T_1 and T_2. The technique was applied to acquire DCE data from a mouse in transaxial imaging plane, with echotrain 3, imaging matrix 64x28, inplane spatial resolution 1.0x1.0 mm^2, and slice thickness 2.0 mm, for 4 slices. Precontrast T_1 mapping data was acquired with 8 different TRs: 0.15, 0.25, 0.4, 0.8, 1.4, 2.2, 3.2, and 5.0 s; TE0=8.1 ms, TE1=19 ms and ΔTE=10 ms, and used to calculate baseline relaxivity $1/T_{10}(\mathbf{r})$, the equilibrium magnetization $M_o(\mathbf{r})$ and $T_{20}(\mathbf{r})$. In dynamic imaging, TRs of 250, 500, and 800 ms; TE_0=8.1 ms, TE_1=19 ms, ΔTE =15 ms were used, of which the number and the values of TRs and TEs may vary depending on the desired temporal resolution. The total number of slices was limited by the shortest TR, i.e., 0.25 s in current protocol.

RESULTS & DISCUSSIONS T_1 recovery curves of different concentrations of $MnCl_2$ solutions are plotted in Fig. 1. We treated the data for 0.1 mMol as the precontrast data and the first 3 points in 0.3 and 0.6 mMol as dynamic data. At TR=6 s, the calculated signal values are close to the measured values; 1854.1 versus 1884.8 for 0.3 mMol and 1265.4 versus 1339.7 for 0.6 mMol. The signal recovery plots in Fig. 2 indicate the shortening of T_1 values of the early few time points within 2.0 min after administration of contrast agent. Imaging duration for each time point was 20.0 s for 3 TRs. Precontrast and dynamic T_1 curves reached the different values at TR=10 s, which is similar to our phantom T_1 results in Fig. 1. This is because of the change of T_2 due to the change of the concentration of the Gd^{3+} based contrast agent. The total imaging time for a time point may be varied by changing acquisition matrix and ETL. Larger ETL can be used to compensate the increased imaging time by larger imaging matrix, at the expense of increased susceptibility-induced distortion. This technique (ms-SEPI) is faster and the result is more accurate than those using fast gradient echo imaging with flipangle variation, of which the calculated T_1 value is sensitive to RF field uniformity.[3, 4] The calculated T_1 and T_2 maps are displayed in Fig. 3 for a middle slice. T_1 does not change at the core of the tumor which indicates the lack of the blood flows, while it rapidly changes near the tumor periphery due to the a high Gd^{3+} concentration by large blood supply, where T_2 correction may be necessary.

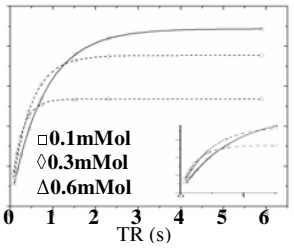

Fig. 1 *T_1 recovery curves of $MnCl_2$ solution phantoms with different concentrations.*

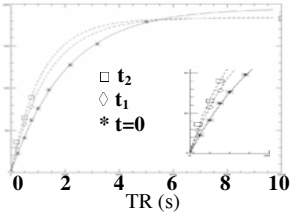

Fig. 2 *T_1 recovery curve of a single pixel at different dynamic time point (0, 0.9, 1.5 min), near a tumor periphery, indicated by the arrow in Fig. 3.*

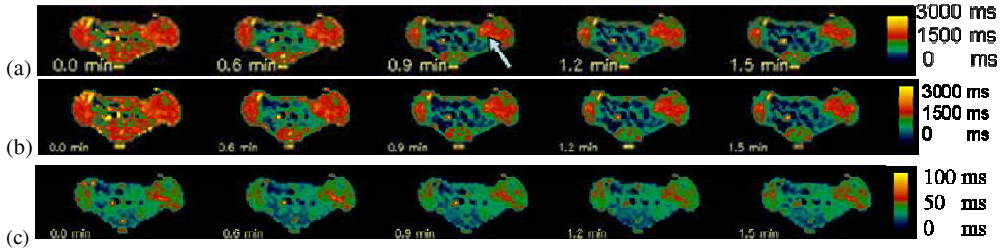

Fig. 3. *T_1 maps (a) with and (b) without T_2 correction, and (c) T_2 maps at different time. The 1st images are the precontrast maps.*

CONCLUSIONS A rapid acquisition technique for simultaneously T_1 and T_2 mapping, multishot EPI with automated TR and TE variations, was developed and applied to acquire quantitative DCE imaging in small animal (mice). The method may be useful to improve the accurate pharmacokinetics study upon drug delivery.

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