Singleshot Measurements of T_1 and Field Variation using 2D Simultaneous Singleshot Spin-, Gradient-, and Stimulated-EPI (2D ss-SGSTEPI)

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INTRODUCTION: A stimulated echo is formed after three 90^o RF pulses. Magnetization-preparation such as diffusion-weighting, T_2 weighting, or displacement encoding, is accomplished immediately after the first RF pulse and the second RF pulse restores a half of prepared magnetization into the longitudinal space. Other half is remained on the transverse plane. In the conventional stimulated-echo pulse sequence, a half of the prepared magnetization is discarded. Considering that nuclear magnetic resonance (NMR) is one of the least sensitive measurement techniques due to the tiny transition energy in nuclear Zeeman interaction (~μeV) and large efforts have been focused to improve SNR in NMR/MRI, stimulated-echo NMR is an ineffective measurement technique.

A novel imaging technique has been developed to utilize other half of the prepared magnetization at the spin-echo position and simultaneously acquire spin-, gradient-, and stimulated-EPI in a singleshot using 2D singleshot spin-/stimulated-EPI (2D ss-SGSTEPI). Preliminary results using 2D ss-SGSTEPI is presented.

METHOD Sequence diagram is shown in Fig.1. This imaging technique is useful for rapid and singleshot measurements of T_1 and phase difference. Phase difference map can be used to estimate field variation or displacement measurement. It reflects the local variation of the static magnetic field. In Fig.1, spin echo (SEPI) ($S_{SEPI}=M_0\sin(\alpha)\sin^2(\alpha/2)\exp(-TE/T_2)$ and stimulated echo (STE) ($S_{STE}=(1/2)M_0\sin^3(\alpha)\exp(-TM/T_1)\exp(-TE/T_2)$) are acquired simultaneously, where $\alpha(=90^o)$ is the flip-angles of the 2nd and 3rd RF pulses. If the flip-angles for the 2nd and 3rd RF pulses are perfect 90^o, the only difference between SEPI and STEPI is T_1 decay during the mixing time TM. Because of the imperfect 90^o in practice, there is an increase in SEPI signal and decrease in STEPI signal, which causes underestimation of the calculated T_1 value. The error in T_1 calculation caused by imperfect 90^o RF pulses can be easily corrected using an additional acquisition (STE_0) with minimum mixing time (TM≈0). The RF correction map can be measured as: $f(r, \alpha) = S_{STE0}/S_{SEPI}$, where S_{SEPI} is spin echo signal. The ratio between STE and SEPI is calculated by equation: $S_{STE}/S_{SEPI}=\exp(-TM/T_1)*f(r, \alpha)$. Finally, T_1 is derived from the equation, $T_1=TM/[\ln(f(r, \alpha)*S_{SEPI})-\ln(S_{STE})]$. Correspondingly, phase difference during ΔTE is calculated by subtracting the phase of SEPI from that of GEPI. Then this phase change is converted into frequency offset: $\Delta f=(\theta_{GEPI}-\theta_{SEPI})/(\Delta TE)$, where ΔTE is the difference of echo time between SEPI and GEPI and θ_{SEPI} and θ_{GEPI} are phase angle of SEPI and GEPI, respectively. Both T_1 and phase maps were constructed in realtime using online imaging construction program. MR imaging experiment was performed on a fluid phantom filled with $MnCl_2$/water solution with TR=3.0s, TM/TE=500/17ms, 160x40 matrix, 5 slices, 1.5x1.5x2 mm^3 spatial resolution, using a transmit/receive head coil. T_1 of the fluid was independently measured as 0.95 s, using spin-echo MRI. 2D ss-SGSTEPI was also applied to a mouse to measure the singleshot T_1 with TR/TE=4.0s, TM/TE=200/16.7ms, 128x28 matrix, 12 slices, 1x1x2 mm^3 spatial resolution, using home-built RF coil.

RESULT & DISCUSSIONS: Figs.2a and 2b indicate the resultant MR images and processed T_1 and phase maps. The phantom was placed near one end of a Transmit/receive head coil, as shown in Fig. 2a. The phase-difference map in Fig. 2b is independent from any phase drift between shot-to-shot. Images in Fig. 2c indicate the magnitude images of SEPI, STEPI, and $STEPI_o$, and calculated T_1 maps with and without RF correction. We expect the minimal spatial variation of the T_1. However, T_1 profile along the vertical dotted line on T_1 without RF correction varied about 30 %, while it was dramatically improved by RF correction. After RF correction, T_1 was measured as 0.9 s, which is comparable to 0.95 s measured using spin-echo imaging.

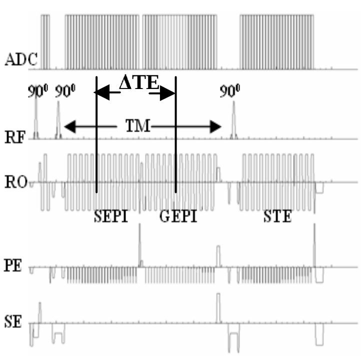

Figure 1: Sequence diagram of T_1 map and Field inhomogeneity measurement.

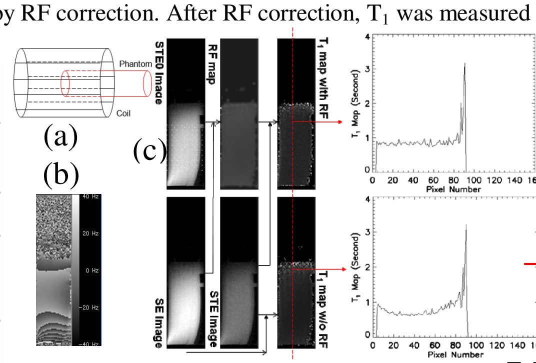

Figure 2: (a) Phantom setup diagram. (b) Field inhomogeneous map. (c) Source images, RF map and resultant T_1 map with or without RF correction.

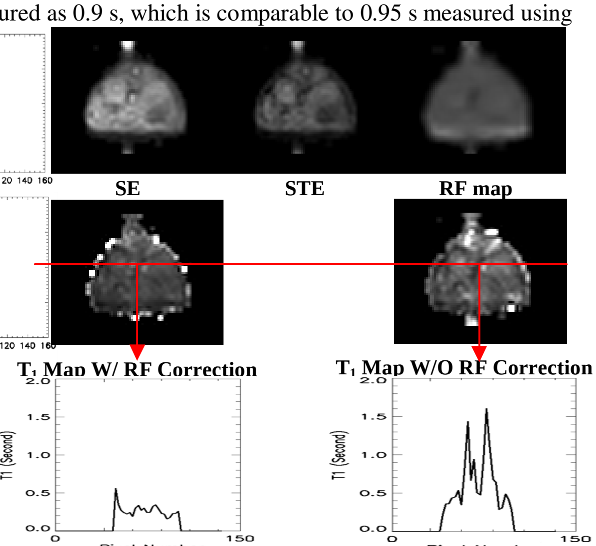

Figure 3: Mouse T1 map measurement.

In Fig.3, singleshot T_1 mapping of an in-vivo mouse is shown. Because the technique is a singleshot imaging technique, the resultant MR images do not suffer from any motion-induced artifact. T_1 profile along the horizontal line on the uncorrected T_1 map indicates large elevation near the center. T_1 value varies more smoothly after RF correction.

These preliminary results demonstrate potential application of this technique which is powerful on rapid T_1 measurement. Moreover online real-time display of T_1 map calculation was integrated into the realtime reconstruction program chain on Siemens 3T Trio scanner. Singleshot T_1 mapping using 2D ss-SGSTEPI may be useful for rapid and accurate estimate of the local concentration of the paramagnetic-ion based contrast agent in dynamic contrast MR imaging, which may improve the accuracy of the pharmacokinetic parameters.

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