Single-shot Measurements of T1 and Field Variation using 2D Simultaneous Single-shot Spin-, Gradient-, and Stimulated-EPI (2D ss-SGSTEPi)

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INTRODUCTION: A stimulated echo is formed after three 90° RF pulses. Magnetization-preparation such as diffusion-weighting, T2 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 single-shot using 2D single-shot 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 T1 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) (SEPI) = S0 exp(-TE/T1) and stimulated echo (STE) (STE) = M0 sin(α)exp(-TM/T1) exp(-TE/T2) are acquired simultaneously, where α=90° is the flip-angles of the 2nd and 3rd RF pulses. If the flip-angles for the 2nd and 3rd RF are perfect 90°, the only difference between SEPI and STEPI is T1 decay during the mixing time TM. Because of the imperfect 90° in practice, there is an increase in SEPI signal and decrease in STEPI signal, which causes underestimation of the calculated T1 value. The error in T1 calculation caused by imperfect 90° RF pulses can be easily corrected using an additional acquisition (STE0) with minimum mixing time (TM=0). The RF correction map can be measured as: S(α) = STE0/STEPI, where SSTEPI is spin echo signal. The ratio between STE and SEPI is calculated by equation: S(α) = STE0/STEPI = exp(-TM/T1) exp(-TE/T2).

Finally, T1 is derived from the equation, T1 = TM[ln(f(r, α) + STE0)/ln(STE0)]. Correspondingly, phase difference during ΔT1 is calculated by subtracting the phase of SEPI from that of GEPI. Then this phase change is converted into frequency offset: Δθ = (θSEPI - θGEPI)/ΔT1, where ΔT1 is the difference of echo time between SEPI and GEPI and θSEPI and θGEPI are phase angle of SEPI and GEPI, respectively. Both T1 and phase maps were constructed in realtime using online imaging construction program. MR imaging experiment was performed on a fluid phantom filled with MnCl2/water solution with TR=3.0s, TM/TE=500/17ms, 160x80 matrix, 5 slices, 1.5x1.5x2 mm3 spatial resolution, using a transmit/receive head coil. T1 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 single-shot T1, TR/TE=T1, T1/TE=200/16.7ms, 128x28 matrix, 12 slices, 1x1x2 mm3 spatial resolution, using home-built RF coil.

RESULT & DISCUSSIONS: Figs.2a and 2b indicate the resultant MR images and processed T1 maps with and without RF correction. We expect the minimal spatial variation of the T1. However, T1 profile along the vertical dotted line on T1 without RF correction varied about 30 %, while it was dramatically improved by RF correction. After RF correction, T1 was measured as 0.9 s, which is comparable to 0.95 s measured using spin-echo imaging.

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