Diffusion Weighted Imaging with Reduced Susceptibility Artifact, using 2D singleshot DW-STimulated EPI (2D ss-DWSTEPI)

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INTRODUCTION Diffusion-weighted imaging (DWI) and diffusion tensor imaging (DTI), using conventional 2D singleshot diffusion weighted-EPI (2D ss-DWEPI), are limited to intracranial applications far from the sinus, due to the severe geometric distortion caused by strong non-linear local magnetic fields at or near tissue/air or tissue/bone interfaces. Although susceptibility-induced distortions can be reduced by using multishot DW acquisition techniques, these multishot techniques, in general, suffer from artifacts caused by phase errors induced by small local or global motions during the application of the large diffusion gradients. These artifacts cause inaccuracy in diffusion measurement. Other alternative singleshot imaging techniques that are based on multi-spin-echo imaging such as single shot fast spin-echo (HASTE) and singleshot gradient-spin-echo (GRASE), are subject to substantial decay due to the long data acquisition duration, which is typically an order of the magnitude longer than T_2 relaxation time of tissue water. This decay causes severe blurring on the resultant DW images. 2D singleshot STimulated EPI (2D ss-STEPI) is presented as a novel technique to perform diffusion MRI of a region with a strong local susceptibility magnetic field gradient. Acquisition of the entire k_y-domain is segmented into the several EP echotrains and is completed after a single diffusion-preparation. The diffusion prepared magnetization decays with T_1, which is typically an order longer than T_2. This technique can acquire the data for the full field-of-view in phase-encoding direction, unlike 3D ss-DWSTEPI that measures a limited FOV in both phase- and slice-directions [1]. 2D ss-STEPI may be advantageous for high-resolution DTI, because the distortion is reduced on the resultant DWI which is proportional to the length of each stimulated echotrain, and because it is a singleshot acquisition technique which is immune to motion.

METHODS 2D ss-DWSTEPI acquires the entire 2D k-space data in a single diffusion prepared (DP) driven-equilibrium (DPDE) (see Fig. 1 on the right). DPDE contains a 90° excitation, 3 reference gradient echoes for EPI phase correction, diffusion-weighting, a dephasing crusher gradient, a 90° tipup, and a spoiling gradient. The imaging part combines stimulated echo imaging and an echoplanar readout. Eq. (1) describes the longitudinal magnetization just before the n^{th} imaging RF pulse with respect to the previous longitudinal magnetization M^z_{n-1} [1].:

$$M^z_n(\vec{r}) = M_o(\vec{r})\left(1 - e^{-\tau/T_1(\vec{r})}\right) + M^z_{n-1}(\vec{r}, t_{n-1}) \cdot \cos\alpha_{n-1} \cdot e^{-\tau/T_1(\vec{r})} \quad (1)$$

Here, α_n is the flip angle of the RF pulse for the n^{th} echotrain and τ is the duration of each segment. The two terms are the freshly recovered and DP magnetizations, respectively. Signal from the first term, which is the freshly recovered and non-DW magnetization, is spoiled after each excitation by the rephasing crusher gradient indicated by left arrows (←), therefore the measured NMR signal reflects only DP magnetization which undergoes T_1 decay. Note that only half of the DP magnetization is used for the imaging, which results in halving the total DP magnetization that is available for the detection. This is a result of spoiling half the magnetization on the transverse plane after tipup RF.

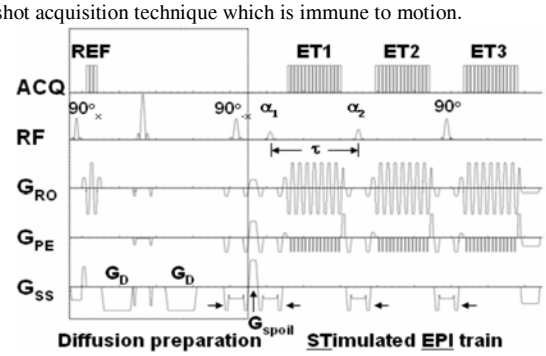

Fig. 1. *Pulse diagram of 2D ss-DWSTEPI, consisting of DPDE and a few stimulated echotrains. A complete set of k_y views is segmented into 3 acquisitions, in this example. τ is the duration of each readout segment, including RF and crusher gradients. The horizontal arrows → and ← represent the dephasing- and rephasing-crusher gradients, respectively, which remove non-DW magnetization from the signal. The vertical arrow indicates spoiling gradient.*

The transverse magnetizations $M^+_{n-1}(\vec{r},t)$ and $M^+_n(\vec{r},t)$ after two consecutive RF pulses (α_{n-1} and α_n) are: $M^+_{n-1}(\vec{r},t) = M^z_{n-1}(\vec{r},t^-_{n-1}) \cdot \sin\alpha_{n-1}$, $M^+_n(\vec{r},t) = M^z_{n-1}(\vec{r},t^-_{n-1}) \cdot \cos\alpha_{n-1} \cdot e^{-\tau/T_1(\vec{r})} \cdot \sin\alpha_n$ (2)

To achieve equal signal amplitude ($M^+_n(\vec{r},t) = M^+_{n-1}(\vec{r},t)$) the relationship between the flip angles of two adjacent RF pulses should satisfy,

$$\tan\alpha_{n-1} = \sin\alpha_n \cdot e^{-\tau/T_1(\vec{r})} \quad (3)$$

The flip-angle for the last RF pulse is set to 90° to consume all remaining longitudinal magnetization, and the flip angles of the proceeding RF pulses can be calculated using the relation in Eq. (3).

All imaging studies were performed on a 3 Tesla MRI scanner (Siemens Medical Solutions, Erlangen, Germany). Typical imaging parameters for ss-DWEPI were used, such as 128 readout matrix, 5/8 partial asymmetric acquisition in the phase-encoding direction to reduce actual data sampling duration. The diffusion gradient was applied on a fluid phantom mixed with $MnCl_2$ (T_1/T_2=1.0 s/60 ms) with b=200 s/mm^2 in z direction. The flipangle was varied along the stimulated echotrain to reduce blurring in the phase direction that is caused by the change in the diffusion prepared longitudinal magnetization due to T_1 decay and the partial excitation [1].

RESULTS & DISCUSSIONS Plot (a) in Fig. 2 demonstrates the signal decay along the stimulated echoes with phase-encoding gradient turned off. The resultant image in Fig. 2b is a 2D slice. The ramped variable flipangle was calculated using Eq. (3) as 29.2, 34.6, 44.5, and 90° for each RF pulse. A reference T_1 value of 1.0 s was used to calculate the variable flipangle, which is close to that of white-matter at 3.0 T. The duration τ for each ET was 17 ms for actual ETL of 14, using the receiver bandwidth 1.86 kHz/px. The plot in Fig. 2a demonstrates that the total signal for each echotrain is similar using the ramped variable flipangle. The same location was imaged using the conventional 2D ss-EPI for the comparison. The total numbers of k_y lines were similar (about 56) for both STEPI and EPI. The actual ETL was reduced by factor of 4 for STEPI (14 instead of 56). It is clear that the susceptibility artifact caused by air bubbles stuck on the acryl wall, was significantly reduced in 2D ss-STEPI, as in Fig. 2b compared to that in Fig. 2c. Although the spatial resolution was limited by low SNR in our preliminary application of 2D ss-STEPI, this imaging technique may be useful for extra-cranial applications of fMRI, DWI, and DTI.

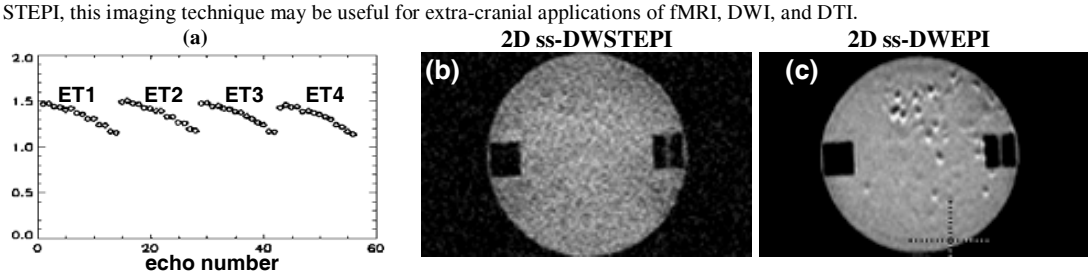

Fig. 2. (a) *Plot of the peak magnitude of the 4 stimulated echotrains of 14 EP echoes for 2D ss-STEPI with the phase-encoding gradient turned off, and (b) resultant image. Image in (c) is a 2D ss-DWEPI image with actual ETL of 56. 2D ss-DWSTEPI image in (b) indicates a significant reduction of susceptibility artifact from the air bubbles as shown in (c).*

CONCLUSIONS 2D ss-STEPI can acquire the diffusion-weighted magnetization of a region with large variation of the local field in a single excitation with very little susceptibility distortion. Even though its SNR is limited, 2D ss-STEPI may allow acquisition of high resolution DTI data from nearly any region of the body, overcoming the limitations of conventional 2D ss-EPI. This new technique not only reduces susceptibility artifacts, but also freezes most physiologic motion because it is a singleshot imaging technique.

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REFERENCES: [1] E. K. Jeong, S. E. Kim, E. G. Kholmovski, et al., Magn Reson Med, in print (December, 2006).