Single Shot ADC

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Introduction

Previous studies have suggested that dynamic ADC changes[1, 2] may provide more accurate spatial localization of brain function than BOLD signal. In particular, the temporal characteristics of the ADC activation revealed that it originated from vascular networks upstream of the BOLD sources. However, the traditional method used to acquire dynamic ADC contrast requires the use of multiple b factors and thus multiple repetitions per ADC map, thereby limiting its temporal resolution. To further assess the temporal behavior of the ADC activation with respect to the BOLD activation, and to characterize the vascular signal propagation from the arterial side, through capillaries, to the venous side, much improved temporal resolution on the order of hundreds of milliseconds is required (It takes less than one second for blood flow with speed of 1-4 mm/s to pass through the capillaries with length less than 1 mm.) In this abstract, we propose using a single-shot method that combines a gradient echo with two spin echoes at different diffusion weightings to simultaneously obtain both the BOLD (from the first spin echo) and ADC measurements. Preliminary results show that images with sufficient SNR to estimate ADC can be achieved with this method within one sec. The method has the potential to acquire an ADC map every 500 ms.

Methods

The single shot ADC sequence, which is shown in Figure 1, acquires one gradient echo and two spin echoes. The gradient echo, which occurs immediately after the excitation, had no diffusion weighting gradients thus providing a good approximation of the proton density. Two different sets of isotropic diffusion weightings[3], with b factors of b_1 and b_2, were applied to the subsequent spin echoes, respectively. Importantly, both sets of gradients are mutually uncorrelated with the BOLD background gradients when this sequence is used in an fMRI scan, thereby eliminating the potential cross-terms between these gradients. If the BOLD gradients are assumed to be constant during a single ADC acquisition, the BOLD related b factor seen by the second spin echo can be derived as twice of that seen by the first spin echo. Under this condition, the signals of the two spin echoes and the corresponding ADC can be represented as:

$$\begin{aligned} S_1 &= S_0 e^{-TE_1/T_2} e^{-b_1 D} e^{-b_0 D} \\ S_2 &= S_0 e^{-TE_2/T_2} e^{-b_1 D} e^{-2b_0 D} e^{-b_2 D} \end{aligned} \longrightarrow \quad D = \frac{2\ln S_1 - \ln S_2 - \ln S_0}{b_2 - b_1} \qquad (b_2 \neq b_1,\ \mathrm{TE}_2 = 2\mathrm{TE}_1)$$

where S_0, S_1 and S_2 are signals of the gradient echo, 1st spin echo and 2nd spin echo respectively; b_1 and b_2 are b factors applied; b_0 is from the BOLD background gradients; TE_1 and TE_2 are echo times of the two spin echoes. By setting $TE_2 = 2TE_1$, both the T_2 effects and the BOLD background b factor are removed in the resultant D, thus the ADC contrast in this sequence is independent of the BOLD effect.

The sequence was implemented on a General Electric 4T MR scanner (GE Healthcare, Milwaukee, WI) equipped with an echo-speed gradient system (40 mT/m at 150 T/m/s slew rate). To accommodate the diffusion weighting gradients and the spiral readout trains, the echo times of the two spin echoes were lengthened. In our experiments, TEs of 73 and 146 ms were used with a one sec TR for two spin echoes respectively. Two sets of b factor pairs were tested. The first set had b factors of 7.05 and 14.1 s/mm^2 for the two echoes, and the other set had b factors of 37.85 and 75.7 s/mm^2. The FOV was set at 24 cm, matrix size at 64x64, and slice thickness was 5 mm. The experiments were carried out using both phantoms and human subjects. Four slices were acquired using an interleaved scan. Image reconstruction and ADC calculations were performed offline. The resultant ADC values and signal to noise ratios (SNR) were compared with the conventional three-shot scan.

Results and Discussion

The gradient-echo image, the two spin-echo images and the resultant ADC map of a human brain are shown in Figure 2. The SNRs are 175, 92 and 42 respectively. Although the TE for the 2nd spin echo is 146 ms in this experiment, the signal level is sufficiently high for deriving the ADC values. During signal acquisition, the 2nd echo and 3rd echo were abnormally suppressed. To adjust for this in the ADC calculation, the spin-echo images were scaled by a constant for each echo to the predicted intensities as shown in Figure 2. The sources of this attenuation will be investigated in future work. Stronger diffusion weighting may be valuable in investigating ADC contrast. With a moderate increase in TE_2 to 165 ms, a total b-value of 300 s/mm^2 can be applied.

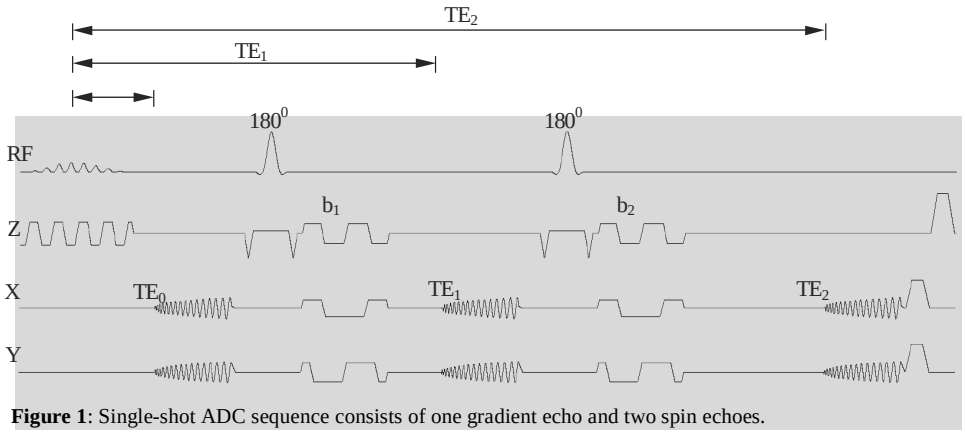

Figure 1: Single-shot ADC sequence consists of one gradient echo and two spin echoes.

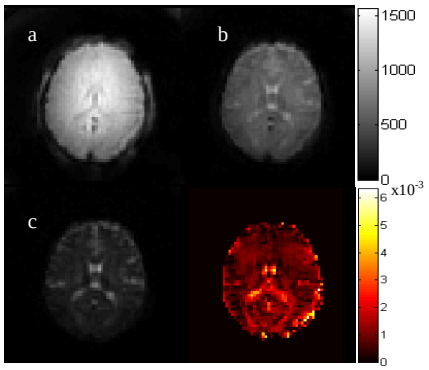

Figure 2: (a) Gradient echo image; (b) 1st spin echo image; (c) 2nd spin echo image; (d) ADC map. The two color bars are for echoes and ADC map respectively.

Conclusion

A single-shot imaging method to simultaneously measure the BOLD (from the first spin echo) and ADC changes has been developed. It can measure ADC changes with a temporal resolution of less than a second. We hope this sequence can be used to assess the vascular signal propagation from the arterial to the venous networks with much improved temporal precision in future fMRI experiments.

Reference

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