High Resolution 3D Multi-slab Multi-shot Spin Echo Diffusion-Weighted Imaging

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Introduction: The need to detect small brain lesions and to reduce partial volume effects in fiber tractography has made high resolution isotropic 3D diffusion-weighted imaging (DWI) a research focus in diffusion acquisition methodology [1-4]. To achieve reasonable scan time, most of the proposed 3D DWI methods used steady-state free precession (SSFP) acquisition [1-3] which has a complicated diffusion-weighted signal model that requires separate T1 and T2 measurements. 3D spin-echo DWI can be described by the standard Stejskal-Tanner signal model, but it is limited by the prolonged acquisition times, especially when combined with cardiac gating. In this work, we propose to use a multi-slab multi-shot DWI sequence (SE) sequence to achieve high resolution 3D DW imaging with whole brain coverage within a reasonable scan time. 3D multi-slab acquisition is proposed to reduce the total scan time relative to single-slab 3D spin-echo DWI acquisition without a loss in SNR efficiency. Motion-induced phase errors resulting from multi-shot acquisition are corrected based on 3D navigators using the KICT algorithm proposed in [5, 6].

Method: 3D multi-slab acquisition: 3D multi-slab acquisition has been used previously in magnetic resonance angiography [7]. In 3D multi-slab acquisition the whole 3D imaging volume is divided into smaller 3D imaging volumes called slabs. Since each of the slabs in a 3D multi-slab acquisition is excited and imaged independently, the slabs can be interleaved to achieve shorter total acquisition time as compared to the conventional, single slab 3D imaging. Shorter acquisition times make the 3D multi-slab acquisition more immune to gross motion within a volume and across volumes, giving high quality DW images and diffusion metric maps. Another advantage of multi-slab acquisition is that when cardiac-gated acquisition is used, multi-slab acquisition allows different parts of the brain to be acquired during different parts of the cardiac cycles giving better cancellation of pulsation effects. This cardiac-gated strategy has been proposed elsewhere [8] for 2D DWI.

Advantages of 3D multi-slab acquisition include boundary artifacts and possible loss of signal-to-noise ratio (SNR) as compared to equivalent single slab 3D acquisitions. Boundary artifact of multi-slab acquisition can be reduced by overlapping adjacent slabs, interleaving odd and even slabs (Figure 1), and discarding some boundary slices of the 3D reconstructed slab images.

3D trajectory and 3D navigator: The 3D trajectory used in the current study was a stack of multishot constant density spirals. For motion-induced phase error correction, a stack of low resolution single shot constant density spirals was used as the navigator. Navigator data were acquired after every data acquisition by adding a second refocusing pulse to the spin echo sequence [6].

Phase error correction and image reconstruction: Cardiac-gating was used to minimize non-linear motion-induced phase errors. Linear motion-induced phase errors were estimated from the navigator data by subtracting the phase of the diffusion encoded navigator images with the phase of the b = 0 navigator images. Slopes and offsets of the linear phase errors were then estimated with least squares fitting and used to correct the k-space of the data [5, 6]. Corrected k-space data were then reconstructed using NUFFT [9] with the quadratic penalized weighted least squares (QPWLS) iterative algorithm. For the navigator data, a faster reconstruction strategy, including gridding in-plane and 1D FFT through-plane was used [6].

Results: SNR analysis: Let SNR0 and SNR3 be the SNRs of 3D (Cartesian sampling in the third dimension) and 2D interleaved spin echo acquisitions, respectively. For cardiac-gated acquisition with at most nex number of excitations per cardiac cycle, the ratio of SNR between 3D multi-slab and single slab single acquisition is given in Equation (1) (assuming similar diffusion encoding, similar readout duration). For 3D single slab acquisition, multiple excitations within a cardiac cycle are not employed due to the resulting short TR and severe SNR degradation. In Equations (1), nslab is the number of slabs in a 3D multi-slab acquisition, nexc is the number of excitations per cardiac cycle, and nex is the number of excitation per cardiac cycle.

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\text{SNR}_{3D} = \frac{\text{SNR}_{0}^{3D}}{\text{SNR}_{0}^{single}} = \left( \frac{n_{slab}}{3} \right)^{3/2}
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SNR_{3D} = 0.59

However, the total acquisition time of 3D multi-slab is 3 times shorter than that of 3D single slab acquisition. If we define as SNR efficiency the ratio of SNR over the square root of the total scan time, 3D multi-slab acquisition has then three times shorter scan time. Similar comparison is done for 3D multi-slab and equivalent interleaved 2D acquisition giving SNR_{3D}^{multi} = 3.33 SNR_{0} with the same total acquisition time.

In vivo data were acquired using Siemens 3 T Trio scanner with a 12-channel head coil on a healthy male subject in accordance with the institutional review board. The obtained resolution was 1.88 x 1.88 x 1.88 mm³ with full brain coverage. Other imaging parameters were: TE1 = 68 ms (for data), TE2 = 120 ms (for the navigator), b = 1000 s/mm², 3-shot constant density spiral in-plane trajectory with readout duration of 20.08 ms, 6 slabs with 16 Cartesian encoding steps per slab, overlapping factor of 25% of slab thickness. The acquisition was cardiac-gated with 3 slabs per R-R interval, resulting in an effective TR of 2 R-R intervals. If slabs were numbered from 1 to 6 starting from the lower part of the brain, slab acquisition order was [1 3 5; 2 4 6]. The obtained resolution was 1.88 x 1.88 x 1.88 mm³ with full brain coverage. Other imaging parameters were: TE1 = 68 ms (for data), TE2 = 120 ms (for the navigator), b = 1000 s/mm², 3-shot constant density spiral in-plane trajectory with readout duration of 20.08 ms, 6 slabs with 16 Cartesian encoding steps per slab, overlapping factor of 25% of slab thickness. The acquisition was cardiac-gated with 3 slabs per R-R interval, resulting in an effective TR of 2 R-R intervals. If slabs were numbered from 1 to 6 starting from the lower part of the brain, slab acquisition order was [1 3 5; 2 4 6]. Cardiac-gating was used to minimize non-linear motion-induced phase errors. Linear motion-induced phase errors were estimated from the navigator data by subtracting the phase of the diffusion encoded navigator images with the phase of the b = 0 navigator images. Slopes and offsets of the linear phase errors were then estimated with least squares fitting and used to correct the k-space of the data [5, 6]. Corrected k-space data were then reconstructed using NUFFT [9] with the quadratic penalized weighted least squares (QPWLS) iterative algorithm. For the navigator data, a faster reconstruction strategy, including gridding in-plane and 1D FFT through-plane was used [6].

Discussion and Conclusion: We have proposed a novel 3D high resolution DWSE acquisition strategy for whole brain coverage. In vivo high quality 1.88 x 1.88 x 1.88 mm³ resolution DW images and FA maps with whole brain coverage were obtained within reasonable scan time. With appropriate slab acquisition order and cardiac gating, the proposed technique is fairly immune to pulsation. Compared with other SSFP based techniques [1-3], our acquisition scheme can achieve similar resolution within similar total acquisition time while using the standard Stejskal-Tanner signal model. Compared to fast spin echo based techniques [4], the current method is more time efficient due to the use of multi-slab acquisition. For future development, optimization on diffusion directions and further reduction of total acquisition time per one 3D volume the sensitivity of 3D diffusion acquisition using imaging can further reduce the higher quality diffusion metrics.