Accurate Gray Matter CBF Mapping in Whole Brain IR 3D PULSAR Imaging through Flip Angle Modulation

N. D. Gai[1], S. L. Talagala[2], and J. Butman[1]

[1]Radiology & Imaging Sciences, National Institutes of Health, Bethesda, MD, United States, [2]NINDS, National Institutes of Health, Bethesda, MD, United States

Introduction: Modification of the PULSAR technique [1] by use of a non-selective background suppression inversion pulse along with 3D-Turbo Field EPI (TFEPI) acquisition, labeled IR-3D-PULSAR, provides whole brain perfusion imaging in about five minutes [2]. Data acquisition is through use of centric-ordered slice encoding (along k_z) followed by single-shot gradient-echo EPI acquisition. In addition to tagged blood signal decay during extended data acquisition, modulation of k-space during non-steady state constant flip angle acquisition results in blurring. This introduces inaccuracies in gray matter and white matter perfusion values. Here we analyze the effect of non-steady state constant flip angle imaging and correct for blurring by modulating the flip angle train such that magnetization in gray matter remains almost constant across all 3D-TFEPI shots.

Materials and Methods: For a spoiled generic gradient echo sequence, the longitudinal and transverse magnetization evolution can be described by [3]

$$M_z(n) = M_z(n-1)E_1.\cos\alpha + M_z(n-1)(1-E_1) \quad \text{and} \quad M_{xy}(n) = M_z(n-1)E_2\sin\alpha\,,\ n=1,\ldots,L$$

where $E_1=\exp(-TR/T1)$, $E_2=\exp(-TE/T2)$, α is the excitation angle and L is the total number of slice encodings. Use of a constant flip angle results in transverse magnetization that decays towards its final steady-state value. This signal modulation which appears along slice encoding (in 3D TFEPI acquisition) will result in blurring. To correct for this modulation, the flip angles need to be adjusted such that the magnetization stays approximately constant across all slice encodings. This is achieved if the flip angle is calculated as $\alpha(n) = \sin^{-1}(\sin\alpha(0)/(E_1\sin\alpha(0) + (1-E_1)\tan\alpha(n-1)))$ [3]. A particular final flip angle can be targeted and the flip angles calculated within a few iterations. Figure 1 shows one such flip angle train calculated using gray matter T1/T2 values, a final flip angle of 20° and 31 slice encodings. Figure 2 shows the transverse magnetization for the constant flip angle (30°) case and for the modulated flip train. Magnetization remains fairly constant for the modulated flip angle train. The PSF of $M_{xy}(n)$ (Figure 3) clearly shows the blurring that results from using the constant flip angle train. Simulations show that blurring is least for smaller maximum flip angle. To fix the maximum angle of the modulated flip angle train, the weighted sum (linear weighting with maximum at center of k-space) of the blood signal is plotted in Fig. 4. From the plot, a maximum flip angle ~20° was chosen.

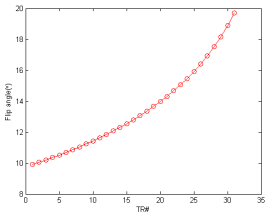

Fig (1)

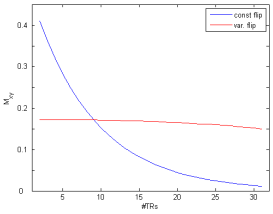

Fig (2)

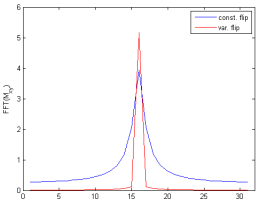

Fig. (3)

MRI Scanning: Five healthy volunteers were scanned under an IRB approved protocol on a Philips 3T Achieva scanner running Release 2.5.3 software. To quantify CBF using the Buxton model [4], a QUIPSSII saturation pulse of the same width as the tagging pulse for bolus cutoff τ ms after tagging was used. Scan parameters for IR-3D-PULSAR were: TR/TD//τ=2380/1800/900 ms; background suppression IR pulse TI=925ms, 62 pairs of control/label images; data acquisition: 3D TFEPI with 24 slices (31 encoding steps with 3D oversampling), 4mm slice thick., 80×80 matrix, SENSE factor=2.5; DAC window≈590ms; scan time≈5 min, flip angle (α)=30° (constant) and modulated flip train. Quantification of CBF was done using the Buxton model [4] with a QUIPSSII like saturation pulse to give a bolus of duration 900ms. Erosion and dilation (to remove skull) followed by automated segmentation based on Otsu's algorithm available in Matlab® was also applied to all CBF maps to separate regions of high perfusion (approximating gray matter-GM) from lower perfusion (approximating white matter-WM). WM matter CBF values were not compared as they were very low and inconsistent due to much longer transit delays (~1.6s) [5, 6].

Results: Figure 4 shows four CBF image slices (same window/level) out of 24 obtained using the two different scans — constant flip angle (top row) and modulated flip angle (bottom row). As can be seen, CBF images obtained with the modulated excitation angle train appear sharper and suffer from lower partial volume effects. This is also reflected in the CBF values (in ml/100gm/min) reported in Table 1. A paired t-test between the two CBF values shows mildly significant differences ($p<0.05$) between them. The number of voxels classified as GM was also lower for the modulated flip angle case by an average of 20% indicating less blurring.

Vol.	GM CBF (const. flip)	GM CBF (mod. Flip)
1	43.5	63.5
2	69.0	77.3
3	51.6	63.0
4	55.8	62.6
5	59.1	62.9

Discussion: Extended data acquisition with multiple TRs in the non-steady state can result in substantial blurring and result in suppressed values for GM CBF as well as misclassification of gray matter CBF. Using a modulated flip angle train to account for this blurring gives higher values in GM and reduces partial volume effects.

References: [1] X. Golay et al., MRM, 2005; 53: 15-21. [2] N. Gai et al., *ISMRM*, 2006: 3486 [3] E.M. Haacke et al., MRI: Physical Principles & Sequence Design, Wiley-Liss, 1999:454-464. [4] R. Buxton, JMRI, 2005 ; 22: 723-726. [5] J. Butman et al., *ISMRM*, 2002: 1706 [6] P. van Gelderen et al., *ISMRM*, 2007: 1416.