Improving ASL using 3D bSSFP with Background Suppression and Two-dimensional GRAPPA

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Introduction In ASL, 2D EPI is commonly used to take advantage of rapid image acquisition. However, EPI images are prone to susceptibility artifacts resulting in signal drop out and image distortion. Additionally, 2D multi-slice readout introduces a slice-dependent bias due to the different slice acquisition times and makes it difficult to incorporate background-suppression (BS) technique, which improves stability of ASL signal. 3D acquisition schemes such as GRASE and FSE-spiral have been proposed to overcome these challenges [1, 2]. However, these methods suffer from undesirable imaging blurring in slice direction due to T_2 decay during a long echo train. An alternative 3D approach is using balanced steady-state free precession (bSSFP) imaging [3, 4]. The method is known for high SNR efficiency and suitable for a fast, distortion-free, and high-resolution imaging. Moreover, it is a 3D sequence and can be incorporated with BS. In this study, we investigated the characteristics of a 3D bSSFP sequence in ASL. We developed a BS-pCASL bSSFP with 2D (in K_y and K_z) GRAPPA acceleration. The results were compared with conventional 2D EPI ASL.

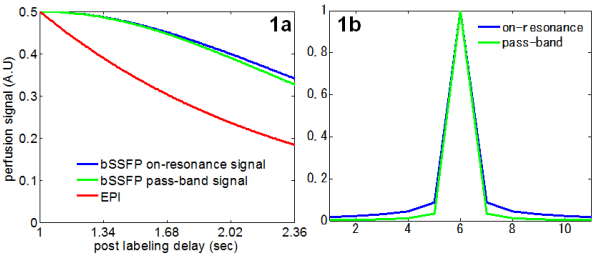

Figure 1. a) Simulation of perfusion signal changes in bSSFP (blue, on-resonance; green: pass-band) and EPI readouts. b) Point-spread function for bSSFP on-resonance and pass-band

Methods A computer simulation was performed to investigate the characteristics of perfusion signal in bSSFP. In a tag condition, perfusion signal was assumed to recover over time as a function of exponential recovery [5]. In a control condition, no signal change was assumed. Then the two signals were subtracted to show perfusion signal decay over time (Fig. 1a, red line). The signal change was plotted from 1 to 2.36 sec of post-label delay (Fig. 1a). Block equation was solved for bSSFP. Signal characteristics of on-resonance spin and pass-band (60% of central off-resonance spins) are plotted (Fig. 1a, blue and green lines respectively). A point spread function in the slice direction was also simulated (Fig. 1b). Data acquisition was performed on a Siemens 3T using a 32-channel head coil. To reduce acquisition time in bSSFP, 2D GRAPPA acceleration was developed. In order to establish an optimal method, we acquired fully encoded data and simulated GRAPPA reconstruction with various acceleration factors (2×2, 3×2 and 4×2) and kernel sizes (2×3, 4×5, 2×3×3, 4×5×3, 2×3×2, 3×2×2 and 4×5×4). Our simulation suggested a 2D acceleration up to factor 6 (y-direction = 3, z-direction = 2; kernel = 3×2×2) provides a moderate increased in RMS errors and g-factors while substantially reducing the acquisition time. The labeling and control RF duration was 1.5 s (FA = 25°, pulse = 0.5 ms, gap = 0.36 ms, and $G_{z,max}$ = 0.6 G/cm) with post-labeling delay of 1 s. For BS, two hyperbolic secant pulses were used at the inversion times of 1.8 s (TI1) and 0.7 s (TI2), respectively [1]. The pCASL labeling was placed in between the two BS pulses. bSSFP images were acquired with centric phase encoding order in K_z. The scan parameters for bSSFP were: TR/TE = 3.4/1.7 ms, FA = 30°, BW = 651 Hz/pixel, matrix = 64×64×40, and FOV = 240×240×200 mm^3. The total bSSFP time was 1.46 sec. For a fast convergence to a steady-state, linear ramp up RFs were used. TR for ASL was 4 sec (i.e. a 3D volume was acquired every 4 sec). For a comparison, bSSFP with no BS was also acquired. A higher resolution bSSFP ASL (96×96×40) was also acquired. For 2D EPI ASL, the following parameters were used: TR/TE = 4000/17 ms, FA = 90°, BW = 1502 Hz/pixel, slice thickness = 5mm, matrix size = 64×64, FOV = 240×240 mm^2, slice number = 24, and GRAPPA factor = 2. For both bSSFP and EPI, the total acquisition time was approximately 5 min acquiring 40 label/control pairs. The spatial SNR was computed as the mean of control images measured in ROI covering whole brain over the standard deviation measured in the background area. The degree of BS was calculated as the ratio of the spatial SNR of the suppressed to the unsuppressed bSSFP mean control images. The temporal SNR (tSNR) of each voxel was calculated as the temporal mean of whole brain perfusion images divided by the standard deviation of voxel time series. Percent signal change of perfusion was calculated as the signal difference between labeling and control images and divided by the control images in gray matter. An fMRI experiment using visual stimulation was performed using bSSFP and EPI. The acquisition period of 70 scans of functional images consisted of 10 epochs, in which 8 scans (64 seconds) of visual stimulation by a flashing checkerboard alternated with 6 scans (48 seconds) of rest condition.

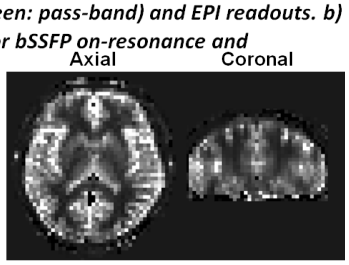

Figure 2. bSSFP perfusion images

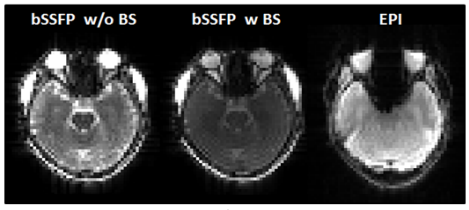

Figure 3. Images from bSSFP and EPI

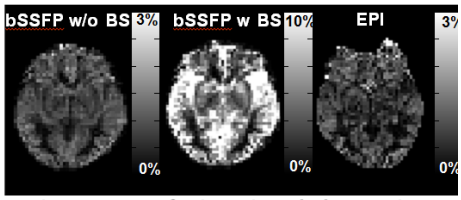

Figure 4. Perfusion signal change in %

Results Figure 1a demonstrated the simulation results of perfusion signal change in bSSFP and EPI. The contribution of stimulated echo signals in bSSFP prolonged perfusion signal. This effect substantially increased the perfusion contrast even at 2.36 sec (70% at bSSFP vs. 40% at EPI). Hence bSSFP allows us to acquire perfusion contrast longer than other acquisition schemes. Since bSSFP is acquired with 3D centric encoding, this prolonged contrast is further weighted by k-space ordering and generates high perfusion signal across all slices. Figure 1b shows the point spread function in the slice direction demonstrating little blurring. This simulation result is confirmed by the coronal perfusion image (Fig. 2). Figure 3 shows the image quality comparison between bSSFP and EPI revealing less distortion in bSSFP. When the BS bSSFP image is compared to bSSFP image without BS, the signal intensity was reduced by 75%. Table 1 summarizes tSNR and percent signal change between the different scans. The BS-bSSFP showed much higher percent signal change of perfusion (Figs. 4 and 5). tSNR was higher in bSSFP as compared to EPI. Hence, bSSFP with BS demonstrated a higher contrast to noise ratio (CNR) than bSSFP with no BS and EPI. In fMRI study, the bilateral visual cortex activation was observed (Fig.6, uncorrected P = 0.005 with cluster size > 50). ASL-bSSFP fMRI provides a wider functional activation area than ASL-EPI fMRI showing a promise of bSSFP in ASL.

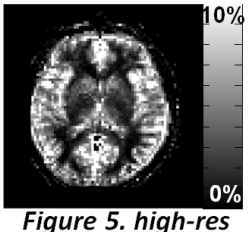

Figure 5. high-res

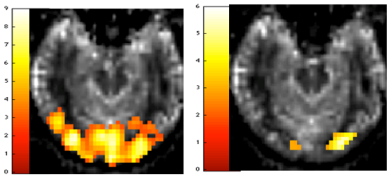

Figure 6. fMRI results. left: bSSFP with BS, right: EPI

Discussion In this study, we demonstrate the advantages of a 3D BS pCASL-bSSFP sequence. It provides high quality data while keeping the acquisition fast and efficient. The prolonged perfusion contrast from the stimulated echo allows us to acquire the contrast longer helping to increase SNR. The slow bSSFP signal decay as compared to T_2 decay in other 3D methods, induces little blurring in the slice direction. The highly accelerated acquisition enabled us to use the method for a single shot (in terms of ASL tagging) ASL suitable for fMRI. Lastly, this study demonstrated the robustness of 3D ASL-bSSFP fMRI with BS for detecting brain activity.

Table 1

	bSSFP No BS 64×64	bSSFP BS 64×64	bSSFP BS 96×96	EPI 64×64
Percent signal change %	1.11	11.1	9.60	0.9
tSNR	2.77	3.01	2.14	1.55

References: [1] Fernandez-Seara, HBM, 2007. [2] Nielsen, MRM 2012 [3] Park, ISMRM 2010 [4] Yan, MRM, 2012 [5] Chen, Magma 2012