A novel method of increasing the contrast to noise ratio of phase images using balanced SSFP

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Introduction

Balanced SSFP (bSSFP) has particular advantages over GRE [1, 2]. However, with few exceptions [2, 3], most of the bSSFP studies have utilized only the image magnitude and ignored the image phase. Recently at high field (≥ 7 Tesla), contrast based on resonance frequency shifts (or phase in gradient-echo MRI) has allowed improved visualization of fine-scale structures in brain [4]. To take the advantage of both bSSFP and phase contrast imaging, we propose a new method that substantially improves the contrast to noise ratio (CNR) of phase image, allowing the visualization of cortical laminar structures *in-vivo* at 3 Tesla.

Methods

In bSSFP, both magnitude and phase of the signal depend on the local resonance frequency. An example of this dependency is shown in Fig 1 (TR = 10 ms, TE = 5 ms, flip angle (FA) = 3°, T1 = 1500 ms, T2 = 70 ms). Here we will focus on the sharp phase transition near the on-resonance frequency (the gray area in Fig.1). The slope of this phase transition is very steep (132° over 10 Hz) and, therefore, can create a large phase difference for a small frequency difference. For example, if white and gray matter have 1.5 Hz frequency difference (e.g. due to magnetic susceptibility and molecular exchange), the resulting phase difference can reach up to 36° regardless of an echo time. The same phase shift would require a 67 ms echo time in GRE. Hence, the combination of this large phase amplification and the SNR efficiency of bSSFP may result in higher CNR compared to GRE. Note that the shape of this phase transition is independent of T1, TR, TE and flip angle and it is only affected by T2. A small flip angle (2~3°) can result in a maximum SNR at the phase transition [2]. It is important to get a good shimming over an imaging region or ROI since the phase transition exists only in a narrow frequency band.

To demonstrate the advantage of the new method, the CNRs of gray-white matter phase contrast in bSSFP and GRE were compared. For a fair comparison, scan parameters were optimized in each sequence considering (1) total scan time, (2) acquisition bandwidth and acquisition window length, (3) TE, (4) FA, (5) excitation slice profile, and (6) readout duty cycle. Two 3D GRE scans and one 3D bSSFP scan were performed. Common parameters were FOV = 20 x 15 cm^2, and resolution = 0.5 x 0.5 x 2 mm^3. For bSSFP, TR = 10 ms, TE = 4.5 ms, BW = ±31.25 kHz, FA = 3°, number of slice = 40, and total scan time = 2 min. The echo time and BW were chosen to use all available readout time within a TR to maximize SNR. The 3° FA gave the maximum intensity around on-resonance. The first GRE scan was performed using a scan time of 2 minutes (matching the scan time of bSSFP), which allowed only 6 slices to be scanned To avoid potential slice profile degradation, an additional 10-minute GRE scan was performed with 30 slices. In both GRE scans, the echo time was set to 45 ms, matching the T2* of the gray and white matter. The duration of the readout acquisition window (42.4 ms, BW = ±4.72 kHz) was also matched to the T2* time to avoid excessive image blurring at a longer readout. FA was set to an Ernst angle (18°) and TR was 66.8 ms. The readout duty cycle of GRE was 63.5% which was close to that of bSSFP (64%). 3D scans for GRE was used and preferred to 2D scan (both have similar SNR) because the 3D scans were easy to match the slab excitation profile to the 3D SSFP scan.

A high resolution bSSFP data was also acquired to demonstrate usefulness of the new method. The scan parameters were FOV = 15 x 15 cm^2, resolution = 0.3 x 0.3 x 2 mm^3, TR = 9.4 ms, TE = 4.3 ms, BW = ±50 kHz, flip angle = 3°, number of slice = 20, number of repetition = 6, and total scan time = 10 min.

For all studies, primary visual cortex area was targeted and linear shimming (including B0) was performed around the area.

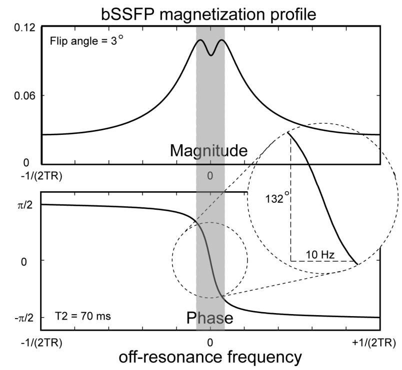

Fig. 1 bSSFP magnitude (upper) and phase (lower) profile. Phase amplification can be observed in shaded frequency band

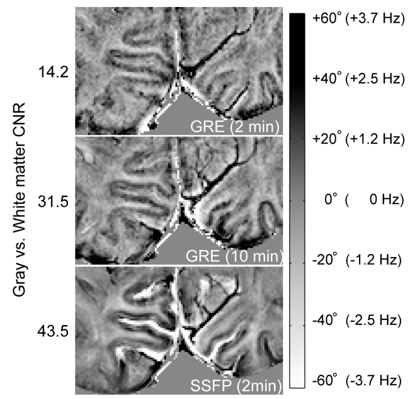

Fig. 2 CNR comparison between GRE and bSSFP: bSSFP shows 3 times higher CNR compared to GRE for the same scan time.

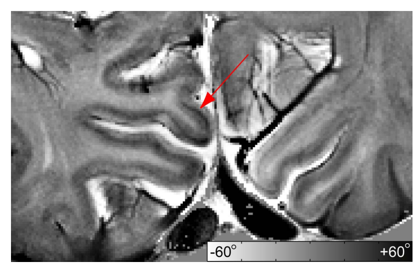

Fig. 3 High resolution result at 3 Tesla (0.3 x 0.3 x 2 mm^3). A distinguishable layer (red arrow) is visible within cortical gray mater.

Results

The bSSFP phase contrast image shows superior gray-white matter CNR (43.5) compared to those of GRE (14.2 for 2 min, 31.5 for 10 min). Per unit scan time, the bSSFP phase image gave 3.1 times larger CNR. The gray-white phase contrast at the echo time 4.5 ms was 26° in bSSFP which was slightly larger than that of GREs (23° and 24°) at the echo time of 45 ms. This is 11 times larger contrast compared to GRE of the same echo time demonstrating the phase amplification effect at the transition band of bSSFP. In certain areas, the phase contrast of bSSFP disappeared due to imperfect shimming, however, a relatively large area was covered by using linear shims. The high resolution result (Fig. 3) demonstrates a detail structure of the visual cortex revealing the Genari line within gray matter (red arrow).

Discussion and Conclusion

Here we demonstrated a new method of generating phase contrast using bSSFP. The method significantly improved the phase contrast to noise ratio compared to GRE. A few interesting features can be observed in the bSSFP phase images. First, CSF is brighter than in GRE. This is because the slope of the phase transition is very steep for long T2 species such as CSF. Second, large veins are thinner in bSSFP than in GRE. This may be due to the long acquisition window in the GRE scans yielding a wider point spread function although subject motion may have affected the slice location. One limitation of the method is the spatial and temporal resonance frequency variation that may limit the spatial coverage and may hamper stability of the method. The temporal variation from the scanner drift and subject respiration can be compensated using a real time method [5]. The spatial coverage can be improved by using high order shims, local shim coils and better shim algorithms [6]. Combined with these, a parallel excitation can be used to compensate for the spatial B0 inhomogeneity [7].

References [1] Scheffler, Eur Radiol, 2003, p2409, [2] Miller, MRM, 2003, p675, [3] Lee, MRM, 2007, p905, [4] Duyn, PNAS, 2007, p11796, [5] Lee, MRM, 2006, p1197, [6] Lee, MRM, 2009, p1500, [7] Heilman, ISMRM, 2009, p251