Midbrain nuclei visualization improved by susceptibility-enhanced 3D multi-echo SSFP for deep brain stimulation guidance

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Introduction

Deep brain stimulation (DBS) is an effective therapy for treating symptoms related to Parkinson's disease. DBS electrodes are implanted in the subthalamic nucleus (STN) using guidance from preoperative MRI registered to patient's brain. Because several passes/adjustments may be needed to achieve optimal electrode placement and each pass or manipulation carries some risk and lengthens the procedure, a method for direct and robust visualization of midbrain nuclei is desirable in order to improve the accuracy of STN targeting on the first pass [1,2]. Due to the higher iron content in midbrain nuclei, such as red nucleus (RN) and substantia nigra (SN), T2- and T2*-weighted MRI have been used for imaging these regions. To provide a high contrast between midbrain nuclei and surrounding white matter in T2*-weighted imaging, the TE generally needs to be set to 20ms or longer. As a result, the scan time is 8-10 seconds per slice for gradient-echo imaging with matrix size of 256x256 [3]. In this study, we use 3D multi-echo steady-state free precession (ME-SSFP) technique for an improved visualization of midbrain nuclei, aiming at achieving high contrast-to-noise ratio with a reduced scan time. Because SSFP makes use of magnetization vectors excited from previous RF excitations, a heavy T2*-weighting can be achieved with a relatively short TE (< 10ms). In addition, the developed ME-SSFP technique enables banding-free image reconstruction, while conventional balanced SSFP (bSSFP) imaging is highly susceptible to banding-artifact, particularly in brain regions with pronounced susceptibility field gradients. To further improve the contrast-to-noise ratio in deep brain nuclei we integrate the ME-SSFP acquisition with susceptibility-weighted imaging (SWI) reconstruction [1,4]. The resulting integrated ME-SSFP SWI method enables robust visualization of midbrain nuclei with scan times as low as 4 seconds per slice.

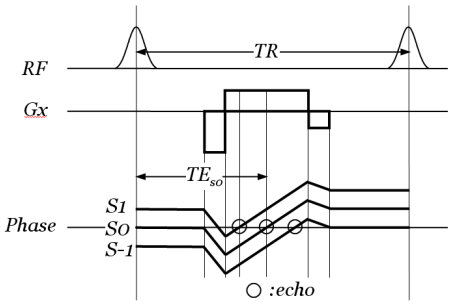

Figure 1. ME-SSFP with unbalanced gradients creates constant nonzero net phase dispersion in each TR cycle. In this study, three echoes are sampled corresponding to steady-states S1, S0 and S-1.

Material and method

In ME-SSFP, the intentionally unbalanced waveform results in constant nonzero net phase dispersion for transverse magnetization during each TR, as shown in figure 1. Through consecutive RF excitations in SSFP, multiple steady states of magnetization are generated, with the phase dispersion in each steady-state equal to integer multiples of the phase dispersion in one TR [5]. In this study, a 3D SSFP sequence was modified to sample three steady-states Sn, n=1,0,-1, where n denotes the integer multiple of total phase dispersion caused by gradient in each TR. Note that images generated from the S0 signal is equivalent to FISP (fast imaging steady precession) and from the S-1 signal to PSIF (reversed FISP). The SWI reconstruction was performed for images from each echo during post-processing. Composite images were then generated by calculating the root mean square (RMS) of the SWI-enhanced echo images. The developed technique was evaluated on healthy volunteers in a 3 Tesla scanner (GE, Milwaukee, WI, USA). Example images shown here were acquired with TE = 8ms, matrix 256x256x20 and scan time of 1 min 22 sec.

Results and discussion

Figures 2a, b and c show axial-plane ME-SSFP images at midbrain level, corresponding to S1, S0 and S-1, respectively. Although S0 showed highest intensity, S1 and S-1 presented stronger T2*-weighting because of their greater signal composition from previous TR cycles. As shown in Figure 2d, the composite image (generated from data shown in a-c) improves resolvability of the midbrain nuclei. By incorporating phase information into magnitude image, the SWI enhanced images (Figure 2e) further improves the image contrast. Figure 3 demonstrated the absence of SSFP banding on the ME-SSFP images.

In this study, 3D ME-SSFP was used to achieve stronger T2*- weighting in a shorter scan time, as compared with gradient echo techniques. The major advantage of ME-SSFP is improved image contrast resulting from signal accumulation over multiple TR cycles. Similar previously reported techniques based on acquisition of S0 and S-1 for cartilage imaging are referred to as double echo steady state (DESS) [6]. We report a more general ME-SSFP method using three echoes to increase sensitivity to iron deposition and thus enable a robust visualization of midbrain nuclei. As shown in this study, SWI processing can be employed to incorporate phase information for enhanced tissue contrast. ME-SSFP is free from banding artifact because complex destructive interference among echoes is avoided by RMS composite image. Although more echoes could be included to obtain even stronger T2*-weighting, it has been shown that signal decays exponentially for higher order echoes, which therefore do not offer sufficient signal-to-noise ratio [5]. We conclude that susceptibility enhanced 3D ME-SSFP offering robust visualization of midbrain nuclei at a reduced scan time folds promise to improve stereotactic guidance of deep brain stimulation.

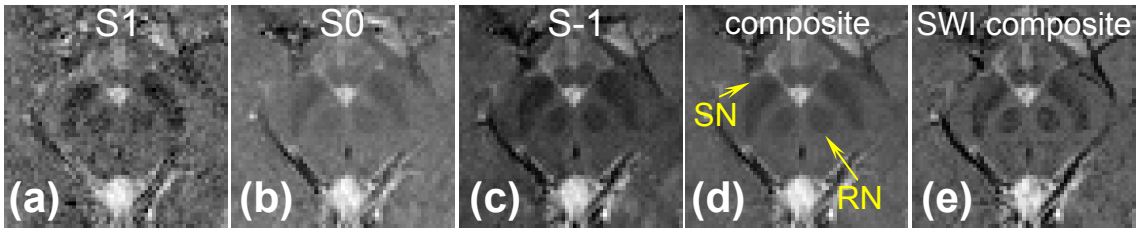

Figure 2. 3D multi-echo SSFP for visualization of midbrain nuclei. Images corresponding to steady-states S1, So, and S-1 are shown in (a), (b) and (c), respectively. The composite image calculated from (a)-(c) is shown in (d) and SWI enhanced composite image is shown in (e). Note that although S0 has highest signal intensity, S-1 and S1 have heavier T2-weighting. The composite image has an improved resolvability for red nucleus (RN) and substantia nigra (SN). SWI reconstruction further improves the image contrast.*

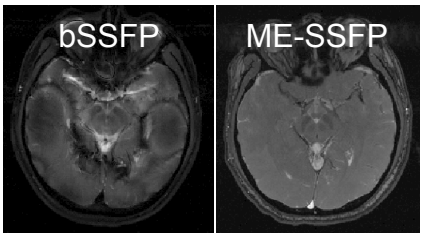

Figure 3: Left: bSSFP is affected by banding-artifact; Right: the developed ME-SSFP is free from banding-artifact.

Reference

[1] Young G.S., et al., Neurosurgery 65(4): 809-15, 2009 [2] Vertinsky A.T., et al., Am J Neuroradiol 30:1717-24, 2009 [3] Elolf E., et al., Am J Neuroradiol 28:1093-94, 2007 [4] Haacke E.M., et al., Magn Reson Med 52:612-18 ,2004 [5] Scheffler K., Concepts Magn Reson 11(5): 291-304, 1999 [6] Moriya S., et al., Acta Radiol 50(5):507-11, 2009