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Original contribution

Fast B_1^+ mapping using three consecutive RF pulses and balanced gradients for improved bSSFP imaging

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ABSTRACT

Purpose: To develop a B_1^+ mapping during the transient phase of balanced steady state free precession (bSSFP) imaging which can be used for subsequent B_1^+ inhomogeneity compensation.
Methods: Two images with different flip angles (FA) are acquired using single-shot spiral technique during the transient phase of bSSFP with three consecutive RF pulses and balanced gradients. Under the assumptions that the transmit (B_1^+) field varies slowly in spatial domain and T_1 and T_2 relaxation effects are negligible during 2·TR, B_1^+ was estimated using the two magnitude images and bSSFP data was sequentially acquired. B_1^+ estimation error due to the assumptions and other factors such as FA and off-resonance were assessed using Bloch simulation. Phantom and in vivo experiments were performed with α-2α-3α scheme.
Results: The simulation results indicated that the proposed method was less sensitive to T_1 relaxation and B_1^+ mapping FA (α) of approximately 60° produced minimum estimation error. The B_1^+-induced intensity variation was reduced with the proposed method in the phantom experiment. For both the phantom and in vivo experiments, the estimated B_1^+ map showed comparable to the conventional B_1^+ map using spin-echo DAM.
Conclusion: B_1^+ map was estimated during the transient phase of bSSFP and subsequently compensated bSSFP images. There was no scan time increment and hence the technique can be used in a prescan manner for B_1^+ mapping or shimming.

1. Introduction

Balanced steady-state free precession (bSSFP) is routinely performed in clinical imaging such as cardiac cine, angiography, and abdominal imaging [1,2]. The bSSFP imaging technique has distinct advantages such as high signal-to-noise ratio (SNR) efficiency compared to conventional spoiled gradient echo (SPGR), and high contrast-to-noise ratio (CNR) especially between flow and stationary tissues [3]. However, it is sensitive to B_0 inhomogeneity. Banding artifacts occur where the phase accumulation due to field inhomogeneity during the TR period is near integer multiples of 2π. Furthermore, B_1^+ inhomogeneity also causes spatially dependent signal variations since the T_2/T_1 contrast in bSSFP is influenced by factors including the flip angle (FA) [4–6]. In clinical applications of bSSFP imaging, unwanted signal modulations due to non-uniform B_1^+ distribution have been reported as a potential error source in field strengths of 3 T or higher [7–11].

The spatial distribution of B_1^+ depends on the main magnetic field strength which determines the resonance frequency and the wavelength of applied RF pulse [12,13]. Moreover, the size and electrical properties of the imaging object also affect the B_1^+ distribution. Thus, the B_1^+ map is patient dependent and needs to be estimated for each subject. B_1^+ information can be useful for many applications e.g. correcting B_1^+ inhomogeneity induced intensity distortions in quantitative studies [14–16], determining local specific absorption rate (SAR) [17], shimming transmit field in parallel transmit system [18–20], or estimating electrical properties [21,22]. B_1^+ map estimation using relationships between signals acquired at different FAs is often used. Among these methods, the double angle method (DAM) is commonly acknowledged as one of the gold standard [13] although this technique requires a long TR thus limiting clinical usage. Several B_1^+ mapping methods based on the SSFP signal have also been proposed [23,24]. Recently, a B_1^+ mapping method using unbalanced SSFP imaging was introduced [25]. Here, the frequency of signal oscillations in the transient phase of SSFP was used and MR spectroscopic analysis was adapted to extract the oscillation frequency. In this technique, tissue relaxation related errors were ignorable due to the short TR acquisition. Furthermore, B_1^+ mapping was particularly insensitive to off-resonance for 2D multi-slice imaging. Since the total acquisition time increases due to the acquisition of more accurate oscillation patterns, it had a trade-off between speed and accuracy. Except for [25], these mapping methods require

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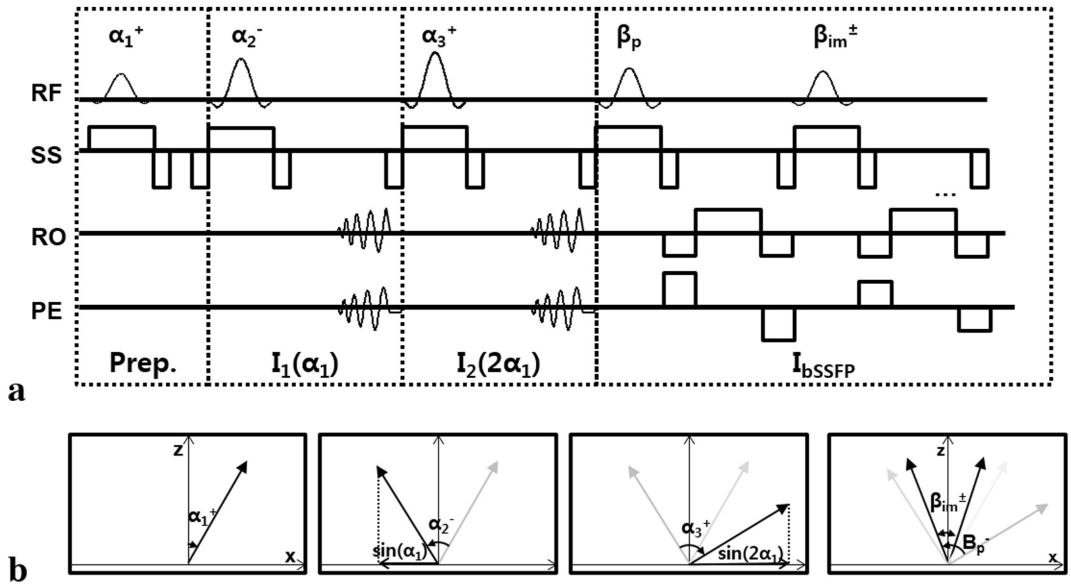

Fig. 1. (a) A pulse sequence diagram of the proposed B_1^+ estimation ($|\alpha_2| = 2|\alpha_1|$ and $|\alpha_3| = 3|\alpha_1|$). (b) Illustration of magnetization changes corresponding to each sequence block in (a).

scan times from few seconds to minutes per slice.

In this paper, we propose a fast B_1^+ mapping method using magnitude information from the early stages of the transient phase in bSSFP. Signal evolution in bSSFP transitions from the transient phase to the steady-state. During the transient phase, the amplitude of signal is governed by factors such as FA, T_1, T_2 and ΔB_0. These transient phase signals are discarded or used to fill the periphery of k-space to increase time efficiency. Here, the transient phase is used to acquire B_1^+ maps. For B_1^+ mapping, the FA of the initial RF pulses are modified and images are acquired with a single-shot spiral readout applied for efficient spatial coverage. A bridge pulse is inserted for smooth transition to the following bSSFP image acquisition stage. Bloch simulations are performed to assess the performance of the proposed B_1^+ mapping technique due to parameters such as T_1, T_2, FA, off-resonance and slice profile. The B_1^+ mapping is demonstrated using phantom and in vivo studies. Subsequent bSSFP image compensation using these measured B_1^+ maps are demonstrated.

2. Method

Fig. 1 shows the proposed sequence and magnetization vectors. In order to enable B_1^+ mapping during the transient phase, the conventional bSSFP sequence is modified such that it consists of 4 blocks; preparation, I_1, I_2, and imaging block. In the preparation block, an α_1 pulse (corresponding to the general α FA used in DAM) is introduced to prepare the signals for the B_1^+ mapping and to avoid the undesired signal oscillations due to off-resonance. This is followed by two opposite-phase pulses, α_2 and α_3, in the I_1 and I_2 blocks respectively for the double angle acquisition. Finally, a β_p pulse is applied as a bridge pulse followed by β_{im} pulses which are used for normal imaging. Therefore, the approach can be thought of as a B_1^+ mapping using three consecutive RF pulses with balanced gradients. The magnetization prior to α_2 can be expressed,

$$\mathbf{M}_1^- = \mathbf{R}_z(\theta)\mathbf{E}_{TR/2}\mathbf{R}_y(\alpha_1)\mathbf{M}_0 + (1 - E_1)\mathbf{M}_0, \quad (1)$$

where $\mathbf{R}_z(\theta)$ is a rotation by an off-resonance dephase angle θ during TR / 2 around z-axis, $\mathbf{E}_{TR/2}$ is a relaxation matrix during TR / 2, $\mathbf{R}_y(\alpha_1)$ is a rotation by α_1 around y-axis representing the RF pulse and E_1 is $e^{-TR/2T1}$. Thus, the magnetizations M_1 and M_2 at TE for I_1 and I_2 are,

$$\mathbf{M}_1 = \mathbf{R}_z(\theta)\mathbf{E}_{TR/2}\mathbf{R}_y(-\alpha_2)\mathbf{M}_1^- + (1 - E_1)\mathbf{M}_0 \quad (2)$$

$$\mathbf{M}_2 = \mathbf{R}_z(\theta)\mathbf{E}_{TR/2}\mathbf{R}_y(\alpha_3)\mathbf{M}_2^- + (1 - E_1)\mathbf{M}_0 \quad (3)$$

Here, $\mathbf{M}_2^-$ represents the magnetization prior to α_3 which can be derived by substituting $\mathbf{E}_{TR}$ instead of $\mathbf{E}_{TR/2}$ in Eq. (2). If the relaxation and off-resonance terms are ignored, the transverse magnetizations at I_1 and I_2 are,

$$M_{1xy} = M_0 \sin(\alpha_1 - \alpha_2) \quad (4)$$

$$M_{2xy} = M_0 \sin(\alpha_1 - \alpha_2 + \alpha_3). \quad (5)$$

The effect of neglecting relaxation and off-resonance terms are considered in the simulations done below. According to Eqs. (4), (5), the FAs α_2 and α_3 can be applied in different ways such that their signals are proportional to $\sin(\alpha_1)$ and $\sin(2\alpha_1)$ or vice versa. As an example, one can let $|\alpha_2| = 2|\alpha_1|$ and $|\alpha_3| = 3|\alpha_1|$ which results in $I_1 \propto \sin(\alpha_1)$ and $I_2 \propto \sin(2\alpha_1)$ (Fig. 1b). Another example is to let $|\alpha_2| = 3|\alpha_1|$ and $|\alpha_3| = 3|\alpha_1|$which results in $I_1 \propto \sin(2\alpha_1)$ and $I_2 \propto \sin(\alpha_1)$. In terms of Fig. 1a, β_p corresponds to $2\alpha_1 + \beta_{im}/2$ and $\alpha_1 + \beta_{im}/2$ for the two example cases respectively. Fig. 1b shows the magnetization changes during the individual sequence blocks for the case when $|\alpha_2| = 2|\alpha_1|$ and $|\alpha_3| = 3|\alpha_1|$. In this study, the relationship for the FAs study were $\alpha_1 = \alpha$, $\alpha_2 = -2\alpha_1$, and $\alpha_3 = 3\alpha_1$. Since the magnitude signals of I_1 and I_2 are proportional to $\sin(\alpha_1)$ and $\sin(2\alpha_1)$ respectively (Fig. 1b), the flip angle distribution can be estimated by,

$$\alpha_{actual} = \cos^{-1}(I_2/2I_1). \quad (6)$$

Meanwhile, as time elapses, the transient phase signal is affected more heavily by T_1 and T_2 relaxation times. In order to minimize the relaxation effect and off-resonance, TR is shortened as much as possible which is also desired in bSSFP to avoid banding artifact. In addition, a single-shot spiral k-space acquisition is used for the efficient acquisition of the I_1 and I_2 signals. Therefore, the proposed fast B_1^+ mapping works best with the assumptions of 1) B_1^+ varies slowly in spatial domain; a low resolution image can represent B_1^+ distribution, and 2) T_1 and T_2 relaxations are negligible if $TR \ll T_1, T_2$; FA dominates overall signal in the early-stage of transient phase in bSSFP. Since balanced gradients are used, the approach will be sensitive to B_0 inhomogeneities. Simulations are performed to evaluate this sensitivity.