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Nonlinear averaging reconstruction method for phase-cycle SSFP

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Abstract

The ability to obtain high-quality images of small structures, such as the nerves of the inner ear, is important for the early diagnosis of numerous conditions. Balanced steady-state free precession (SSFP; e.g., true fast imaging with steady-state precession) is a fast acquisition method, but its use has been limited by the presence of off-resonance banding artifacts. To reduce these artifacts multiacquisition balanced SSFP with phase cycling is used, yielding multiple data sets in which the banding artifacts are spatially shifted with respect to each other (e.g., as in CISS). We present a new method, called nonlinear averaging (NLA), for combining these data sets to reduce banding artifacts. The NLA method arithmetically averages the three highest magnitude signals from four-phase-cycle SSFP data on a pixel-by-pixel basis. Simulations indicate that NLA offers improved signal-to-noise ratio (SNR) over the more standard maximum intensity projection (MIP) reconstruction.

NLA is compared to MIP in simulations and volunteer tests. Simulations suggest that NLA provides substantially improved SNR compared to MIP. In a randomized blinded comparison of 10 volunteer studies, two radiologists found that NLA, compared to MIP, gave improved results. NLA also provided superior noise reduction and enhanced edge sharpness compared to MIP. We demonstrate that NLA, similar to MIP, improves SNR and image quality. It does so consistently in all situations to which it is applied.

Keywords: Phase cycle; SSFP; Banding artifact; Postprocessing; Artifact reduction; Cochlear nerves

1. Introduction

Balanced steady-state free precession (SSFP) pulse sequences, such as true fast imaging with steady-state precession and fast imaging employing steady-state acquisition (FIESTA), allow the rapid acquisition of images with a high signal-to-noise ratio (SNR), with relative immunity to flow artifacts and good contrast between the cerebrospinal fluid (CSF) and other structures such as nerves and bone. This allows for excellent depiction of structures within the internal auditory canal (IAC), particularly the facial nerve (FN), cochlear nerve (CN), superior vestibular nerve (SVN) and inferior vestibular nerve (IVN) [1,2]. The visualization of these structures is useful for the clinical diagnosis of acoustic neuromas as it allows an accurate measurement of tumor size [3]; other inner ear conditions, such as fibrous dysplasia, cholesterol granuloma and metastasis, can also be readily visualized [4]. This sequence is also valuable in detecting and characterizing inner ear malformations in patients with sensorineural hearing loss [5]. Other uses of SSFP include the evaluation of spinal cysts (syringomyelia) [6] and the study of hydrocephalus [7].

Balanced SSFP is prone to banding artifacts due to inhomogeneity in the main magnetic field, usually due to susceptibility variation. This has been an obstacle to the application of balanced SSFP to very small fields of view (e.g., $\leq$12 cm) that require longer T_R values (e.g., $\geq$7 ms). This problem has been overcome by acquiring a number (e.g., two or four) of balanced SSFP data sets, with each data set having its own distinct radiofrequency (RF) phase cycle. This phase cycling spatially shifts band artifacts on each image set. Different data sets are then combined to minimize band artifacts. Several methods aiming to combine image data sets have been proposed: N-point discrete Fourier transforms [8], root sum of squares (RSS) [9,10] and complex averaging [11]. Maximum intensity projection (MIP), however, has been the most widely used reconstruction method and is used in the constructive interference in steady state (CISS) method [4]. MIP

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calculates a new image using the maximum of the original magnitude images on a pixel-by-pixel basis. MIP greatly reduces banding artifacts in individual data sets, but is known to have SNR properties less optimal than averaging [12]. In this paper, we also show that MIP can accentuate a ridge of higher signals that can flank each side of the band.

We also present a new construction method for four-phase-cycle SSFP, called nonlinear averaging (NLA). NLA is compared to MIP using simulations, phantom studies and volunteer studies with four-phase-cycle data sets. To the best of our knowledge, at the time of writing, MIP is the most widely used reconstruction method, as other methods require complex data or images. We therefore restrict ourselves to comparing NLA only to MIP.

2. Methods

Given $N>1$ magnitude image data sets $M_1,\ldots,M_N$, MIP and NLA are calculated on a pixel-by-pixel basis according to:

$$MIP = \max(M_1, M_2, M_3,\ldots, M_N)$$

$$NLA = \frac{M_1+M_2+M_3+\cdots+M_N - \min(M_1, M_2, M_3,\ldots, M_N)}{N-1}. \tag{1}$$

In our experience, the most common acquisitions use either $N=2$ or $N=4$. For $N=2$, NLA and MIP are equivalent. For $N=4$, NLA is a distinct method. In this paper, we focus on $N=4$. Four-phase balanced SSFP acquires four consecutive data sets of the same anatomical site, with RF sign alternation and RF phase increment of $+90°$ and $-90°$ per T_R, respectively. The result of this phase cycling is the displacement of banding artifacts due to inhomogeneity in the B_0 field on different image sets. Provided that the inhomogeneity is not too large, a pixel that is part of a band in one image will not be part of a band in any of the other three images. The SSFP sequence used in this study is called phase-cycle FIESTA (FIESTA-C), which is implemented on a 1.5-T scanner (GE Health Care, Milwaukee WI).

2.1. Simulations

The first simulation tested the ability of MIP and NLA to suppress banding in the presence of noise. Four 256×256 images were generated. Signals from the real and imaginary signal channels, both with Gaussian noise distributions, are combined to form a magnitude image by their square root sum of squares. The noise distribution of the resulting magnitude image will be nonnegative and will follow a Rician distribution [13]. Rician noise of various magnitudes was added, as a percentage of the signal level, to all four images. Noise levels of increasing intensity were added to the four images, which were combined to form one MIP and one NLA image. The SNR for each new image was measured as a function of added noise levels (Fig. 1).

The second simulation was designed to determine how well each of the methods (MIP and NLA) deals with the structure of bands when a signal is plotted against phase precession angle β, which denotes free precession per TR. The actual structure is well known to be a complicated function of flip angle, T1, T2, TR and precession angles [14]. For example, dependence on these parameters is seen in the differing band structures of Fig. 2A (flip angle, 10°) and Fig. 2B (flip angle, 70°). Note that at 10°, V-shaped structures have high ridge structures on either side of the minima. These ridges often are signal maxima that can produce artificially high signals in the combined image. The width of the band is dependent on flip angle, with larger flip angles generally yielding higher, more uniform signals over a large part of the image. Fig. 2B illustrates the trend at higher flip angles (70°), where, apart from the band, the signal is generally uniform. The data sets were generated then combined using MIP and NLA for flip angles from 1° to 90°. Ideally, the final combined image in this simulation should be perfectly flat. The effect of ridges is to produce ripple-like structures in the final image; thus, whichever reconstruction method minimizes the ripple at a given flip angle has an advantage. The metric for measuring this ripple for various flip angles was to subtract the maximum ripple height (max) from the minimum height (min) and to divide this by the mean to give Eq. (2):

$$\text{ripple} = \frac{\max - \min}{\text{mean}}. \tag{2}$$

A third simulation tested low-contrast detectability for MIP and NLA. Four images were generated, with each image containing a set of squares laid out on a Cartesian grid. These contrast patterns were 1024×1024 pixels in size, with random noise added. None of the simulated images had any banding in this case. There were four rows and six

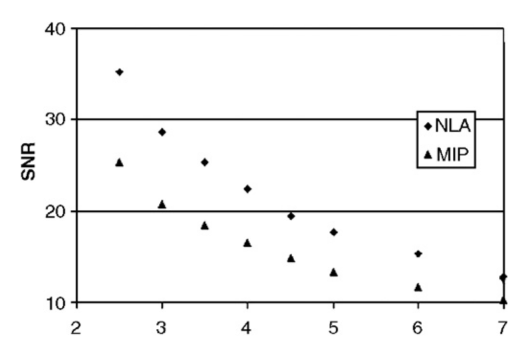

Fig. 1. SNR at the band location versus percentage noise added to the simulation for NLA and MIP reconstructed images. Noise was added in 0.5% increments until the 5% noise level had been reached and in 1% increments from then on. The horizontal axis begins at 2%; thus, a clearer separation of higher percentage points can be seen.