Available online at www.sciencedirect.com
ScienceDirect
Magnetic Resonance Imaging 25 (2007) 1138–1147
MAGNETIC RESONANCE IMAGING

Reduction of flow- and eddy-currents-induced image artifacts in coronary magnetic resonance angiography using a linear centric-encoding SSFP sequence

Xiaoming Bi[a,b], Jaeseok Park[c], Vibhas Deshpande[c], Orlando Simonetti[d], Gerhard Laub[c], Debiao Li[a,b,*]

[a]*Department of Radiology, Northwestern University, Chicago, IL 60611, USA*
[b]*Department of Biomedical Engineering, Northwestern University, Evanston, IL 60201, USA*
[c]*Cardiovascular Research Center, Siemens Medical Solutions, Los Angeles, CA 90095, USA*
[d]*Department of Cardiovascular Medicine, Ohio State University, Columbus, OH 43210, USA*

Received 23 October 2006; revised 27 December 2006; accepted 5 January 2007

Abstract

Coronary magnetic resonance angiography (MRA) acquired using steady-state free precession (SSFP) sequences tends to suffer from image artifacts caused by local magnetic field inhomogeneities. Flow- and gradient-switching-induced eddy currents are important sources of such phase errors, especially under off-resonant conditions. In this study, we propose to reduce these image artifacts by using a linear centric-encoding (LCE) scheme in the phase-encoding (PE) direction. Abrupt change in gradients, including magnitude and polarity between consecutive radiofrequency cycles, is minimized using the LCE scheme. Results from numeric simulations and phantom studies demonstrated that signal oscillation can be markedly reduced using LCE as compared to conventional alternating centric-encoding (ACE) scheme. The image quality of coronary arteries was improved at both 1.5 and 3.0 T using LCE compared to those acquired using ACE PE scheme (1.5 T: ACE/LCE=2.2±0.8/3.0±0.6, P=.02; 3.0 T: ACE/LCE=2.1±1.1/3.0±0.8, P=.01). In conclusion, flow- and eddy-currents-induced imaging artifacts in coronary MRA using SSFP sequence can be markedly reduced with LCE acquisition of PE lines.

Keywords: Magnetic resonance angiography (MRA); Coronary arteries; High-field imaging; SSFP; Flow; Eddy currents

1. Introduction

Balanced steady-state free precession (SSFP) sequences (TrueFISP, FIESTA, b-FFE and b-SSFP) have become the major method of choice for the magnetic resonance angiography (MRA) of coronary arteries at 1.5 T. Higher signal-to-noise ratio (SNR) and contrast-to-noise ratio of coronary images can be achieved compared to conventional gradient-echo sequence [1,2]. However, the balanced structure of SSFP sequence makes it sensitive to residual phase from each radiofrequency (RF) cycle. Besides those well-known sources of spin dephasing, such as local inhomogeneity of the main field, chemical shift between spectrally different components and susceptibility-induced field variations, eddy current and flow are also important sources of such phase error [3–5].

Eddy current is induced by electric fields resulting from changing magnetic flux [6]. Such current alters local magnetic field, which, in turn, increases the dispersion of spins and leads to imaging artifacts [7]. The onset of eddy current is related to the dynamic switching of imaging gradients and the electromagnetic properties of conducting components, including RF coils, RF shield and the subject being imaged. Complete compensation of such spatially and temporally varying currents using preemphasis requires careful characterization and delicate design of the applied gradient. Recent studies have shown that by improving the order of k-space data acquisition,

* Corresponding author. Chicago, IL 60611, USA. Tel.: +1 312 926 4245; fax: +1 312 896 5665.
E-mail address: d-li2@northwestern.edu (D. Li).

0730-725X/$ – see front matter © 2007 Elsevier Inc. All rights reserved.
doi:10.1016/j.mri.2007.01.004

eddy-current-induced imaging artifact can be substantially reduced [3,5,8–10].

Coronary MRA data are typically acquired during the middiastole of each cardiac cycle to minimize cardiac-motion-related artifacts. Blood flow, however, is at a relatively high velocity at this stage of isovolumic filling [11]. Numerous studies have reported the dependence of SSFP signal on flow and frequency offset [12–14]. Flow-related artifacts manifesting as low or inhomogeneous signal intensity in blood pools can be reduced by careful shimming and dedicated design of the data acquisition scheme [14,15].

To assure sufficient spatial resolution and minimal motion artifact, the full *k*-space for reconstructing a coronary artery image is typically filled by combining segmented data sets acquired from multiple heartbeats. For coronary MRA, centric-encoding order has been widely used in the phase-encoding (PE) direction for studies using Cartesian sampling scheme. By collecting central *k*-space lines at the beginning of each readout train, optimal fat saturation and minimal respiratory motion artifact (in case of navigator-gated free-breathing imaging, although breath-holding was used in this study) can be achieved [16]. Typically, data are collected from low to high spatial frequencies in both positive and negative *k*-space regions in each readout train in an oscillating mode (Fig. 1A) [2,17,18]. The sign and amplitude of PE gradients alternate in consecutive PE lines, accumulating phase errors from eddy currents and flow within consecutive RF cycles (see Theory for details). For simplicity, such PE order is referred as alternating centric-encoding (ACE) in the following text.

We propose the use of a linear centric-encoding (LCE) PE order for coronary MRA using Cartesian SSFP sequence. With such acquisition scheme, the *k*-space trajectory moves unidirectionally from center to outer lines in each segment (Fig. 1B). Not only can the advantages of ACE order be preserved by collecting central lines first but, more importantly, smooth and minimal changes of PE gradients between consecutive RF cycles (Fig. 1D) can potentially reduce imaging artifacts from flow and eddy currents. Phantom and human studies were performed in this study to validate the efficacy of such encoding scheme in both 1.5- and 3.0-T clinical scanners.

2. Theory

The net transverse magnetization of spin ensemble is a function of tissue parameters, including T_1 and T_2, imaging parameters such as the flip angle, repetition time (T_R) and the accumulated phase angle φ between RF pulses for SSFP sequence. Phase dispersion depends on the local magnetic field strength spins have experienced, in which variations originate from static field inhomogeneity, eddy-currents-induced local magnetic field and motion of spins (e.g., from flow). The static field inhomogeneity from main

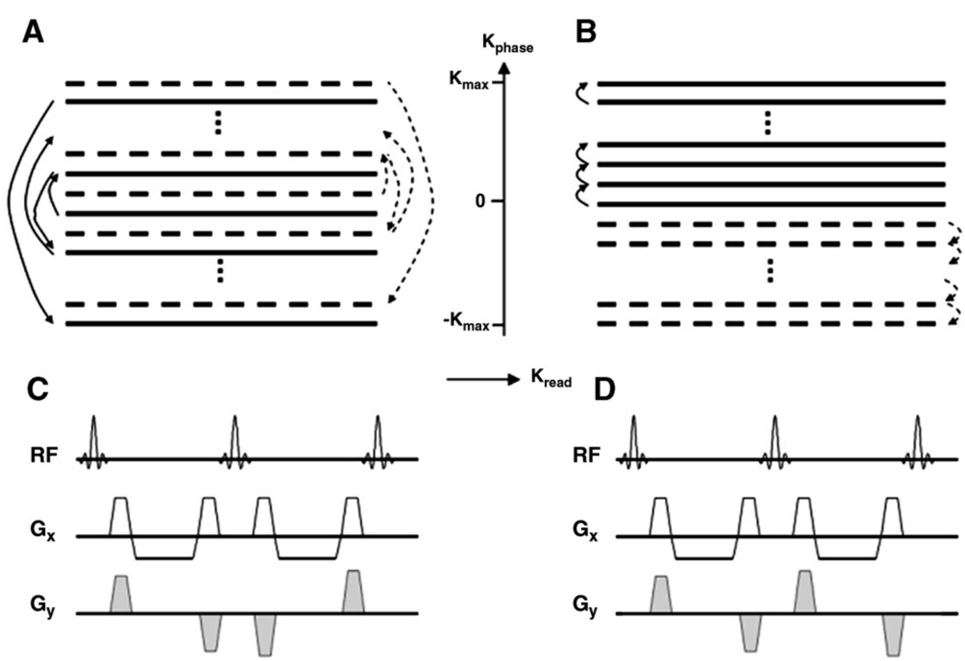

Fig. 1. Schematic of the *k*-space trajectory of conventional ACE (A) and proposed LCE (B) PE orders. The *k*-space data of one imaging slice are filled by interleaving two segments in this example. The trajectory of ACE scheme moves from central to outer lines in the PE direction and alternates between positive and negative regions in each segment, as illustrated by two segments (first: solid line; second: dashed line) in (A). Note the sign and the magnitude of the PE gradient (G_y) alternating from one RF cycle to another (C). Using LCE PE order, the *k*-space trajectory travels unidirectionally from central to outer lines in each segment, as illustrated in (B). The magnitude of G_y is slightly increased from RF cycle to another. The polarity of gradient pairs remains unchanged in the whole readout train, as shown in (D).