Magnetic Resonance Imaging 34 (2016) 1017–1025

Contents lists available at ScienceDirect

Magnetic Resonance Imaging

journal homepage: www.mrijournal.com

Accelerating dual cardiac phase images using undersampled radial phase encoding trajectories☆

Karis Letelier [a,b], Jesus Urbina [a], Marcelo Andía [a,c], Cristián Tejos [a,b], Pablo Irarrazaval [a,b], Claudia Prieto [d], Sergio Uribe [a,c,*]

[a] Biomedical Imaging Center, Pontificia Universidad Católica de Chile, Santiago, Chile
[b] Electrical Engineering Department, Faculty of Engineering, Pontificia Universidad Católica de Chile, Santiago, Chile
[c] Radiology Department, School of Medicine, Pontificia Universidad Católica de Chile, Santiago, Chile
[d] Division of imaging sciences, King's College London, London, UK

ARTICLE INFO

Article history:
Received 18 July 2015
Revised 25 March 2016
Accepted 27 March 2016
Available online xxxx

Keywords:
Whole-heart MRI
Dual phase
View sharing
Radial phase encoding

ABSTRACT

A three-dimensional dual-cardiac-phase (3D-DCP) scan has been proposed to acquire two data sets of the whole heart and great vessels during the end-diastolic and end-systolic cardiac phases in a single free-breathing scan. This method has shown accurate assessment of cardiac anatomy and function but is limited by long acquisition times. This work proposes to accelerate the acquisition and reconstruction of 3D-DCP scans by exploiting redundant information of the outer *k*-space regions of both cardiac phases. This is achieved using a modified radial-phase-encoding trajectory and gridding reconstruction with uniform coil combination. The end-diastolic acquisition trajectory was angularly shifted with respect to the end-systolic phase. Initially, a fully-sampled 3D-DCP scan was acquired to determine the optimal percentage of the outer *k*-space data that can be combined between cardiac phases. Thereafter, prospectively undersampled data were reconstructed based on this percentage. As gold standard images, the undersampled data were also reconstructed using iterative SENSE. To validate the method, image quality assessments and a cardiac volume analysis were performed. The proposed method was tested in thirteen healthy volunteers (mean age, 30 years). Prospectively undersampled data ($R = 4$) reconstructed with 50% combination led high quality images. There were no significant differences in the image quality and in the cardiac volume analysis between our method and iterative SENSE. In addition, the proposed approach reduced the reconstruction time from 40 min to 1 min. In conclusion, the proposed method obtains 3D-DCP scans with an image quality comparable to those reconstructed with iterative SENSE, and within a clinically acceptable reconstruction time.

1. Introduction

Assessment of cardiac anatomy and function using magnetic resonance (MR) imaging is still mainly based on the acquisition of multiple two dimensional (2D) cine images using balanced steady state free precession (b-SSFP) sequences [1]. This approach has several disadvantages, including the need of expertise to plan different views of the heart; the possibility of slice misalignment; and the need of several breath-holds, which makes it difficult to apply this approach in children.

Isotropic non-angulated three dimensional (3D) cardiac MR can overcome some of these problems since it requires minimal planning and the data can be reformatted in any plane [2]. For this purpose, different techniques have been proposed to study anatomy [3–5]; function [6] and flow of the whole heart and great vessels [7]. Among them, a sequence called 3D dual cardiac phase [8] allows for assessment of the heart anatomy and evaluation of functional parameters from a single free breathing scan [8,9]. This approach uses a 3D b-SSFP sequence, Cartesian sampling and SENSE reconstruction, typically with acceleration factors of two; however, the acquisition time still remains long.

In order to speed up the acquisition of static whole heart images, a radial phase encoding (RPE) trajectory has been proposed [10]. RPE is a non-Cartesian acquisition trajectory that combines Cartesian readouts and radial phase encoding steps. The images from RPE trajectories are usually reconstructed using iterative SENSE. This technique uses additional information from coil sensitivity maps to remove undersampling artifacts [11,12].

For dynamic MRI, the redundant information between images has been used to speed up the acquisition of the MR data [13–15]. The most common approaches are *k*–*t* methods [16] and compressed

☆ Informed consent was obtained from all individual participant included in the study

* Corresponding author at: Biomedical Imaging Center, Pontificia Universidad Católica de Chile, Santiago, Chile. Tel.: +56 2 2354 8468.
E-mail address: suribe@med.puc.cl (S. Uribe).

http://dx.doi.org/10.1016/j.mri.2016.03.017
0730-725X/

sensing methods [17,18]. Another method that combines *k*-space data between images acquired at different time points is key-hole [14,15]. This approach continuously collects low frequencies from *k*-space and acquires high *k*-space frequencies only once. These high frequencies are then used to fill the high frequencies that were not acquired during other time points. An alternative method, based on a golden ratio RPE scan, has been recently proposed to exploit redundant information and to speed up the acquisition of dynamic contrast enhanced MRI [12]. This method takes advantage of the RPE trajectory that oversamples the center of *k*-space, and continuously updates low frequencies, similar to key-hole approaches. However, in contrast to key-hole, high frequencies are also acquired at each time frame. In this approach, it is important to know the amount of *k*-space information that can be shared between different time points [19].

In this work, we propose a new technique that combines *k*-space data from two different RPE trajectories to accelerate the acquisition of the 3D dual cardiac phase sequence. This approach is based on the use of complementary and undersampled RPE trajectories for each cardiac phase. Our main hypothesis is that the outer *k*-space information can be combined between both cardiac phases, allowing the use of a simple multiple-coil gridding reconstruction to obtain artifact-free images with similar quality as those provided by iterative SENSE, but with a reduced reconstruction time.

2. Methods

2.1. Acquisition scheme

A 3D dual cardiac phase scan with a RPE acquisition trajectory was implemented. The RPE trajectories for end-systole and end-diastole were angularly shifted with respect to each other (Fig. 1). The angular shift was equal to half of the angular step used in each RPE trajectory.

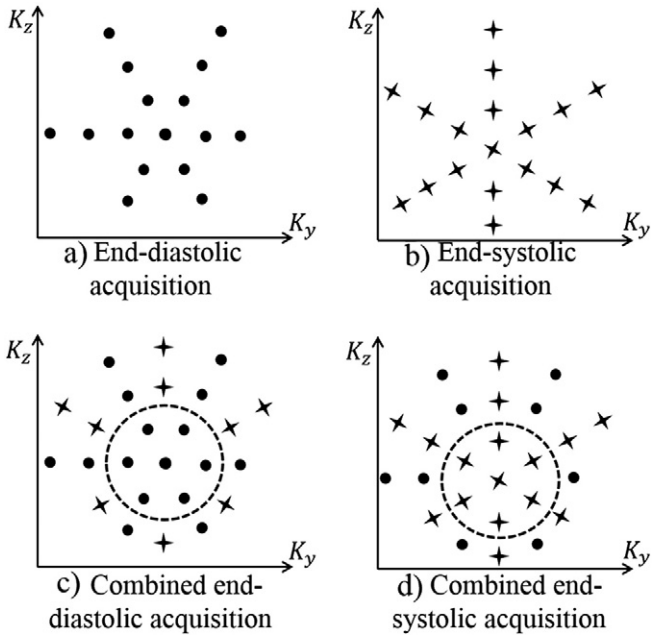

Fig. 1. Panels a and b show the acquisition scheme of the RPE trajectory in one representative K_y–K_z plane of each cardiac phase, end-systole (black points) and end-diastole (black stars). Panels c and d show the combined acquisition schemes for end-diastole and end-systole where the circle corresponds to the limit (K_r) of those points which are and are not shared between different cardiac phases.

To speed up the acquisition, the RPE trajectories were undersampled in the angular (R_α) and radial (R_r) directions in a similar fashion as proposed in Ref. [10]. The undersampling along the radial direction was performed using an interleaved undersampling scheme, such that the sampled locations were shifted in the radial direction by one position from one angular direction to the next one. The total acceleration was the product of both factors: $R = R_\alpha * R_r$. We defined the RPE trajectory as fully sampled when the number of *k*-space encodings was identical to a Cartesian trajectory for the same spatial resolution.

The proposed acquisition scheme was implemented on a 1.5 T Achieva Clinical MR scanner (Philips Healthcare, Best, NL).

2.2. Reconstruction methods

Since RPE trajectories for end-systole and end-diastole were angularly shifted with respect to each other, it was possible to combine *k*-space data between both cardiac phases. If 60% of the data were shared, it meant that 60% of the outermost *k*-space points from one cardiac phase were added to the other cardiac phase and vice versa as shown in Fig. 1.

After *k*-space data were combined, a Fourier transform was applied along the readout direction. Then, for each readout point, the phase encoding steps remained distributed on a radial trajectory. Data were reconstructed using gridding with uniform coil combination as proposed by Roemer et al. [20]. For comparison purposes, we reconstructed the original data, i.e., without combining *k*-space information between cardiac phases, using non-Cartesian iterative SENSE. The stopping criterion of the iterative method was that the residual needed to be lower than 10^{-3} or reach a maximum of eight iterations.

For gridding and non-Cartesian iterative SENSE reconstruction, we used the following density compensation function:

$$\mathrm{DCF}\left(K_y, K_z\right) = \begin{Bmatrix} \left(\sqrt{K_y^2 + K_z^2}\right)^{-1} & < K_r \\ 2^*\left(\sqrt{K_y^2 + K_z^2}\right)^{-1} & \geq K_r \end{Bmatrix} \quad (2)$$

where K_y and K_z correspond to both phase encoding directions, K_r is the ratio of areas in *k*-space that defines the area between those points which were and were not shared between different cardiac phases.

2.3. Experiments

2.3.1. 1D simulation

Two simulations were performed to analyze the effect of combining outer *k*-space information from two data sets. The first simulation evaluated the level of blurring introduced by the proposed method (Fig. 2a to f). The second simulation evaluated the definition of the edges when combining profiles from different simulated cardiac phases (Fig. 2g to l).

For both simulations, different pairs of 1D *k*-space profiles with different waveforms, centers and widths were generated. Each profile simulated an image acquired at end-systole or at end-diastole. The following function was used to simulate the profiles:

$$y(k) = \exp^{-\left(\frac{k-d}{w}\right)^c} \quad (3)$$

where *d* represents the center of the profile, *w* represents the width and *c* is a parameter that controls the shape of the profile. As *c* increases, the Fourier transform of the profile starts to resemble a rect function.

To simulate the proposed acquisition and reconstruction approach, the *k*-space data of each profile were combined as follows.