Magnetic Resonance Imaging 31 (2013) 1390–1398

Contents lists available at ScienceDirect

Magnetic Resonance Imaging

journal homepage: www.mrijournal.com

CUDA accelerated method for motion correction in MR PROPELLER imaging☆

Chaolu Feng [a,b,*], Jingzhu Yang [b,c], Dazhe Zhao [b,c], Jiren Liu [a,c]

[a] School of Information Science and Engineering, Northeastern University, Shenyang, CO 110819, China
[b] Key Laboratory of Medical Image Computing of Ministry of Education, Northeastern University, Shenyang, CO 110819, China
[c] Neusoft Group Co., Ltd., Shenyang, CO 110179, China

ARTICLE INFO

Article history:
Received 31 December 2012
Revised 12 March 2013
Accepted 27 April 2013

Keywords:
Artifact
CUDA
Motion correction
PROPELLER

ABSTRACT

In PROPELLER, raw data are collected in N strips, each locating at the center of k-space and consisting of M_x sampling points in frequency encoding direction and L lines in phase encoding direction. Phase correction, rotation correction, and translation correction are used to remove artifacts caused by physiological motion and physical movement, but their time complexities reach $O(M_x \times M_x \times L \times N)$, $O(N \times R_A \times M_x \times L \times (M_x \times L + R_N \times R_N))$, and $O(N \times (R_N \times R_N + M_x \times L))$ where $R_N \times R_N$ is the coordinate space each strip gridded onto and R_A denotes the rotation range. A CUDA accelerated method is proposed in this paper to improve their performances. Although our method is implemented on a general PC with Geforce 8800GT and Intel Core(TM)2 E6550 2.33 GHz, it can directly run on more modern GPUs and achieve a greater speedup ratio without being changed. Experiments demonstrate that (1) our CUDA accelerated phase correction achieves exactly the same result with the non-accelerated implementation, (2) the results of our CUDA accelerated rotation correction and translation correction have only slight differences with those of their non-accelerated implementation, (3) images reconstructed from the motion correction results of CUDA accelerated methods proposed in this paper satisfy the clinical requirements, and (4) the speed up ratio is close to 6.5.

1. Introduction

Magnetic resonance imaging (MRI) is popularly used in medical diagnosis over other modalities for its various advantages, such as no ionizing radiation, high spatial resolution, the ability to produce anatomical images at arbitrary cross-sections, and high soft-tissue contrast [1]. In order to obtain such high quality and diagnostically interpretable images, a relatively long-time is required for MRI to collect data and reconstruct images, which makes magnetic resonance images vulnerable to artifacts caused by physiological motion (e.g. cardiac and respiratory movements) or physical movement (patient's voluntary or involuntary movements, e.g. eye movements, swallowing, and head or limb movements) [2], especially for children, elderly people or patients suffering from certain medical conditions (e.g. Parkinson's disease and stroke), who can hardly keep still during the scan [3].

In order to reduce the misidentification of location information, promote the vital anatomical details, improve the diagnostic confidence of the images, and avoid the requirement of scan repetition, various investigations have focused on reducing the artifacts from magnetic resonance images which can be classified into prospective motion correction and retrospective motion correction. In prospective correction methods, motion and movement are compensated during data acquisition by changing the scanner coordinate system according to the patient in real-time [4], introducing additional navigator echoes [5,6], attaching active markers to the patient [3,7], or utilizing external tracking systems [8]. This implies a real-time detection of the patient's pose during the scan, scanning with a variety of k-space trajectory shapes [9–12], introducing micro radio frequency (RF) coils [13], or using optical monitors or other hardware such as a mounted camera system and patient bite bar to provide rapid and slice-wise motion correction [7,14]. On the other hand, retrospective motion correction, which is also named post-processing correction, compensates for the influence of motion by performing image-based registration of multiple acquired volumes for dynamic scans [15], corrupting individual slices for segmented scans [16], or acquiring additional data on top of the regular imaging data to measure and correct motion and movement [17]. The periodically rotated overlapping parallel lines

☆ Funding: This work was supported by the Fundamental Research Funds for the Central Universities under grant N100604015, China. This work was also supported by Chinese National Natural Science Foundation under grant No. 61001047. This work was also partly supported by project under grant N100418002.

* Corresponding author.
E-mail addresses: fengchl@neusoft.com, Chaolu.Feng@gmail.com (C. Feng).

0730-725X/$ – see front matter © 2013 Elsevier Inc. All rights reserved.
http://dx.doi.org/10.1016/j.mri.2013.04.013

with enhanced reconstruction (PROPELLER) known as a type of self-navigating data collection and motion correction method was introduced into clinical practice by J.G. Pipe in [18]. The correlative characteristics of center oversampling of k-space are used to correct the artifacts resulting from the translational and/or the rotational motion that occurred during the scan. In PROPELLER, raw data, which are used to reconstruct images after motion correction, are collected in k-space in N strips, each consisting of L parallel linear trajectories. These linear trajectories correspond to the L lowest frequency phase encoded lines in conventional methods, in which data are collected in a Cartesian trajectory. Each strip of phase encoded lines can be collected with a single radio frequency excitation in one echo train. The overlap between strips merges into a central circle in k-space as the overlap decreases at higher spatial frequencies and peripheral k-space values are measured by a single strip. Because Contrast Ratio (CR) and Signal Noise Ratio (SNR) are mainly decided by the low frequency signals which are located at the center of k-space, images reconstructed by PROPELLER have a higher CR and SNR. In order to obtain diagnostically interpretable images, a motion correction procedure is used in PROPELLER to eliminate artifacts caused by patients' movements or physiological motion. Because k-space is non-uniformly sampled, after motion correction, raw data have to be re-sampled on Cartesian coordinate space and the final image is reconstructed by Inverse Fast Fourier Transform (IFFT). However, the time complexity of motion correction slows down the reconstruction, which limits its clinical application and development.

In our previous work, a CUDA accelerated reverse gridding method was proposed to re-sample the raw data uniformly and we obtained nearly octuple speedup [19]. If not specially specified, we use this reverse gridding as the default gridding method in this paper. As a continuous work, we propose a CUDA accelerated method in this paper to improve the performance of motion correction methods existing in PROPELLER, which include phase correction, bulk rotation correction, and bulk translation correction. This paper is organized as follows. The principles of motion correction methods and image reconstruction are briefly introduced in the following section. In section 3, we give a pseudo-code for each of these correction methods and analyze their time complexities individually. We then introduce the CUDA model and propose a CUDA accelerated method for each of these correction methods in section 4. In section 5, one simulation water phantom, one no-motion brain data, and one motion brain data are used to validate the correctness and evaluate the performance of our method. We conclude our work in section 6.

2. Motion correction and image reconstruction

In PROPELLER, raw data are collected in k-space with M_y parallel linear trajectories, which are also named phase encoding lines. Each linear trajectory consists of M_x sampling points in frequency-encoding direction. The k-space is sampled by N concentric rotated strips, each consisting of the L lowest frequency phase encoded lines which are located at the center of M_y. Therefore, the overlap between strips merges into a central circle in k-space and the whole k-space is filled with trajectory S. The acquisition assumes a circular coverage and a circular field of view (FOV) in k-space. In order to eliminate artifacts caused by patients' movements or physiological motion, rotation correction is used to compensate for the trajectory; whereas phase correction and translation correction are used to correct the raw data. Due to this non-uniform sampling, after motion correction, raw data have to be uniformly re-sampled onto the Cartesian coordinate space before the final image is reconstructed by IFFT.

2.1. Phase correction

Due to the imperfect gradient balancing along the readout direction and the eddy currents, the point of rotation center is not exactly the center of k-space. On the basis of Fourier Transformation, this displacement of rotation center results in a low-frequency spatially varying phase in image space for each strip. One can remove the k-space translation and eliminate the motion-related phase by first windowing the raw data $M(n)$ of strip n, where $1 \le n \le N$, with a triangular function, then transforming the result and the original raw data $M(n)$ of strip n into complex image I and I_Λ using Fast Fourier Transform (FFT), removing the phase of I_Λ from I, and finally transforming the result back to achieve the corrected raw data $M'(n)$ using IFFT. The triangular function can be written as

$$\Lambda(x) = \begin{cases} 2x/T, & \text{for } 0 \le x < T/2 \\ 2(T-x)/T, & \text{for } T/2 \le x < T \end{cases} \tag{1}$$

which can be easily extended into two-dimension by taking $T = M_x$ in frequency-encoding direction and setting $T = L$ in phase-encoding direction, respectively. Suppose Ph(•) denotes the phase of complex •, removing the phase of I_Λ from I can be expressed by the following formula as

$$I'(p,q) = |I(p,q)|e^{-i\phi}. \tag{2}$$

where (p,q) is the pixel index and $\varphi = Ph(I(p,q)\text{-}I_\Lambda(p,q))$.

2.2. Rotation correction

In Fourier transformation theory, rotation of an object in image space results in identical rotations of its Fourier transform in k-space. Thus, the rotation of an object can be assessed by the magnitude of k-space data. As mentioned above, there is a circle (the diameter is L/FOV) in the center of k-space, which is spanned into a set of Cartesian coordinates defined as R (the resolution is $R_N \times R_N$). A reference data set M_A in k-space is formed by gridding the raw data $M(n)$ of the n-th strip inside this circle onto R, calculating complex modulus of the re-sampled result, and averaging the modulus of all strips together. For each strip, the trajectory $S(n)$ is then rotated by ra radian (ranging from $-R_A$ to R_A). Raw data of this strip are re-sampled from this new trajectory onto R. At last, a real number matrix M_n is formed by setting its elements to be the module of the gridding result. The correlation between M_n and M_A is measured by the following equation

$$C_n = \frac{\sum_{i=1}^{R_N}\sum_{j=1}^{R_N}\left(M_A(i,j)-\overline{M_A}\right)\left(M_n(i,j)-\overline{M_n}\right)}{\sqrt{\sum_{i=1}^{R_N}\sum_{j=1}^{R_N}\left(M_A(i,j)-\overline{M_A}\right)^2\sum_{i=1}^{R_N}\sum_{j=1}^{R_N}\left(M_n(i,j)-\overline{M_n}\right)^2}} \tag{3}$$

where $\overline{M_A}$ and $\overline{M_n}$ denote the average of elements in M_A and M_n, respectively. The correlation C_n should be largest when M_n is rotated to the magnitude-weighted average rotation of M_A. Once the largest correction is determined, the rotation angle R_a used to reduce motion from this strip is estimated and the trajectory of this strip is rotated by the correction angle R_a to match this estimation.

2.3. Translation correction

It is different from rotation of an object in image space producing identical rotations of its Fourier transform in k-space that shifts of an object in image space result in linear phase shifts in k-space. Translation correction is used to eliminate these shifts