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Cardiac Magnetic Resonance Imaging using Radial k-space Sampling and Self-Calibrated Partial Parallel Reconstruction

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Abstract

Radial sampling has been demonstrated potentially useful in cardiac magnetic resonance imaging because it is less susceptible to motion than Cartesian sampling. Nevertheless, its capability of imaging acceleration remains limited by undersampling-induced streaking artifacts. In this study, a self-calibrated reconstruction method was developed to suppress streaking artifacts for highly accelerated parallel radial acquisitions in cardiac magnetic resonance imaging. Two- (2D) and three-dimensional (3D) radial k-space data were collected from a phantom and healthy volunteers. Images reconstructed using the proposed method and the conventional regridding method were compared based on statistical analysis on a 4-point scale imaging scoring. It was demonstrated that the proposed method can effectively remove undersampling streaking artifacts and significantly improve image quality ($p < 0.05$). Using the proposed method, image score (1–4, 1=poor, 2=good, 3=very good, 4=excellent) was improved from 2.14 to 3.34 with the use of an undersampling factor of 4 and from 1.09 to 2.5 with the use of an undersampling factor of 8. Our study demonstrates that the proposed reconstruction method is effective for highly-accelerated cardiac imaging applications using parallel radial acquisitions without calibration data.

Keywords

Parallel imaging; Radial sampling; VIPR

1. Introduction

Motion is a long-standing issue in cardiac magnetic resonance imaging (CMRI). A few important motion sources include respiration, myocardial contraction and blood flow. Because of motion in CMRI scans, ghosting or blurring artifacts may be generated in image reconstruction (1,2). High-definition CMRI usually requires high spatial or temporal resolution and thereby a long scan time. This makes the cardiac imaging data acquisition extremely challenging. In many CMRI applications, patients are required to hold breath in order to reduce respiratory motion (3). However, this approach gives patients discomfort and the data acquisition time may still be long for those patients with low breath-hold capability. Free-breathing acquisition has been used with respiratory gating techniques (4–6). This

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technique can reduce the motion artifacts, but it significantly prolongs the scan and increases the susceptibility to motion other than respiration. Therefore, CMRI critically requires advanced fast imaging techniques for reduced scan time.

Extensive investigations have been made on the application of non-Cartesian k-space sampling trajectory in MRI (7–12). It has been demonstrated that radial k-space sampling is potentially useful in CMRI, because radial trajectory has two desirable features. First, its repeated sampling of center k-space gives signal averaging over the image space and is intrinsically not sensitive to motion effects. Second, compared to patterned aliasing artifacts resulted in undersampled Cartesian acquisition, noise-like streaks from undersampled radial acquisitions disperse over the entire image space and are more tolerable in final reconstructed images. Therefore, high undersampling can be employed to reduce the scan time and motion-related challenges in CMRI while maintaining adequate image quality. In general, SNR will be reduced if undersampling is used in data acquisition. In radial imaging, this SNR reduction does not affect the final image quality as severely as that in Cartesian imaging because the oversampling in center k-space increases the base level of SNR. However, the technical challenge in radial imaging is how to efficiently suppress streak artifacts in reconstruction. This has been the key issue in accelerated radial imaging. Conventional radial reconstruction methods perform filtered back projection or regridding to Cartesian k-space (13–15). Both have limited capability in reconstructing high-quality images from highly undersampled data. More recently, several novel techniques have been proposed to address this issue including focal underdetermined system solver (FOCUSS) and the use of the total variation constraint (16, 17). These studies have demonstrated the feasibility to improve the radial reconstruction by taking advantage of intrinsic properties of streaking artifacts. Many other investigations have attempted to exploit parallel imaging in radial reconstruction (18–23). However, combination of partial parallel imaging (PPI) and radial acquisitions is not a trivial task. Many algorithms suited for Cartesian PPI require resolving a large system of linear equations in radial imaging. This makes the algorithm very complicated and the computation cost substantially huge (18). Another disadvantage of some existing methods is the need to acquire extra calibration signal, which not only causes the inconsistency problem between the external calibration data with the accelerated data due to cardiac motion, but also can considerably reduce the data acquisition speed. These unfavorable features make this technique much less attractive to CMRI. Several self-calibrated parallel imaging methods (21–23) have been proposed to address this limitation.

Radial acquisitions have been previously used in CMRI (11, 12, 24, 25). In most previous studies, attention has been paid to SNR and artifact level. It should be pointed out that resolution is also an important concern in some cardiac applications, e.g., coronary imaging (25), because the blood vessel size is very small. Radial sampling has a lower sampling density in outer k-space than in center k-space. This intrinsically reduces the reconstruction accuracy of high spatial-frequency signals. Furthermore, filtering along the radial projection used in conventional reconstruction methods (26, 27) introduces a trade-off between SNR and resolution. If this trade-off were not balanced properly, radial reconstruction could end up with either high artifact level or poor resolution, which can fundamentally degrade the clinical value of images. To resolve this issue, the development of an efficient reconstruction method is critical.

The purpose of this study is to investigate a new method for reconstruction of radial CMRI data. In this study, we take advantage of the intrinsic properties of streaking artifacts that deteriorate radial imaging and efficiently combine radial imaging with parallel imaging together. A novel reconstruction technique for parallel radial CMRI was developed with the following three features: efficient computation, no requirement of extra calibration signals, and a balance between SNR and spatial resolution. To validate its effectiveness in CMRI

applications, we applied 2D, 3D hybrid (2D in-plane radial and 1D Cartesian) and 3D radial imaging (VIPR) in cardiac function imaging, and whole heart imaging. It is demonstrated that this technique can efficiently balance the tradeoff between SNR and resolution in reconstruction. The proposed method produces high quality reconstructions from a small number of projections in CMRI applications.

2. Theory

Mathematically, the image reconstructed from a set of radial data can be modeled as the convolution of the real image and the polar point spread function (PSF) corresponding to the radial sampling trajectory. Due to an imperfect PSF in under-sampled radial acquisitions for accelerated imaging, streaking artifacts may be generated (28). In this section, the polar PSF will be first discussed and a self-calibrated reconstruction will be introduced based on the property of polar PSF. For simplicity, only the equations for 2D radial acquisition are presented. The equations for 3D acquisition (VIPR) (29) can be derived simply by adding one dimension in the 2D equations.

2.1 Properties of Polar Point Spread Function

In 2D radial acquisition, the k-space data are sampled equidistantly in radial k_r and angular k_θ direction. Without loss of generality, a regular radial trajectory with N_θ projections that covers 180° can be defined as that with the first projection positioned at the angle of zero. Its sampling function can be written as a sum over the product of two delta functions:

$$s(k_r, k_\theta) = \left(\frac{\pi}{N_\theta}\right) \Delta k_r \sum_{m=0}^{2N_\theta-1} \sum_{n=0}^{N_r-1} \left[\delta\left(k_\theta - m\frac{\pi}{N_\theta}\right) \delta(k_r - n\Delta k_r) \right],$$

where Δk_r is the radial increment in k-space and there are $2N_r$ data points along every radial projection. The PSF for this radial trajectory in image-space is the Fourier transform of the above equation:

$$PSF(r, \theta) = \left(\frac{\pi \Delta k_r^2}{N_\theta}\right) \sum_{m=0}^{2N_\theta-1} \sum_{n=0}^{N_r-1} \left[n \exp\left(-j2\pi n \Delta k_r r \cos\left(\theta - \frac{m\pi}{N_\theta}\right)\right) \right],$$

where (r, θ) represents the position of a pixel in the polar coordinate system in image-space.

Equation 2 can be calculated by numerical method. A few typical artifact patterns generated by this PSF have been investigated in a previous study (23). In accelerated imaging, we always want to reduce the number of projections in data sampling, i.e., undersample the data in angular direction. In radial direction, the Nyquist criterion is satisfied, which implies:

$$\Delta k_r r \leq 1.$$

In this work, only the artifacts related to the reduced number of projections are discussed. In Fig. 1, three calculated PSFs are shown with 16, 32 and 64 projections. It can be seen that there exists a circular region in the center of each PSF. Inside this circular region, the PSF is flat except a huge peak at the center of the circle.. It should be noted that those streaks outside the circular region will make no contribution to the convolution between an image and the polar PSF if the image object dimension is smaller than the radius of the circle in the PSF. This forms the critical requirement for radial sampling without considerable streaking

artifacts. As shown from (a) to (c) in Fig. 1, the diameter of the center flat region in a polar PSF is proportional to the number of projections (23). As a result, considerable artifacts will always be seen in the image reconstructed from the reduced number of projections using the gridding or back-projection method.

Equation 2 can be used to determine the location of streaks in a PSF. If the term $\cos(\theta - m\pi/N_0) \neq 0$, the magnitude of PSF is small due to phase incoherence of the terms in summation for any r that satisfies Eq. 3. When $\cos(\theta - m\pi/N_0) = 0$, i.e., $\theta = m\pi/N_0 \pm \pi/2$, the magnitude of PSF is large due to phase coherence of those terms summed in Eq. 2. This implies the streaks are located at the angles of $m\pi/N_0 \pm \pi/2$. It can then be understood that the number of streaks that may cause strong artifacts is always equal to the number of projections. In other words, more projections make the artifact energy be distributed at more locations, which follows lower artifact level at each location. Figure 2(a) shows a typical pattern of a regular PSF with 16 projections and (b) shows the plot of its magnitude against the angle θ at a particular value of r . It can be clearly seen that the corresponding 16 streaks are located at the angles equal to $m\pi/N_0 \pm \pi/2$ with $m=0, 1, 2, \dots, 15$.

Now let S be a positive integer and N be a multiple of S . Consider a regular sampling function with N projections. In k -space, these N projections can be equally divided into S groups, each of which contains N/S projections spaced equal-distantly in angular direction. Apparently, these S groups of projections are just rotated versions of the regular radial sampling function with N/S projections. The PSFs of these rotated sampling functions can be obtained by rotating the regular PSF of N/S projections by an angle of $i\pi/N$:

$$PSF_{N/S}^i(r, \theta) = PSF_{N/S}\left(r, \theta - \frac{i\pi}{N}\right) = \left(\frac{S\pi\Delta k_r^2}{N_0}\right) \sum_{m=0}^{\frac{2N}{S}-1} \sum_{n=0}^{N_r-1} \left[n \exp\left(-j2\pi n \Delta k_r r \cos\left(\theta - \frac{(mS+i)\pi}{N}\right)\right) \right],$$

where $i = 0, 1, 2, \dots, S-1$ indicates the PSF of the i th group of projections. From this equation, it can be known that the streaks in these PSFs are not overlapped and they are simply the rotated versions of those in the regular PSF with N/S projections. By comparing Eqs. 2 and 4, we have:

$$PSF_N = \frac{1}{S} \sum_{i=0}^{S-1} PSF_{N/S}^i,$$

Because the magnitudes of streaks are much larger than the magnitudes of PSF at other positions, it is easy to find out that the below equation is true:

$$PSF_N\left(r, \frac{(mS+i)\pi}{N} \pm \frac{\pi}{2}\right) \approx \frac{1}{S} PSF_{N/S}^i\left(r, \frac{(mS+i)\pi}{N} \pm \frac{\pi}{2}\right),$$

for $m=0, 1, 2, \dots, N$, and $i=0, 1, 2, \dots, S$. This relationship can be illustrated using an example with $N=16$ and $S=4$ in Fig. 2. From Figure 2(c)~(j), the PSFs are shown and the plots of PSF along the angular direction are also given. It can be seen that all the streaks in the PSF with N projections are divided into S groups, each of which appear in one PSF with N/S projections and have magnitudes S times higher than those in the PSF with N projections.

2.2 Determination of Streak Locations and Generation of Calibration Images

In this section, the property (Eq. 6) of polar PSF is used to generate a calibration image to train the PPI operator. The artifact level in the calibration image is S times larger than that in the original image. Let the number of projections in a set of radial imaging data be N . These N projections can be divided into S groups, each of which contains equally spaced N/S projections. By a regridding or back-projection method, one can generate an original image from all the N projections and S derivative images from S groups of N/S projections. Because the streaking artifacts in a radial image are caused by the convolution between the real image and the streaks in the polar PSF, the implication of Eq. 6 on the above operation can be stated as follow: The streaks in the original image are also divided into S groups, each of which appears in only one of the derivative images with the magnitudes S times higher than that in the original image. Figure 2 presents one example.

It has been demonstrated in a previous study that the magnitude variation of an image will dramatically increase due to the existence of streaking artifacts (14). As a result, the magnitude variation at the locations of streaks in one of the S derivative images should be higher than that at the same locations in the other $S-1$ images because the streak at a particular location appears in only one of those derivative images. Simply by comparing the magnitude variation pixel by pixel in those derivative images, we can generate an image with all streaking artifacts and its artifact level will be S times higher than that in the original image. This image can be used for the calibration of a reconstruction operator. The details of this image generation are presented below.

Let $I_c(x, y)$ be the images reconstructed from N radial projections acquired from M receive channels, where (x, y) represents the pixel index in image-space, and $c=1, 2, \dots, M$ is the channel number. These images can be reconstructed using either a regular regridding or back-projection method. Correspondingly, the $S \times M$ derivative images for this set of data can be written as $I_{c,i}(x, y)$, where $i=0, 1, 2, \dots, S-1$. The generation of calibration image can follow the below calculation:

Calculate the sum of square images:

$$IS_i(x, y) = \sum_{c=1}^M |I_{c,i}(x, y)|^2, i=0, 1, 2, \dots, S-1.$$

Calculate variation:

$$V_i(x, y) = |IS_i(x, y) - IS_i(x-1, y)| + |IS_i(x, y) - IS_i(x, y-1)|,$$

Select the image values for every pixel (x, y) from $I_{c,i}(x, y)$:

$$I_{c,a}(x, y) = I_{c,l}(x, y), l \in \{0, 1, \dots, S-1\} \text{ and } V_l(x, y) = \max \{V_i(x, y), i=0, 1, \dots, S-1\}.$$

Because the images $I_{c,a}(x, y)$ are those selected pixels from the S derivative images $I_{c,i}(x, y)$ with the highest variation, they include all the streaking artifacts in the images $I_c(x, y)$, but with the artifact level S times higher. Notice, calibration image is calculated for image from each channel.

2.3 Calculation and Application of PPI Reconstruction Operator

Once the set of calibration images is available, the PPI reconstruction operator can be calculated in image space. The operator is defined as spatially adaptive weights in image space. The weighted summation of images from M channels can suppress streaking artifacts by a factor of S . To calculate the operator, the calibration images $I_{c,a}(x, y)$, $c=1, 2, \dots, M$, are linearly combined to approximate the original image set $I_c(x, y)$, $c=1, 2, \dots, M$, which has S times lower artifact level. An error function for i th channel can be defined as:

$$e_i(x, y) = \left| I_i(x, y) - \sum_{c=1}^M r_{i,c}(x, y) I_{c,a}(x, y) \right|^2,$$

where $r_{i,c}$'s, $i=1, 2, \dots, M$ and $c=1, 2, \dots, M$, represent a set of coefficients for linear reconstruction operator. By letting the partial derivative

$$\frac{\partial e_c(x, y)}{\partial r_{c,l}} = 0, l=1, 2, \dots, M,$$

a set of linear equations will be generated and the coefficients $r_{i,c}$'s can be resolved by the calculation of pseudo-inverse. After the calibration, the reconstruction operator can be directly applied to the images $I_c(x, y)$, $c=1, 2, \dots, M$, and a reconstructed image can be generated by:

$$\bar{I}_i(x, y) = \sum_{c=1}^M r_{i,c}(x, y) I_c(x, y).$$

Because streaks in the images $I_{c,a}(x, y)$ have a level S times higher than that in the images $I_c(x, y)$, the reconstruction operator calibrated by Eqs. 10 and 11 is capable of suppressing the streaking artifacts by a factor of S . For this reason, “ S ” is referred to as “suppressing factor” in this work.

It should be understood that the PPI reconstruction operator defined above is a general image space reconstruction operator. The purpose is to take advantage of spatial encoding capability of coil sensitivities to reduce the noise amplification in reconstruction. Eq. 10 may still generate a set of valid coefficients if the channel number $M=1$, i.e., without parallel imaging. However, this generated reconstruction operator will amplify noise more in comparison to that when the channel number M is much larger than 1, i.e., with parallel imaging.

3. Methods

The method described in the previous section was applied to a 2D phantom imaging, and three in-vivo CMRI applications: 2D cardiac function, 3D hybrid (2D radial in-plane and Cartesian sampling through-plane) whole-heart and 3D VIPR whole-heart imaging. Written consent was obtained from all volunteers before each experiment in compliance with our institutional review board guidelines.

3.1 Data acquisition

A phantom image was acquired on a Philips 3T scanner with an 8-channel coil array (Invivo Diagnostic Imaging, Gainesville, FL). A 2D gradient echo sequence (FOV=250 mm, flip angle = 60 degree, TR/TE=500/16.11 ms, slice thickness = 5 mm) was used with a radial sampling trajectory including 512 projections and 512 points per projection. At this high sampling rate, no considerable streaking artifacts were seen in reconstruction using the regular regridding or filtered back-projection method.

Breath hold 2D cine images were acquired on a Siemens TIM Trio magnet with 2D TrueFISP radial sequence with 256 projections and 512 points/projection. The field of view was 30 cm, flip angle was 65 degrees, slice thickness 5 mm, TR = 36.64 ms, TE = 2.29 ms, The data were acquired with a 4-channel cardiac coil, and have 17 cardiac phases.

Whole-heart imaging data were acquired using 3D hybrid and VIPR sampling trajectories. 3D hybrid imaging was implemented on a 1.5T Siemens Espree system. Ten volunteers were scanned during free breathing using an ECG-triggered, self-gating 3D SSFP sequence with TR = 4.0ms, TE = 1.8ms, flip angle = 70 degree, resolution = $1.2 \times 1.2 \times 2.0 \text{ mm}^3$, 48 partitions, 256 projections with 512 readout points per projection. The scan time varies from 10 to 12 minutes subject to volunteer's respiratory acceptance rate. 3D VIPR imaging was implemented on a 3.0T Siemens Trio system. Eight volunteers were scanned using a segmented SSFP sequence with navigator gating (NAV). A whole-heart slab was scanned using the following parameters: $350 \times 350 \times 350 \text{ mm}^3$ FOV, 280 readout points; $1.3 \times 1.3 \times 1.3 \text{ mm}^3$ isotropic resolution; 11520 projections; 40 lines/segment; 60° flip angle; TR/TE = 3.0/1.5 ms; The scan time for one measurement during free-breathing is approximately 8 minutes. A 12-channel body coil was used in both studies.

3.2 Reconstruction

In 2D phantom imaging, the image reconstructed by regridding from 512 projections was used as the reference image. Three sets of data with 16, 32 and 64 projections were generated by artificially undersampling the 512 acquired projections. These data were used to simulate the accelerated data acquisition by 32, 16, 8 times respectively. The reconstructed images using the proposed method were compared with the reference image. In 2D cardiac function imaging, the image reconstructed by regridding from 256 projections was used as the reference image. Two sets of data with 16 and 32 projections were generated by artificially undersampling the 256 acquired projections. Both the regridding and the proposed methods were used to reconstruct these data. The reconstructed images were compared with the reference image and the reconstruction results were compared with that using 256 projections. In 3D in vivo imaging (hybrid and VIPR), the images reconstructed by regridding from all the acquired projections were used as reference images. The data sets with acceleration factors of 4 and 8 with respect to the number of the acquired projections were generated by artificially undersampling the acquired projections. Both the regridding and the proposed methods were used in reconstruction. The reconstructed images were compared with the reference images. In all the reconstruction, a standard algorithm (31, 32) was used for regridding.

3.3 Quantitative evaluation

To compare the reconstructed image quality with the proposed parallel imaging method and with directly regridding images, two objective observers reviewed and scored the images reconstructed randomly based on a 4-point scale (25). The image dataset includes 2D cine, 3D hybrid and 3D VIPR image results. The scoring scale and criteria for evaluation were as follows: 1 = poor delineation or uninterpretable of scanned objects (with markedly blurred borders or edges), 2 = good (object visible, but moderately blurred), 3 = very good (object

clearly visible but blurring is still mildly present), and 4 = excellent (object visible, well delineated with no visible artifacts). The final quantitative results of these images are determined as the mean of grading. At last, a two-tailed, paired sample t-test was used to evaluate the differences in image grades between images reconstructed with and without parallel imaging methods. P-value < 0.05 is considered as statistically significant.

4. Results

4.1 Phantom Studies

The proposed self-calibrated reconstruction method is illustrated in Fig. 3. The phantom data with 32 projections were used to simulate the undersampled data (corresponding to an undersampling factor of 8). A suppression factor of 4 was used in this reconstruction. Correspondingly, the 32 projections were first divided into 4 groups, each of which contains 8 projections. The images reconstructed from these 4 groups of projections using gridding method are shown in Figs. 3(a) ~ (d). The positions of these streaks can be detected using the method in Esq. 7 and 8. The results are shown using the binary images in Figs. 3(e) ~ (h), where 1 indicates the position of streaks. It can be seen that these streaks in the four images are located differently. By selecting the image values as in Eq. 9, the calibration image (Fig. 3(i)) was generated with streaking artifacts level 4 times higher than the original image (Figure 3(j)), which is reconstructed from 32 projections using regridding method. It should be noted that the streaks in Fig. 3(i) are located at the same positions as those in Fig. 3(j). Then, we used these two images to calibrate a reconstruction operator using Eqs. 10 and 11. This reconstruction operator was applied to the original image in Fig. 3(j). The reconstruction will generate an image with streaks at a level about 4 time lower than the original image. This reconstructed image is shown in Fig. 3(k). Figure 3(l) is the reference image, which was reconstructed from 256 projections using regridding method. In Fig. 3, what is shown is only sum of square images over all the channels. In the calibration of reconstruction operator and the final reconstruction, the images from all the channels were used.

4.2 In vivo Studies

4.2.1 2D imaging—Figure 4 shows an example of 2D cardiac function imaging. The data with 256 projections in the first time frame of 2D cine images was used. A reference image was generated from 256 projections using regridding method and is shown in Fig. 4(a). Fig 4(b) (c) (d) show image reconstructed from 32 projections with suppression factor of 2, 4 and 6, suggesting with higher suppression factor, there will be more artifacts suppressed, however more resolution will be lost. In order to determine a suitable suppression factor for CMRI, the statistic method as described in “Methods” section for quantitative evaluation is used. A series of 2D images were generated from the data with undersampling factor from 4 to 16 using the proposed method with suppression factors from 2 to 8. The mean and variance of image scores are given in Table 1. It can be seen that the use of suppression factor 4 gives highest mean and lowest SD. Based on this statistical analysis, we used the suppression factor of 4 in reconstruction for all the in-vivo imaging except the VIPR whole heart imaging. In VIPR whole heart imaging applications, we used the suppression factor of 3 instead of 4 for reduced computation time cost. The selection of suppression factor will be discussed more thoroughly later.

Figure 5 shows how the proposed method performs with regard to the removal of streaking artifacts in 2D cardiac function imaging. A reference image was generated from 256 projections using regridding method and is shown in Fig. 5(a). Figure 5(b) shows the image reconstructed from 32 projections using regridding method. It can be seen that there exist strong streaking artifacts when an undersampling factor of 8 is used. Using the proposed

self-calibrated reconstruction method, the streaking artifacts were comparably well suppressed with a suppression factor of 4, as shown in Fig. 5(c). One more comparison with undersampling factor of 16 is shown in Fig. 5(d) and (e).

4.2.2 3D hybrid imaging—Figure 6 shows the reconstruction results for two volunteers as examples. Figures 6(a) and (b) show the images reconstructed from 256 projections using gridding method. Figures 6(c) and (d) are the images reconstructed from 64 projections (corresponding undersampling factor 4 and the acquisition time close to 3 minutes) using regridding method. Strong streaks can be seen over the entire FOV in this reconstruction. By applying the proposed self-calibrated reconstruction method, it can be seen that streaking artifacts are comparably well suppressed in the reconstructed images shown in Figs 6(e) and (f).

4.2.3 3D VIPR imaging—An example of reconstruction results in 3D whole heart imaging is shown in Fig. 7. Figure 7(a) gives the image reconstructed from total 11520 projections using regridding method. Figure 7(b) shows the image reconstructed from 2880 projections (corresponding undersampling factor 4 and the acquisition time of close to 2 minutes). In this image, streaking artifacts are shown in a noise-like pattern. After the application of the proposed technique, it can be seen that the noise-like artifacts are effectively reduced, as shown in Fig. 7(c).

4.3 Quantitative evaluation

The mean image quality scores and standard deviations from reference images, images directly reconstructed with undersampling factors 4 and 8, and images reconstructed using the proposed parallel imaging method with reduction factors 4 and 8 are showed in Table 2. The image quality with the proposed parallel imaging method increased significantly ($p < 0.05$) as compared to the directly reconstructed images with the same reduction factor and maintained a good score of 3.24 even with a reduction factor of 4. Compared to the reference images, there was no significant loss in image quality with reduction factor of 4. However, image quality degraded with reduction factor of 8.

5. Discussion

In this study, we proposed a self-calibrated reconstruction technique for radial CMRI imaging. This technique takes advantages of two intrinsic properties of radial sampling: 1) Regridding reconstruction using a subgroup of acquired radial projections will not change the location of streaking artifacts, but will enhance the intensity level of streaking artifacts, in comparison with the reconstruction using all the acquired projections. 2) Streaking artifacts will increase the level of total magnitude variation in radial images. Based on these two properties, two sets of images with different intensity level of streak artifacts can be generated and a linear reconstruction operator can be calibrated for artifact suppression. The effectiveness of this method has been demonstrated in several cardiac imaging applications.

There is a critical parameter in the proposed technique: suppression factor S . It is important to understand what role this parameter plays in the algorithm. Ideally, the suppression factor represents the ratio of streak level between the derivative and original images discussed in the section of theory. It should be noted the generation of the derivative image relies on the determination of streak locations, which uses the total variation method. This total variation method works well when the streaks related to different subgroups of projections are separated well from each other. However, if these streaks have some overlap, e.g., in the regions marked with arrows in Fig. 3(i), the comparison of total variation will not be reliable. In this situation, it is easy to cause mis-calibration from the derivative image to the

original image and introduce some loss in image information. Because the overlap of streaks often happens at the edges of imaging objects, this may introduce the loss of imaging details, i.e., the loss of image resolution. In general, the use of a higher suppression factor will reduce the reliability in determining the streak locations because a smaller number of projections will give lower SNR in derivative images. At the same time, it should be noted that a higher suppression factor can also reduce the level of streaking artifacts, which means the suppression of noise level or higher SNR. Therefore, the suppression factor physically represents the tradeoff between SNR and image resolution. Figure 4 shows an example of cardiac function imaging. A reference image reconstructed from 256 projections is given in Fig. 4(a). The other images in Figs. 4(b) ~ (d) are those reconstructed from 16 projections with suppression factors of 2, 4 and 6. From this comparison, it can be seen that the imaging information at the edge of heart tissues marked with arrows is more and more likely lost when the suppression factor goes higher and higher. On the other hand, the streaking artifacts became lower when a higher suppression factor is used. This implies that the suppression factor in this technique has to be carefully determined based on the requirements for resolution in cardiac imaging. Particularly in some imaging applications that requires high resolution, it is expected that a relatively low suppression factor is used. In this work, we empirically used $S=4$ for the in-vivo imaging applications.

Compared with some other techniques, e.g., radial GRAPPA, the proposed technique is relatively simple with respect to the computation complexity. More importantly, the introduction of a suppression factor offers an efficient way to balance between artifact level and resolution. This is especially useful in coronary imaging. By appropriately adjusting the suppression factor, it is easy to optimize the reconstruction that can keep both resolution and low streaking artifacts. For CMAI diagnosis, the imaging requirement may change case by case. It would be of significant clinical importance to have a suppression factor adjustable.

It should also be pointed out that the data acquisition hardware and pulse sequence in this work have not been optimized for the proposed parallel imaging reconstruction algorithm. Some ongoing research work is focusing on the real implementation of fast whole heart imaging within 2~3 minutes. By the use of a high-quality cardiac coil array, e.g., 32ch cardiac coil, and some further improvement on pulse sequence, it is expected that the proposed technique should give better performance than the presented results in this paper.

6. Conclusion

A self-calibrated reconstruction method for parallel radial imaging was proposed in this work. Experimental results from phantom and in vivo imaging demonstrate its effectiveness in cardiac applications when a high undersampling factor in radial trajectories is used. This will make it potentially possible to implement a cardiac whole heart imaging approximately within 2~3 minutes. Motion concerns in CMRI can be efficiently reduced. This method offers two advantages: First, it does not need any additional calibration data. Second, it can efficiently balance the trade-off between artifact level and resolution. It should be noted that this method can be easily used in many other imaging applications besides the cardiac imaging.

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Figure 1.
Polar PSF with 16, 32, and 64 projections. The diameter of center circular region depends on the total number of projections.

Figure 2.

Polar PSF for 16 projections (a), and its plots of streaks in angular direction (b), polar PSF of first 4-projection subset (c) and its plots of streaks in angular direction (d), polar PSF of second 4-projection subset (e) and its plots of streaks in angular direction (f), polar PSF of third 4-projection subset (g) and its plots of streaks in angular direction (h), and polar PSF of fourth 4-projection subset (i) and its plots of streaks in angular direction (j). Three features should be noted: 1) The number of projections and number of streaks are equal. 2) The rotation of PSF will correspondingly change the angular position of streaks. 3) The reduction in projection number will increase the magnitude of streaks.

Figure 3.

Illustration of the proposed parallel imaging method using phantom imaging data. (a)~(d) show 4 phantom images reconstructed from 4 different 8-projection subsets from 32 projections. Corresponding binary images (calculated from eqs. 7 and 8) that show location of streaks are presented in (e) ~ (h). (i) shows the generated image with high artifact level using Eq. 9. (j) shows the directly reconstruction results from 32 projections. (k) shows the final reconstruction result with 32 projections based on reconstruction operator calibrated from (i) and (j) using Eq. 11. Reference image reconstructed from fully-sampled 256 projections is given in (l).

Figure 4. Reconstructed images from cardiac function imaging data using different suppression factors. Reference image (a) was reconstructed from fully-sampled 256 projections. The other images were reconstructed from 32 projections out of 256 projections with suppression factors of 2 (b), 4 (c), and 6 (d).

Figure 5.

Reconstructed images showing the 16th out of 17 cardiac phase cine images. All 256 projections were used to reconstruct the reference images (a). (b) shows the reconstructed image from 32 projections using regridding without parallel imaging methods. (c) shows the reconstructed image from 32 projections using the proposed parallel imaging method. (d) shows the reconstructed image from 16 projections using regridding without parallel imaging methods. (e) shows the reconstructed image from 16 projections using the proposed parallel imaging method. It can be seen that the proposed method effectively suppress the streak artifacts.

Figure 6. 2 different slices of whole heart 3D HYBRID from 2 volunteers. Reference image (a) and (b) are images reconstructed from 256 projections. (c) and (d) are reconstructed image using the direct regridding from 64 projections. (e) and (f) are images reconstructed using proposed technique from 64 projections with suppression factor 4. Note there are much less streak artifacts in (e) and (f) than (c) and (d).

Figure 7.

An exemplary slice of whole heart 3D VIPR image. Reference image (a) was reconstructed from all the acquired 11520 projections. (b) is the image reconstructed using direct regridding from 2880 projections. In comparison, the image (c) reconstructed using proposed parallel imaging methods from the same 2880 projections has much less streak artifacts than (b).

Statistical analysis on the dependence of 2D cardiac image quality on suppression factors in reconstruction. The mean and variance of image quality scores corresponding to a suppression factor are estimated from the scores of those images reconstructed from the data with undersampling factors of 4, 6, 8, 10, 12, 14, and 16.

Table 1

Quantitative analysis	Suppression factor						
	2	3	4*	5	6	7	8
Mean	2.7143	2.8571	3.1429*	2.9286	2.8571	2.7857	2.7500
Standard Deviations (SD)	0.9940	0.9880	0.8018*	0.9759	0.9449	0.8092	1.0840

* marks the suggested value, which has the highest mean and lowest SD.

Table 2

Statistical analysis on image scores calibrated from all sets of in-vivo images, including 2D cine, 3D hybrid and 3D VIPR results. The mean and variance of image quality scores are calculated with reconstruction methods of proposed parallel imaging method with reduction factors of 4 and 8, directly regridding method with reduction factors of 4 and 8.

Quantitative analysis	Reconstruction Method			
	Fully acquired	Directly gridding with RF = 4	Parallel Imaging reconstructed with RF = 4*	Directly gridding with RF = 8
	Mean	3.98	2.14	3.34*
Standard Deviations (SD)	0.13	0.31	0.16*	1.09
				2.5*
				0.32*

* indicates statistically significant improvement.