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Whole-heart single breath-hold cardiac cine: A robust motion-compensated compressed sensing reconstruction method

Javier Royuela-del-Val¹, Muhammad Usman², Lucilio Cordero-Grande²,
Marcos Martin-Fernandez¹, Federico Simmross-Wattenberg¹, Claudia Prieto²,
and Carlos Alberola López¹

¹ Laboratorio de Procesado de Imagen, Universidad de Valladolid, Spain.

`{jroyval, marcma, fedsim, carlos}@lpi.tel.uva.es`

² Division of Imaging Sciences and Biomedical Engineering, King's College London,
United Kingdom.

`{muhammad.usman, claudia.prieto}@kcl.ac.uk`

Abstract. In this paper we propose a methodology to achieve single breath-hold whole-heart cine MRI with a temporal resolution of ~ 46 ms and a spatial resolution of $2 \times 2 \text{mm}^2$ out of a previously described method (JW-tTV) for single slice reconstruction. Its feasibility is tested by itself and in comparison with another state-of-the-art reconstruction method (MASTeR); both methods are adapted to Golden Radial k-space trajectories. From a formal viewpoint, we make use of a realistic numerical phantom to have a ground truth of deformation fields so that the methods performances against noise can be quantified and the sparsity regularization parameter involved in the reconstructions can be fixed according to the signal to noise ratio. Phantom results show that the adapted JW-tTV method is more robust against noise and provides more precise motion estimations and better reconstructions than MASTeR. Both methods are then applied to the reconstruction of 12-14 short axis slices covering the whole heart on eight volunteers. Finer details are better preserved with JW-tTV. Ventricle volumes and ejection fractions were computed from the volunteers data and preliminary results show agreement with conventional multiple breath-hold Cartesian acquisitions.

1 Introduction

Cine MRI has become the preferred technique for the noninvasive study of the ventricular volumes and cardiac function [4], where a stack of short axis slices covering the ventricles are acquired during several breath-holds (BH), typically one per slice, to avoid respiratory motion artifacts. This approach leads to long scan durations, misalignments between the slices acquired at slightly different BH positions and requires patient cooperation, a need that may be hindered by the patient pathology degree. The development of a technique able to cover the whole heart in a single BH would significantly palliate these shortcomings.

However, such a technique would involve the application of very high acceleration factors given the short time available for data acquisition.

Compressed sensing (CS) [7] based reconstruction methods have been developed and successfully applied to multi-slice 2D SSFP sequences. In [8], a free-breathing approach is presented in which a pseudorandom, eight-fold undersampling acquisition pattern in the Cartesian k-t space is applied. In [14] a closely related approach is applied to acquire four short axis and three long axis views of the heart in a single BH. shape model is fitted to estimate the volume of the left ventricle (LV).

In the mentioned methods, the application of a Cartesian acquisition scheme leads to limited spatio-temporal resolution and, in some cases, aliasing problems along the phase-encoding direction [14]. Moreover, the targeted temporal resolution must be strictly defined before the examination, as it would determine the acquisition pattern (roughly speaking, the temporal resolution will determine how often the center of the k-space is sampled). These limitations can be overcome by the application of a Golden radial (GR) trajectory in k-space [16]. Since no phase-encoding is used, there is no interference from objects outside the field of view and the wisely selected golden-angle step allows to continuously acquire data regardless of the final temporal resolution of the reconstructed images, resulting in a more versatile and simpler procedure.

Recently, motion estimation (ME) and motion compensation (MC) techniques have been introduced in CS methods in order to minimize the degradation of CS reconstructions due to inter frame motion [1, 9, 12]. In CS with ME/MC, motion is jointly estimated from the reconstructed images themselves which, in turn, will be affected by both remaining undersampling artifacts and noise. The degree to which these errors in the images affect the estimated motion will determine the final quality of the reconstructions. Therefore, the application of a robust ME method turns out crucial [13]. However, the lack of a noise-free ground-truth for the images and the deformation fields hinders the performance analysis of the ME techniques and the CS reconstruction methods against noise in real situations. Signal to noise ratio (SNR) is also determinant in the choice of the regularization parameters involved in CS reconstruction [6, 8]. A common procedure is to sweep the parameter space and select those parameters that provide the best reconstruction according to an expert criterion. This approach has clear limitations in real-world applications.

In this work we achieve higher acceleration factors with respect to recently published methods so whole-heart multi-slice cine single BH coverage is achieved. This is done by extending the Jacobian weighted temporal total variation (JW-tTV) [12] method, originally applied to the reconstruction of single-slice Cartesian acquired data, to use a GR acquisition scheme. This allows us to reach a x16 acceleration factor (with respect to a fully sampled Cartesian acquisition) that had remained unachieved so far. This proposal is compared with MASTeR [1], a related method which differs mainly in the ME technique used and which we have also extended to use GR. We show that quantitative cardiac function indicators, namely end diastolic (ED), end systolic (ES) and ejection fraction (EF)

values calculated from the images reconstructed with the proposed method are in accordance with those obtained from a gold standard multi-BH Cartesian acquisition.

In addition, and as a preliminary step, we analyze the effect of noise on the acquired data in two aspects. First, in the selection of the optimal value for the regularization parameter involved in the CS reconstruction in order to obtain an empirical rule to select it. Second, in the errors induced in the estimated motion fields and in the quality of the final images. To this end, we use the numerical phantom XCAT [11, 17] to generate realistic ground truth deformation fields and synthetic k-space data affected by different levels of noise. The resulting extensions of both compared methods (JW-tTV-GR and MASTeR-GR) are tested.

2 Methods

CS reconstruction of undersampled dynamic MRI with ME/MC

In CS with ME/MC, the reconstruction of an MRI image $\mathbf{m}$ from the acquired undersampled k-space data $\mathbf{y}$ can be formulated as [13]

$$\underset{\mathbf{m}}{\text{minimize}} \quad \frac{1}{2} \|\mathbf{y} - \mathbf{E}\mathbf{m}\|_{\ell_2}^2 + \lambda \|\Phi \mathcal{T}_\Theta \mathbf{m}\|_{\ell_1} \quad (1)$$

where $\mathbf{E}$ is the encoding operator that models the acquisition process. In Cartesian acquisitions, $\mathbf{E}$ comprises the multiplication of the image $\mathbf{m}$ by the coils sensitivity profiles and the application of the regular fast Fourier transform (FFT) followed by the undersampling of the data with the used sampling pattern. In cine cardiac MRI, the sparsifying transform Φ is typically chosen to be the temporal Fourier transform or the temporal total variation, in order to exploit the temporal redundancy between different cardiac phases. However, inter-frame motion will affect the redundancy of the signal along time reducing, reducing its sparsity level. The MC operator $\mathcal{T}_\Theta$ comprises a set of spatial deformations, governed by the set of parameters Θ , that deform each temporal instant in the dynamic image $\mathbf{m}$ to a common reference motion state. This way, the sparsity of the the resulting motion compensated sequence along the temporal dimension is restored and the quality of the reconstruction improved, consequently. The parameter λ , referred to as regularization parameter in the introduction section, establishes a trade off between data consistency and the sparsity of the solution.

Since the motion in $\mathbf{m}$ is not known beforehand, a regular CS reconstruction is performed as a first step, which is equivalent to solve Eq. (1), with $\mathcal{T}_\Theta$ set to the identity. Then, a ME step follows and $\mathcal{T}_\Theta$ is estimated. The ME technique consists on a groupwise registration method based on a B-spline deformation model [10], controlled by the set of control points Θ , and a simple groupwise registration metric based on the variance of the intensity along time. Those control points that minimize the value of the registration metric are found by solving the following optimization problem:

$$\underset{\Theta}{\text{minimize}} \quad \left\| \sum_{n=1}^N \left(\mathcal{T}_{\Theta,n} \mathbf{m}_n - \frac{1}{N} \sum_{k=1}^N \mathcal{T}_{\Theta,k} \mathbf{m}_k \right) \right\|^2 + R(\Theta) \quad (2)$$

where $R(\Theta)$ stands for an additional regularization term given by the second spatio-temporal derivatives of the motion fields that encourages smoothness of the estimated spatial deformations [13]. Eq. (2) is solved by means of a non linear conjugate gradient algorithm. Once $\mathcal{T}_\Theta$ has been obtained, CS with ME/MC is performed by solving Eq. (1). The NESTA optimization algorithm [3] is used to this end. Once Eq. (1) has been solved, and iterative procedure can be followed by alternating the ME/MC and the image reconstruction steps to obtain successively refined motion fields and images. In this work, the number of ME iterations has been set to three.

Finally, in JW-tTV a modification is introduced in the computation of the sparsity regularization term. The ℓ_1 term in Eq. (1) is substituted by a motion-based sparsity promoting metric given by

$$\|m\|_{\mathcal{T}_\Theta} = \left\| \sum_{n=1}^N |\mathcal{T}_{\Theta,n+1}\mathbf{m}_{n+1} - \mathcal{T}_{\Theta,n}\mathbf{m}_n| \mathcal{J}_{n+1/2} \right\|_{\ell_1} \quad (3)$$

where $\mathcal{J}_{n+1/2}$ stands for the Jacobian of the deformation, averaged between instants n and $n+1$. The Jacobian is introduced in order to weigh the contributions of the temporal differences in Eq. (3) according to their corresponding spatial extent in the original sequence (before MC), instead of in the motion compensated one. Therefore, the following reconstruction problem results:

$$\underset{\mathbf{m}}{\text{minimize}} \quad \frac{1}{2} \|\mathbf{y} - \mathbf{E}\mathbf{m}\|_{\ell_2}^2 + \lambda \|\mathbf{m}\|_{\mathcal{T}_\Theta} \quad (4)$$

Adaptation to Golden Radial acquisition

As stated in the introduction, in this work we have adapted the previously proposed JW-tTV method to work with GR trajectories in k-space in order to overcome the limitations derived from Cartesian acquisitions. In the GR scheme [16], radial profiles are continuously acquired with an angular separation given by the golden angle ($\sim 111.24^\circ$).

In order to work with non Cartesian k-space data, the encoding operator in Eq. (4) is modified by substituting the regular FFT and the undersampling pattern by the calculation of a non-uniform FFT (NUFFT). To implement the NUFFT, we resort to the well established gridding algorithm [2]. In this algorithm, the k-space data is convolved with a finite kernel, sampled onto a Cartesian grid, converted to image domain performing a regular FFT and multiplied by a deapodization function to compensate for the effect of the convolution kernel in the frequency domain.

Even if efficient implementations exist, given the need of gridding operations the computational cost of the NUFFT is significantly higher than the Cartesian FFT. However, the benefits resumed in the introduction justify its application. In order to briefly compare the results obtained with the proposed GR method and the equivalent Cartesian counterpart, in Figure 1 the reconstruction of a single SA view of the heart of one volunteer is shown. The GR pattern covers

Fig. 1. Comparison of a Cartesian (top) and GR (bottom) reconstruction of a medial SA slice of the heart for an equivalent acceleration factor of 16 at different cardiac phases (left to right).

the k-space more efficiently leading to better reconstructions, specially for the high acceleration factors involved in the single BH application.

Comparison with MASTeR

The proposed method is compared in the results section with MASTeR [1] which mainly differs from JW-tTV in the method used to estimate the motion in the image. In MASTeR, motion is estimated sequentially between each pair of consecutive frames to define a set of operators that predict each frame $\mathbf{m}_i$ out of its leading and trailing frames:

$$\begin{aligned}\mathbf{m}_i &= \mathcal{F}_{i-1}\mathbf{m}_{i-1} + \mathbf{f}_i \\ \mathbf{m}_i &= \mathcal{B}_{i+1}\mathbf{m}_{i+1} + \mathbf{b}_i\end{aligned}\tag{5}$$

where $\mathcal{F}_{i-1}$ and $\mathcal{B}_{i+1}$ denote the forward and backward MC operators. Residuals $\mathbf{f}_i$ and $\mathbf{b}_i$ are assumed to be sparse and are used as sparsity term:

$$\underset{\mathbf{m}}{\text{minimize}} \frac{1}{2} \|\mathbf{y} - \mathbf{E}\mathbf{m}\|_{\ell_2}^2 + \alpha \|\mathbf{f}\|_{\ell_1} + \beta \|\mathbf{b}\|_{\ell_1}\tag{6}$$

where $\mathbf{f}$ and $\mathbf{b}$ are the concatenation of the residuals for all the cardiac phases. In our implementation, we choose $\alpha = \beta = 0.5\lambda$, since there is no apparent reason to favor any of the two terms. A method based on complex wavelets is used for estimating inter-frame motion.

Numerical phantom for heart motion and MR images simulation

XCAT [11] was used to obtain a ground truth both for reconstructed images and for the motion deformation fields. We modified MRXCAT [17] to simulate the k-space data with a desired SNR of a medial short axis slice with the following main parameters: b-SSFP sequence, TR/TE/flip angle = 3ms/1.5ms/60°,

in-plane resolution of $1 \times 1 \text{ mm}^2$, field of view = $256 \times 256 \text{ mm}^2$. 320 radial profiles were acquired and grouped into 20 cardiac phases (16 profiles per phase). 8 virtual coils were simulated. MASTeR-GR and JW-tTV-GR were used for reconstruction and the motion fields estimated by each of those methods were stored for posterior analysis. A dense sweep on the regularization parameter λ was performed for each SNR in order to find its optimal value in order to get both the maximum signal to error ratio (SER) and the best structure similarity index (SSIM) [15], calculated using the noise-free phantom image as the reference.

In-vivo experiments: single BH whole-heart acquisitions

Eight healthy volunteers were scanned with a 32-element cardiac coil in two 1.5T Philips scanners (four each). Relevant scan parameters include: b-SSFP sequence, TR/TE/flip angle = $2.9 \text{ ms}/1.44 \text{ ms}/60^\circ$, field of view = $320 \times 320 \text{ mm}^2$, spatial resolution = $2 \times 2 \text{ mm}^2$, slice thickness = 8mm with no gap between slices. For each slice (12 to 14 were acquired, depending on volunteers heart size) between 216 and 272 radial profiles with 320 frequency encoding samples were acquired during a single cardiac cycle with ECG triggering. 13-16 cardiac phases were retrospectively reconstructed with a fixed temporal resolution of 46.4ms (16 profiles per frame). A fully sampled, Cartesian, multi BH scan was performed with similar parameters for comparison. Breath-hold duration ranged 9 to 15 seconds. Sensitivity maps were estimated from a separate scan.

All the reconstructions were run off line on a server with two Intel Xeon E5-2695 v3 CPUs @ 2.30 GHz and 64 GB of RAM using MATLAB (R2015a, The MathWorks, Natick, MA). At the current implementation, both methods are applied to each slice independently from the others, what makes parallelization of the reconstruction straight forward.

3 Results

Figure 2 shows the SER and SSIM values for the JW-tTV reconstruction of the phantom data for a range of SNR and λ values. As the SNR diminishes, the optimal value of λ increases steadily. Therefore, higher level of sparse regularization results in better reconstructed images, as predicted by CS theory [7]. Only small differences in optimal λ appear depending on the quality metric used. These results were used as a reference to select λ in the reconstruction of in vivo data, where the SNR was estimated from the fully sampled, Cartesian images taking a background region as a reference for noise level. After revision by an experienced observer, the final values of λ were increased in order to maximize the subjective quality perception.

The reconstructions obtained for a selection of the SNR values simulated with the numerical phantom are summarized in Figure 3. Sharper edges are recovered with JW-tTV than with MASTeR, specially for lower SNR values. More interestingly, MASTeR reconstructions show an erratic motion in the temporal

Fig. 2. SER (left) and SSIM index (center) obtained for each combination of λ in Eq. (4) and SNR value are plotted –dashed lines–. The envelope –continuous line– marks the optimal value of λ according to each metric for each SNR value and is plotted separately (right). The value of λ used to reconstruct the real data from volunteers is indicated in blue.

profiles than can be appreciated even for high SNR values. The presence of erratic motion grows as the SNR decreases. These results agree with the motion fields obtained with both ME/MC CS based methods, analyzed in Figure 4. For the whole range of SNR values simulated, the groupwise ME/MC technique employed in JW-tTV results in more precise estimations. The differences with the phantom ground-truth, measured as the root mean squared error on a mask over the myocardium, remain almost constant even for the lowest SNR values, while higher errors and faster degradation can be observed for MASTeR.

Figure 5 shows the results obtained from in-vivo data for two of the eight volunteers acquired. Sharper edges and finer details are better preserved with JW-tTV, while contrast between myocardium and blood pool is preserved. For each volunteer, ventricular volumes at end systole (ESV), end diastole (EDV) and ejection fraction (EF) were computed from the short axis slices using Simpson’s rule and manual segmentation on a dedicated software for the three reconstructions. Functional parameters computed from fully sampled Cartesian reconstructions were used as reference values. Bland-Altman plots of the results are shown in Figure 6. The mean and standard deviation of the differences between the functional parameters computed with each method and the reference scan are summarized in Table 1. JW-tTV results in lower mean ESV, EDV and EF differences and lower standard deviation for ESV and EF than MASTeR reconstructions.

4 Discussion and conclusion

In this paper we have proposed a new reconstruction method based on the adaptation of a previously described ME/MC CS based method to comply with Golden Radial, multi-slice acquisitions. We have achieved an undersampling factor that allows us to cover the whole-heart (12-14 slices) with a temporal resolution of 46.4 ms, 13-16 cardiac phases with 16 profiles per frame, in a single BH of 10-13 seconds.

Fig. 3. Phantom reconstructions for different SNR values. Reference noisy images reconstructed from fully sampled Cartesian data (a, b) and from Golden Radial under-sampled data (16 spokes per frame) with MASTeR-GR (c,d) and JW-tTV-GR (e,f) at systole and diastole, respectively. Error images are shown scaled by a factor of 2. Temporal evolution of the dashed line in (a) for the corresponding reconstructions (g, h, i). In JW-tTV-GR reconstructions thin details such as borders of the papillary muscle (red arrows) or the wall of the right ventricle are better preserved and present sharper edges than MASTeR-GR. Errors in MASTeR-GR are mainly concentrated in the edges of the structures, due to edge blurring and motion errors (white arrows), while JW-tTV-GR presents noisy errors due to low SNR. In the temporal evolutions yellow arrows indicate areas where MASTeR-GR presents some erratic motion in the heart even for high SNR values, while in JW-tTV-GR this effect is much less present.

Fig. 4. Absolute value of motion fields estimated with MASTeR and JW-tTV in the 14th phase of the cardiac cycle out of 20, corresponding with large inter frame motion after end-systole for different SNR values (left and center). Phantom ground truth is shown at the left of each image. Red mask represents left and right myocardium. Plot of the RMSE of the motion fields calculated over the myocardium mask for a range of SNR values (right). Larger errors and faster degradation for low SNR can be observed for MASTeR, while JW-tTV shows to be almost immune to the presence of noise.

Method	ESV difference (ml)	EDV difference (ml)	EF difference (%)
MASTeR-GR	1.65 ± 3.62	-4.23 ± 2.62	-2.63 ± 2.61
JW-tTV-GR	-0.28 ± 3.06	-2.98 ± 5.00	-0.73 ± 2.19

Table 1. Left ventricular volumes and EF differences between accelerated and reference scan. Mean $\pm$ standard deviations are shown.

For a given undersampling factor, determined by the single BH application that we pursue, the experiments on the numerical phantom allowed us to study the effect of noise on the reconstructions and on the selection of the reconstruction parameters. How to choose the weight of the sparsity regularization in CS reconstructions is still an open problem of the technique. In this work we try to establish, for this specific application, a choice rule based on the SNR of the data. On real practice, this SNR can be easily obtained by the MRI scanner with a short noise acquisition at the beginning of the examination. However, after visual examination, the final value of the reconstruction parameter had to be increased to get better results. Two aspects could explain this fact: on the one hand, the metrics used (SER and SSIM) are known to be limited when used to quantify image quality perception; on the other, different systematic errors in the real data reconstruction that are absent on the simulated data could affect the results (e.g. errors in the sensitivity maps and in the k-space trajectories). Consequently, a higher regularization term could partially mitigate the effect of the systematic errors mentioned.

Simulated and in vivo results show that better ME and more detailed reconstructions are obtained with the adapted JW-tTV than with MASTeR. For the latter, some erratic motion is observed in the reconstructions that could be explained by the lower performance of the ME procedure in the whole range of SNR considered. These results illustrate how ME errors can propagate to the final solution and the importance of using a robust ME technique.

Fig. 5. Cartesian fully sampled, MASTeR and JW-tTV reconstructions of the data from two of the volunteers acquired. Basal (a,b), medial (c,d) and apical (e,f) short axis views are shown both at systole and diastole. White arrows indicate areas where thinner details are recovered with JW-tTV therefore permitting better delineation of the papillary muscles at systole.

Fig. 6. Bland-Altman plots of the ESV (left), EDV (center) and EF (right) calculated from the fully sampled data (reference values) and from the accelerated single BH acquisition with MASTeR and JW-tTV.

Accordingly with this observation, the erratic motion in MASTeR reconstructions seems to affect the LV functional parameters calculated, introducing a larger mean difference in ESV, EDV and EF, since it can introduce artificial volume variations in the images and makes more difficult to identify systolic and diastolic cardiac phases. It is worth mentioning the higher standard deviation in the calculation of the EDV with JW-tTV. As Figure 6 shows, in MASTeR reconstructions EDV values are systematically underestimated. With JW-tTV that bias does not occur, but underestimation takes place in volunteers with larger EDV. This fact increases the standard deviation of EDV differences in JW-tTV. However, the mean difference remains lower (in absolute value) than the one calculated from MASTeR reconstructions.

For the current MATLAB implementation, the reconstruction times for the single BH acquisitions lasted 12.5 min per slice with all the slices being reconstructed in parallel independently, what clearly limits the practical application of the proposed method. However, an initial reconstruction is provided in approximately 2 min that can be useful to check the correct planning of the acquisition. The introduction of coil compression techniques [5] and the development of a high performance implementation of the algorithm on graphics processing units could significantly reduce these reconstruction times.

Although the number of volunteers is reduced, reconstructed images and functional values computed from the highly accelerated data show preliminary evidence of the feasibility of the extended JW-tTV method to realize a whole heart cine examination in a single BH of short duration. For comparison, the acquisition time in the conventional Cartesian acquisition was between 1.7 and 2.2 minutes, without considering resting intervals between breath holds. This reduction in scan time could drastically impact on patient comfort and medical resources exploitation. However, further studies including intra and inter observer variability and both healthy volunteers and patients with heart disease are needed to fully validate these results.

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