Accelerated cardiac MRI by 2D Fourier Inversion of the Entire Image Sequence

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INTRODUCTION

The image acquisition of cardiovascular MRI can be accelerated due to the fact that the heart takes approximately 10% of the field of view (FOV) from one timeframe to the next. In reduced FOV techniques, Noquist [1] and its application in parallel imaging [2] separate the dynamic region from its static surrounding across the FOV and reconstruct the images by one-dimensional inverse Fourier transform (IFT). To further reduce the data acquisition time, the dynamic region can be reduced to an arbitrary shape that encloses the heart compactly. The whole image sequence of a slice is reconstructed together by two-dimensional (2D) IFT. Our earlier simulation results demonstrate that the proposed method requires 44% of *k-t* space lines, 18% less compared to Noquist, to reconstruct the resized image sequence with no visual difference.

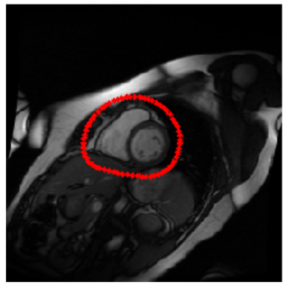

Fig. 1: dynamic region bounded in the red dashed curve.

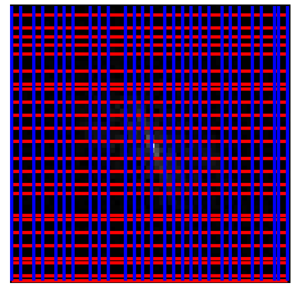

Fig. 2: Phase encoding line samples: horizontal in red at even timeframes; vertical in blue at odd timeframes

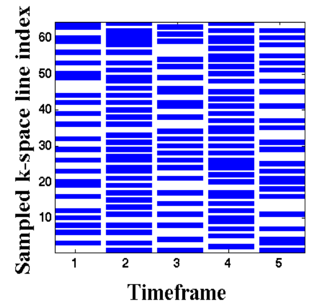

Fig. 3: Sampled *k*-space line indices

METHOD

The dynamic region was identified from a scout image. Fig.1 illustrates the desired dynamic region bounded by a B-spline curve. Only the pixels inside this dynamic region were reconstructed for each timeframe. All the other pixels were assumed to be unchanged and were reconstructed once for the entire image sequence.

A. Problem Formulation

An $N\times N$ image at a time t ($0\leq t<T$) was vectorized as $x_t=[x_{t,S}, x_{t,D}]^{\mathrm{T}}$, then $y_t = [F_{t,S}, F_{t,D}]x_t$, where y_t is the vectorized *k*-space measurements at *t*, $F_{t,S}$ and $F_{t,D}$ are 2D Fourier coefficients corresponding to the static and dynamic pixels respectively. Similarly to Noquist, images of all timeframes were concatenated into a column vector and then restructured such that the static pixels were followed by all the dynamic pixels in chronological order yielding a linear system of the form $Y=AX$:

$$\begin{bmatrix} y_0 \\ y_1 \\ \vdots \\ y_{T-1} \end{bmatrix} = \begin{bmatrix} F_{0,S} & F_{0,D} & 0 & \cdots & 0 \\ F_{1,S} & 0 & F_{1,D} & \cdots & \vdots \\ \vdots & \vdots & \vdots & \ddots & 0 \\ F_{T-1,S} & 0 & \cdots & 0 & F_{T-1,D} \end{bmatrix} \begin{bmatrix} x_S \\ x_{0,D} \\ \vdots \\ x_{T-1,D} \end{bmatrix}$$

where A (size $N^2T\times N_{unknowns}$) is the transformation matrix with each row corresponding to one measurement in *k-t* space. Let N_S and N_D be the number of static and dynamic pixels at a time *t* respectively, then $N_{unknowns} = N_S + N_DT$.

a. b.

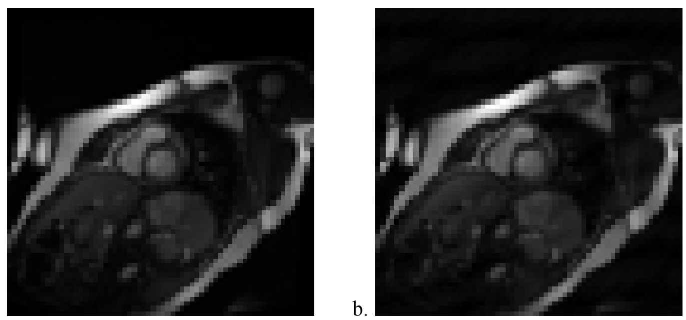

Fig. 4: Original (a.) and reconstructed (b.) images at time 5

B. Phase Encoding Direction

As shown in Fig. 2,The phase encoding alternated horizontally and vertically at even and odd timeframes to provide more *k*-space coverage relative to keeping the phase encoding direction constant for all timeframes.

C. Optimized Sampling

The sequential backward selection (SBS) [3] was used to select $M\leq N^2T$ elements of Y that provide the best possible reconstruction of X. SBS was modified to either sample or skip full *k-t* space lines instead of selecting individual samples. Consequently, in the row elimination process, either N consecutive rows at even times or N equally spaced rows with step-length N at odd times were evaluated and deleted simultaneously. Let A_r denote a candidate set of rows of A for elimination. Using the Sherman-Morrison formula, the multi-row elimination criterion at i^{th} iteration assuming a least-squares reconstruction was

$$C_i = \min\left\{ tr\{\frac{A_{r_i}\left(A^HA\right)^{-2}A_{r_i}^H}{1-A_{r_i}\left(A^HA\right)^{-1}A_{r_i}^H}\}\right\},$$

where the superimposed H denotes Hermitian transpose. The A matrix was updated after each elimination. Note that sample selection can be performed prior one time to image acquisition based on an assumed dynamic region.

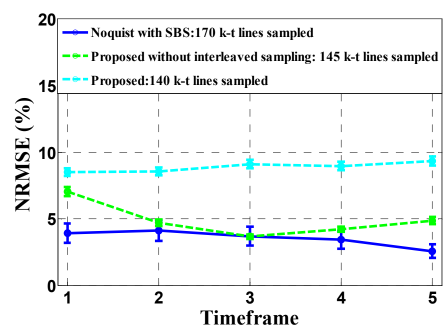

Fig. 5: NRMSE comparison

D. Image Reconstruction

The entire image sequence was reconstructed by solving the reduced $Y=AX$ system using conjugate gradients. The reconstruction quality was evaluated by normalized root mean square error (NRMSE) in the dynamic region.

SIMULATION RESULTS AND DISCUSSION

40 mid-ventricular short-axis cardiac MR image sequences from different subjects were used to simulate the *k-t* space measurements. Images were acquired using a gated SSFP sequence with 8mm slice thickness, 40cm FOV, scan matrix 256×128, flip angle 45°, repetition/echo times 3.81/1.6ms. The phase encoding was along the vertical direction in each sequence. The original reconstructed image size, 256×256×20, was resized to 64×64×5. Fig.3 shows the indices of sampled lines at each timeframe not reflecting the sampling direction. Fig.4 displays the original and reconstructed images at the 5$^{\text{th}}$ timeframe. Fig.5 compares the NRMSE with error bars in dynamic region among the proposed method, the proposed method with the same sampling direction, and Noquist with SBS. NRMSE less than 10% implies subtle differences.

CONCLUSIONS

The proposed method is capable of a 56% reduction in *k-t* space data with negligible reconstruction error. Compared to existing techniques, our method offers greater flexibility to shape the dynamic region and a further reduction in image acquisition time while maintaining comparable NRMSE.

REFERENCES

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