A Novel Reduced Field-of-View Technique for Cardiac MRI: NoQuist

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Abstract

A novel cardiac magnetic resonance imaging (MRI) acceleration strategy called "NoQuist" is presented here. Using this new technique, a more sparsely sampled dynamic image sequence is correctly reconstructed without Nyquist foldover artifact by reductions in the size of the Fourier model of the k-space data for the dynamic image. It does not rely either on substitution or interpolation to arrive at a data set sufficient for reconstruction of the dynamic sequence.

The proposed "NoQuist" method allows reduction of acquisition time in dynamic MRI scans by eliminating the data redundancy that is associated with the presence of static regions in the dynamic scene. A reduction of acquisition time can be achieved, asymptotically equal to the static fraction of the field-of-view (FOV), by omitting acquisition of a strategically selected subset of phase encoding views from a conventional equidistant Cartesian acquisition grid.

1. Introduction

In cardiac imaging of a patient with a severe cardiovascular disease, the shorter breath hold time required to complete a whole movie will increase the success rate in imaging it. This fact, associated with the constant motion of the heart, are the main problems still encountered in cardiac MRI.

Thus, faster acquisition is desirable since it allows the same image to be acquired in a shorter breath hold. Alternatively, it also allows higher-resolution images in the same breath hold time, or complete multi-slice coverage of anatomy in fewer breath holds.

This is why a reduction of imaging time remains of great interest in many applications of dynamic MRI, and mainly in cardiac imaging.

A repeated acquisition of image planes in which only part of the field-of-view (FOV) changes over time is involved in many important dynamic MR imaging applications. A clear example is cine-imaging of the heart, where in a breath-held acquisition, the heart moves during the cardiac cycle, but the lungs and shoulders do not move. For each image of a conventional dynamic acquisition technique, all the raw data space, called *k*-space, will be acquired. Various approaches have been proposed to trade the associated redundancy for reduction in acquisition time or, alternatively, increase temporal resolution. These existing techniques have in common the fact that improvement of acquisition efficiency is achieved by reduction of spatial sampling density, followed by one form or another of temporal interpolation or filtering of the data points in order to eliminate the foldover artifact [1-4]. A common feature of most reduced FOV techniques to date is the associated time savings. If the size of the dynamic region represents a fraction 1/K of the total FOV, the proposed techniques all offer acquisition efficiency improvement by a factor exactly or approximately exactly equal to K.

The novel acceleration strategy introduced here does not rely on substitution or interpolation. A more sparsely sampled dynamic image sequence is correctly reconstructed without Nyquist foldover artifact by reductions in the size of the Fourier model of the *k*-space data for the dynamic image. The image is reconstructed by direct inversion of this reduced model. It is more general and flexible than the approach introduced in [5] because it does not require explicit synthesis of omitted conventional-grid data. As a result, the NoQuist approach accommodates variations in the dynamic fraction of the field of view naturally.

2. Materials and methods

2.1. Theory

The novel technique presented here is based on the formulation of the image reconstruction as an inverse problem. In a single (static) two-dimensional (2-D) MRI image, the unknown image function $f(x,y)$ that is desired to be reconstructed is by good approximation [6] related to the measured k-space data $F(k_x,k_y)$ through a Fourier transform.

The formulation of the image reconstruction problem can be reduced to a 1-D case. This is done for better clarity and because imaging time constraints in spin-warp imaging is typically dominated by sampling in the phase encoding direction in the separable case of 2-D image reconstruction. The Fourier relation between the measured k-space data F and the ideal image f can be expressed as:

$$F(k) = \int_{-\infty}^{\infty} f(x) e^{-2\pi jxk}\, dx \qquad (1)$$

In fact, only a sampled-data version of $F(k)$ is accessible and both the data sampling in k-space and truncation of the sample train have an impact on the reconstructed image. After normalizing the k-space sampling interval and the pixel size in the reconstruction, the reconstructed image $f_{\Pi}(x)$ is related to the data $F_{\Pi}(x)$ by a discrete Fourier Transform (DFT). Its inverse will be called the "data modeling" operation:

$$F(k) = \frac{1}{N} \sum_{x=-N/2}^{N/2-1} f_{\Pi}(x) e^{-2\pi jkx/N} \qquad (2)$$

For clarity, relation (2) can be rewritten in vector format, and the reconstructed image can be denoted by f instead of $f_{\Pi}(x)$:

$$F = M f \qquad (3)$$

In (3) F is an N-dimensional vector whose elements represent the k-space data points. Similarly, the vector $\vec{f}$ represents the DFT-reconstructed image. The elements of the square *data modeling matrix M* contain the DFT coefficients m_{xk}:

$$m_{xk} = \frac{1}{N} e^{-2\pi jkx/N} \qquad (4)$$

Conversely, the image reconstruction matrix M^{-1} defines the image reconstruction:

$$f = M^{-1} F \qquad (5)$$

This formulation requires equal numbers of data points and image points, not taking into account subsequent interpolation to larger image grid sizes.

2.1.1. Dynamic imaging

In the following, a situation where only part of the FOV of the image changes over time will be considered. In cardiac imaging the heart moves during the cardiac cycle but large areas in the adjacent lungs and shoulders often stay in the same position and do not change image intensity. In functional MRI only a small part of the image is expected to respond to a certain activation experiment. The interest in contrast agent uptake imaging often involves relatively small areas of early enhancement in an unchanged remainder of the FOV. In order to take advantage of these observations for accelerating acquisition in dynamic imaging, the total FOV will be partitioned into a "static" part with N_S points that remains unchanged throughout the dynamic scene, and a "dynamic" part with N_D points that contains all pixels that change in intensity during the scan (see Fig.1).

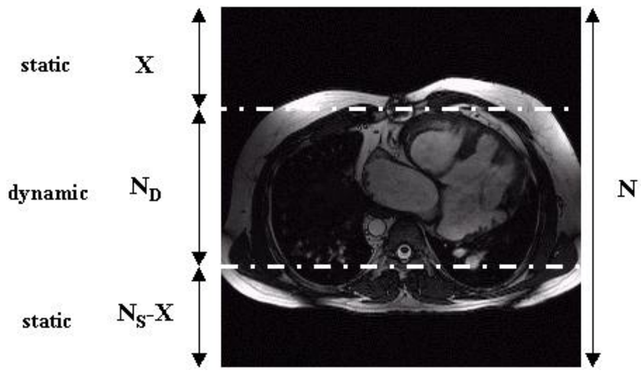


Figure 1. Phase encoding (vertical) FOV of size N with static region of size N_S, and dynamic region of size N_D.

2.1.2. NoQuist algorithm

In order to eliminate the associated data redundancy, "NoQuist" proposes to reduce the data acquisition of the system by minimizing the total number of data points while retaining the critical requirement of M of being invertible (5). If the number of data points is reduced below the number of image points this requirement cannot be met. When the reconstruction matrix is uniquely defined, several reconstructions of the image sequence with exactly this minimum data requirement have been investigated. For the case of cardiac-gated cine-imaging, all time samples should represent a partitioning of the cardiac cycle into T equal intervals. This implies acquisition of:

$$N_{frame} = N_{nq} / T \qquad (6)$$

samples per frame, where N_{nq} corresponds to the number of image points for a NoQuist reconstruction.

2.1.3. Data selection

The condition of the reconstruction problem, quantified by the conditioning number of the data modeling matrix *M*, will be determined by the choice of *k*-space views. Any selection that yields a non-singular matrix *M* is a feasible solution. Several heuristic algorithms for selection of an appropriate subset of *k*-space data points have been designed and tested. As a rule, algorithms have been designed upon the conjecture that for stability the sampling grid should ideally be equally dense across the entire field of view in *k*-space.

2.1.4. Stability and noise propagation

The condition of the data modeling matrix defines the propagation of data noise into the reconstructed image. Assuming a constant noise level during *k*-space acquisition, all elements of the data vector in F will have an identically and normally distributed zero-mean uncorrelated complex-data noise term ν defined by standard deviation σ. A pixel in the image with phase encoding direction coordinate x in the image vector f is reconstructed by a linear system M^{-1} (with matrix elements m_{xk}):

$$f(x) = \sum_{k=0}^{N_{nq}} m_{xk} F(k) \qquad (7)$$

The noise terms $\nu(k)$ in the measured data $\hat{F}(k) = F(k) + \nu(k)$ propagate into image noise $\phi(x)$.

Since ϕ_x is a linear combination of normally distributed random variables, it is also a normally distributed white noise term. Using the expression of its variance, and the expression of the noise variance of a conventional DFT reconstruction, it is possible to calculate a linear noise amplification factor $\Phi(x)$ for NoQuist reconstruction at location *x*, relative to DFT reconstruction but independent of the SNR of the data and the contents of the image:

$$\Phi(x) = \sqrt{\frac{Var\{\phi(x)\}}{Var\{\phi_{DFT}(x)\}}} = \sqrt{\frac{1}{N}\sum_{k=0}^{N_{nq}} |m_{xk}|^2} \qquad (8)$$

2.2. Experimental methods

Construction of the data modeling matrix and all image reconstructions shown here has been implemented in MATLAB 6.1 (MathWorks, Inc., Natick, MA, USA).

All MRI image data shown in illustrations have been acquired on a Philips Gyroscan Intera scanner (Philips Medical Systems, Inc., software release 7.1.2) using a balanced Fast Field Echo (bFFE) steady-state free precession sequence and a five-element phased-array receiver coil designed for cardiac applications. Typical parameters are: 192-256 phase encodings, TR: 3.5 ms, TE: 1.7 ms, flip angle: 60 degrees, FOV: 250-350 mm with oversampling, 12-24 frames per cardiac cycle.

3. Results and discussion

3.1. MRI reconstruction

Figure 2 shows frame 1 of a 12-frame axial movie with 192 phase encodings per frame using a 5-element cardiac surface coil.

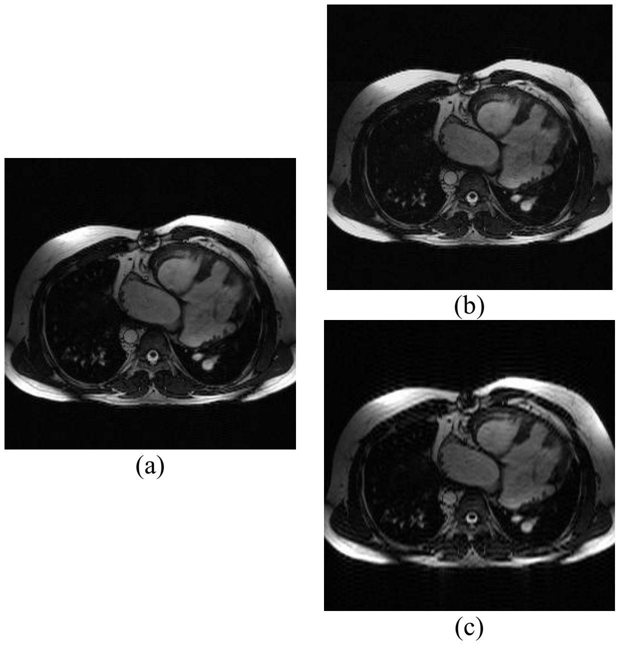
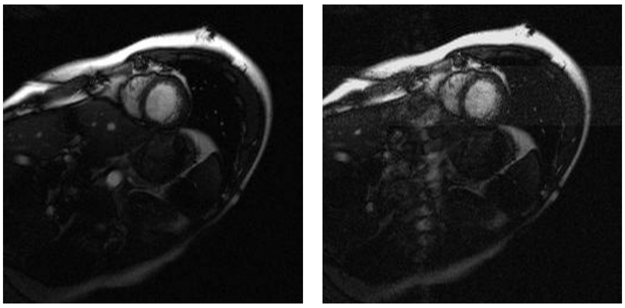


Figure 2. First of 12 frames in an axial bFFE movie with 256x256 image grid from 192 phase-encodings. The center 50% of the FOV is treated as dynamic for NoQuist. (a) Full-grid conventional reconstruction; (b) NoQuist reduced-data reconstruction using 53% of the data; (c) Reduced-data conventional reconstruction from 53% of the data. Images (b) and (c) have equal hypothetical acquisition times.

Figure 2(a) displays the full-grid reconstruction. Figure 2(b) shows the NoQuist reconstruction from 53% of full-grid data or 102 phase encodings per frame. The 50% dynamic portion of the FOV is located into the center half of the FOV. Figure 2(c) represents a conventional reconstruction from 102 low spatial frequency phase encodings. The conventional reduced-data reconstruction shows substantial loss of image detail, preserved in the NoQuist reconstruction generated from an equal number of data points.

3.2. Short axis image

The reconstructions of a short axis slice through the heart of a normal volunteer are shown in Figure 3. Compared to the conventional full-grid (160 phase encoding views) reconstruction (shown in (a)), NoQuist reconstructions with increasing data reduction levels have been investigated. They show faithful preservation of image detail, with signal-to-noise ratio (SNR) decreasing gradually with increasing data reduction. Image (b), using only 25% of the data, still appears to show adequate resolution for evaluation of myocardial functional parameters such as ejection fraction and radial thickening. It also illustrates how moving areas (descending aorta) in image locations outside the region classified as dynamic for the NoQuist reconstruction cause ghost artifacts, similar to other uncompensated motion. If, as in the example, such ghost artifact columns can be rotated away from the areas of interest by appropriate choice of phase encoding direction the reduced-data image may still have undiminished diagnostic value despite 75% reduction in acquisition time.



(a) (b)

Figure 3. First of 16 frames in a short-axis study, acquired using 4 elements of a phased-array coil, with 160 phase encoding views per frame. The phase encoding direction is vertical. A conventional FFT reconstruction on 256x256 image grid is shown in (a). Image (b) shows NoQuist reconstruction with a 75% of data reduction (80% static FOV with N_{frame}=40).

4. Conclusions

The proposed reconstruction by direct Fourier inversion offers a flexible approach to avoid collecting redundant data. As such, it can reduce scan time or breath holding time, reduce degradation from motion during a cardiac phase, or allow cardiac imaging on scanners with slower gradients. No interpolation or substitution of data from other time points is required to reconstruct each frame. This reduces the artifacts from rapid movement in comparison with methods that use interpolation or substitution.

The shorter breath hold time required to complete a whole movie will increase the success rate in imaging severe cardiovascular disease, where difficulty with breath-holding occurs frequently. Alternatively, acquisition of more frames per cardiac cycle in the same acquisition time allows more accurate evaluation of functional indices, even during stress. It can be noticed that a shorter acquisition time yields increased image sharpness, resolution and/or contrast.

A penalty in SNR is associated with this technique. A resolution/SNR trade-off depends on the choice of k_y values acquired.

The proposed method can be implemented immediately in most of today's imaging equipment without requiring additional hardware. Optimization of data subset selection may yield further SNR improvements. This novel technique holds promise for acceleration of phase velocity encoded imaging for quantitative flow mapping.

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