# Toward Robust, Clinically-Practical Single-Breathhold 3D Cardiac Cine MRI with High Acceleration and Rapid Online Reconstruction

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**TARGETED AUDIENCE:** Clinicians & scientists interested in cardiac MRI and fast imaging

**PURPOSE:**

Recently, k-t acceleration methods have demonstrated potential for 3D whole-heart cardiac cine MRI within a single breathhold [1-3]. However, sufficient spatiotemporal resolution and whole-heart coverage require high acceleration, which could result in considerable residual artifacts and through-frame flickering effects. Furthermore, retrospective cardiac gating generates irregular k-t sampling and high acceleration necessitates use of high density coil, both of which pose intensive computation. In this work, we explored a new k-t sampling scheme and several newly developed reconstruction methods to improve image quality and enable fast online processing for highly accelerated 3D cine MRI.

**METHODS:**

Data Acquisition: variable density sampling with k-space energy-based density can improve overall reconstruction accuracy [4]. Random k-space sampling produces incoherent artifacts [5], which is less disturbing than coherent artifacts of regular sampling and could further be suppressed by denoising. A variable-density random k-t (VDR-kt) sampling pattern was developed as below. K-space samples are jittered in [$k_y$, $k_z$] with small random shifts and converted to polar coordinate. The jittered k-space is sorted by radius into k-space rings and each ring is assigned an acceleration factor ($R_{kr}$, increasing linearly from center to outer k-space). K-space samples in each ring are sorted by angle and divided into blocks with $R_{kr}$ samples in each block. Next, the samples in each block are assigned to $R_{kr}$ neighboring cardiac phases in an interleaved manner. By such means, k-space is sampled with k-radius-specific acceleration and randomized time-shifting acquisition (Fig.1).

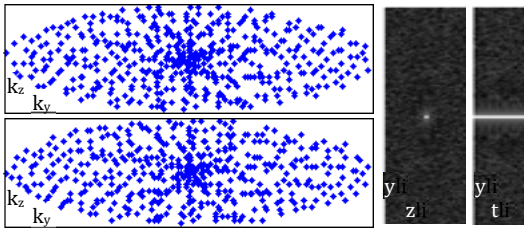


Fig. 1. Sampling at t=1 (left upper) & 4 (left bottom). Aliasing artifacts pattern in [y,z] (middle) &[y,t] (right)

Reconstruction: Fig.2 shows the reconstruction flowchart. A k-t autocalibrating parallel imaging method (kat ARC, [3]) was used for reconstruction. The original k-space data were compressed to 8 virtual channels using a geometry-decomposition coil compression method (GCC [6]). A static tissue removal (SSR [7]) scheme was used to estimate and remove static tissue signals (originated from chest wall and abdomen) in the original data to suppress aliasing and improve condition for the subsequent k-t reconstruction. To address the large number (~2000/phase) of synthesis patterns generated by VDR-kt, a data decoupling calibration (DDC [8]) method with computation insensitive to the number of sampling patterns was used to calculate coil weights for kat ARC data synthesis. After reconstructing the dynamic tissue signals (originated from the heart and aorta) using kat ARC, the static tissue signals were added back to generate the final image.

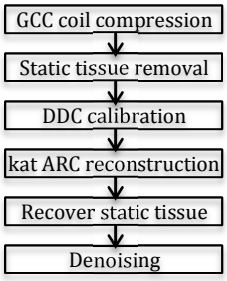


Fig. 2. Recon pipeline

Denoising: VDR-kt sampling produces noise-like artifacts in both space and time (Fig.1). After reconstruction, a wavelet-based filtering was used on coil-combined images to suppress residual aliasing artifacts in each 3D volume frame by frame and next a locally-low-rank (LLR) filter [9] was applied along time to further suppress temporal flickering effects. For LLR, signal dynamics (cardiac motion) in each local block (10x10x5 voxels) is fit into a set of basis functions corresponding to the large single values based on SVD and thus incoherent artifacts & noise corresponding to the trivial singular values are suppressed without impairing motion depiction.

To evaluate the proposed method, 5 pediatric patients were scanned on GE 1.5T (MR450w) using 32-channel cardiac coil. Short-axis 3D cine MRI was performed with regular and VDR-kt sampling. Imaging parameters were: 2x2mm$^2$ resolution, 24 slices, 5mm thickness, 20 phases/cycle, ~8× net acceleration, 20s scan time. Each dataset was processed using different reconstruction methods for comparison on a linux workstation.

**RESULTS:**

As shown in Fig.3, coherent artifacts overlaying the heart is observed with regular k-t sampling (arrows in (a)), while VDR-kt produces more diffuse artifacts (arrows in (b)). SSR (c) further reduces the residual aliasing artifacts in (b) without sacrificing anatomy depiction. DDC (e) provides visually identical reconstruction as conventional calibration (d), but reduces the reconstruction time (all 20 phases with single-core computation) from >1hr to ~10min. GCC (f) further reduces the processing time to <2min without visible image quality loss. Denoising (g) suppresses incoherent artifacts and noise in background air, liver and myocardium and shows reduced temporal flickering effects when played in cine mode.

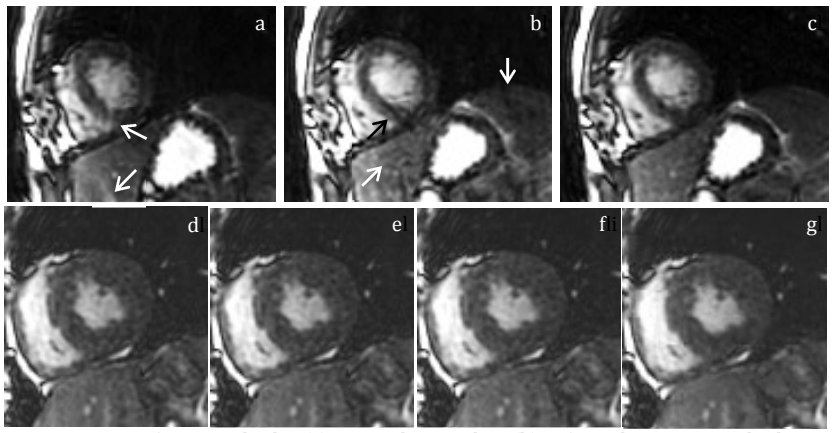


Fig. 3. Images from regular k-t (a) & VDR-kt sampling (b) & from the same VDR-kt data using SSR (c). Images reconstructed using conventional calibration (d), DDC (e), GCC(f) and after denoising (g). [are top and bottom rows from the same dataset?]

**CONCLUSION:**

This work developed a new VDR-kt sampling scheme and an online reconstruction pipeline for kt-accelerated 3D cine MRI. Our results show VDR-kt with SSR is promising for more robust 3D cine MRI with high acceleration. Data-decoupling calibration and coil compression enables <2min reconstruction with single-thread computation. With VDR-kt sampling, denoising could further suppress residual artifacts and flickering effects with rapid post-reconstruction processing.

**REFERENCES:** [1] Tsao, MRM 2003:1031; [2] Huang, MRM 2005:1172; [3] Lai, ISMRM 2009:766; [4] Hennig, Euro Rad 1999:1020; [5] Lustig, MRM 2007:1182; [6] Zhang, MRM 2013:571; [7] Lai, ISMRM 2013:128; [8] Lai, ISMRM 2012:4245; [9] Liang ZP. ISBI 2007; 988-991