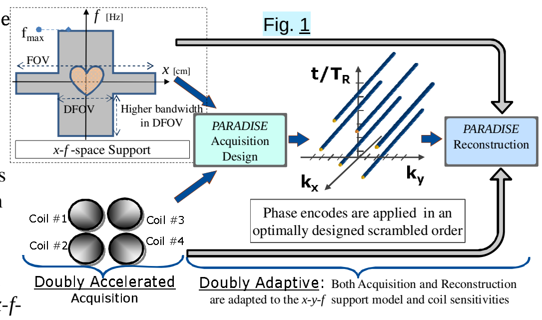
REAL-TIME SHALLOW-BREATHING CARDIAC MRI USING PATIENT-ADAPTIVE PARALLEL IMAGING

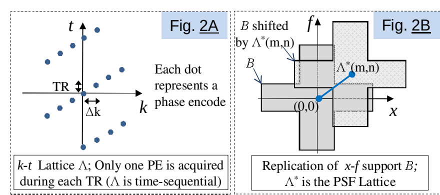
B. Sharif[1], J. A. Derbyshire[2], A. Z. Farnesh[2], R. J. Lederman[2], and Y. Bresler[1]

[1]Coordinated Science Lab, Department of Electrical & Computer Engineering, University of Illinois at Urbana-Champaign, Urbana, IL, United States, [2]Cardiovascular Branch, NHLBI, National Institutes of Health, DHHS, Bethesda, MD, United States

Introduction The goal of real-time cardiac MR imaging is to reconstruct high resolution and high quality images from MR data acquired with no cardiac synchronization (ECG triggering). Patient-Adaptive Reconstruction and Acquisition Dynamic Imaging with Sensitivity Encoding (PARADISE) [1,2], is a non-gated cardiac MR technique that optimally adapts data acquisition and image reconstruction to characteristics of the imaged slice and receiver coil sensitivities. PARADISE provides performance guarantees on achievable SNR and spatio-temporal resolution and enables high SNR artifact-free imaging for such resolutions. MR data acquisition and image reconstruction are typically formulated using Fourier transform (FT) theory, i.e., the classical *k*-space relationship. However, in dynamic imaging it takes a few milliseconds from collecting one k-space phase-encode (PE) line to the next, i.e., sampling in the PE direction is time-sequential and not instantaneous [3]. Hence, a more accurate model for MR data acquisition is the so called "*k-t*-space perspective" (*k*=spatial frequency, *t*=time) [3-5]. *k-t* sampling results in aliasing of the signal spectrum in the reciprocal domain, i.e., the *x-f* space (*x*=PE direction, *f*=temporal frequency) [4,5]. Several k-t based techniques such as UNFOLD [5] (and UNFOLD-based parallel imaging) and PARADISE use a support region in *x-f*-space. PARADISE is a *k-t*-based parallel imaging technique that adapts a general *k-t* lattice to the assumed *x-f*-space support (instead of using conventional *k-t* sampling patterns). However, the previously proposed PARADISE technique [1,2] is restricted to breathhold imaging since it uses a multi-banded patient-adaptive *x-f* support model [4] that formulates the approximate periodicity of the heart motion. In this work, we propose an alternative PARADISE scheme that allows for slow and shallow breathing (SB). Advantages of allowing for shallow breathing real-time imaging include: (1) Patient comfort: the subject is not required to hold his/her breath and can perform slow and shallow breathing (2) Potential diagnostic information in beat-to-beat heart motion variability as a result of interplay between respiratory activity, vascular, and cardiac dynamics [6]. This paper presents the first implementation of shallow-breathing PARADISE (referred to as SB-PDISE) for real-time cardiac imaging.

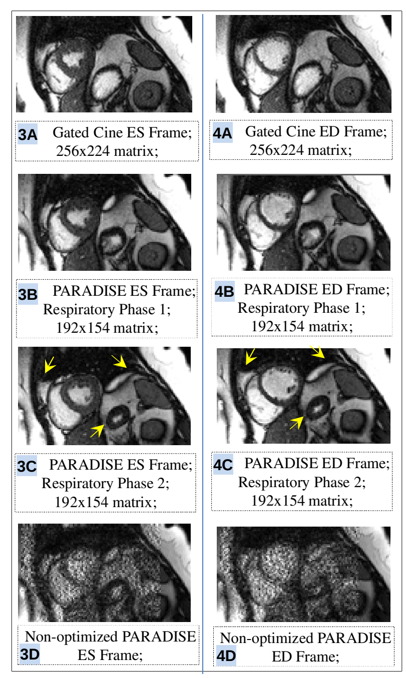
Theory PARADISE is a *doubly accelerated* technique, i.e., in addition to using parallel imaging, the method gains acceleration from a spectral support model (in the *x-f* space). It is also *doubly adaptive* as it uses the support information and coil sensitivities to adapt both the acquisition and reconstruction. Fig 1 summarizes features of the technique. As shown in the Fig, the *x-f*-support support model (denoted by *B*) is characterized by a localized dynamic FOV (DFOV) (similar to the one used in UNFOLD [5]) and a temporal bandwidth that is varying in *x*. Unlike the original PARADISE scheme [1,2], the support model in SB-PDISE does not depend on the specifics of the subject's heart dynamics (heart-rate, etc.). However, as noted in Fig 1, the *k-t* acquisition is adapted to the assumed *x-f* support in order to optimize the expected reconstruction SNR (or equivalently the average spatio-temporal g-factor) [1]. As in other *k-t*-based dynamic imaging techniques, the acceleration in PARADISE results from multi-fold undersampling (relative to the Nyquist rate) provided by the *k-t* sampling scheme. Based on sampling theory, *k-t* sampling on a lattice (Fig 2A) will result in aliasing of the object's spectrum in the *x-f*-space (Fig 2B) and the replication pattern is on the corresponding point-spread function (PSF) which is the 2D FT of the sampling lattice [4,5]. Given a fixed spatial resolution, a *k-t* lattice Λ shown in Fig 2a is parameterized by (TR, Δk) [4]. Fig 2B depicts two representative replicas of the *x-f* support (Fig 1), one centered at the origin and another on a grid point Λ^*(m,n) of the PSF lattice Λ^*. The degrees of freedom in designing the k-t acquisition scheme are: (1) Repetition time (TR) (2) PE step size (Δk) (3) Phase-encode (PE) ordering (possibly scrambled). The *k-t* acquisition lattice is optimized so that the resulting overlap patterns will correspond to well-conditioned (low g-factor) SENSE equations for *x-f*-space reconstruction. Finally, PARADISE reconstruction recovers the signal in the *x-f* domain by undoing the aliasing using sensitivity encoding and utilizing the support model *B* as prior information.





Methods MR imaging (N=4) with informed consent was performed under the NHLBI IRB using a 1.5T Siemens Avanto scanner with a 32-element cardiac array. Initially, a gated segmented SSFP cine scan during a single breathhold was acquired (|FOV|=420mm square, matrix=256x224, 30 phases, TR=3.5ms, temporal resolution=28ms, GRAPPA rate 4). A customized SSFP pulse sequence was developed to allow for operator-defined TR and ordering of PEs -- hence capable of acquiring a general *k-t* lattice (Fig 2A). MR data for the non-gated SB-PDISE scheme with a 192x154 image matrix (FOV as in cine) was collected during shallow breathing (scan time=29.2s). The *x-f* support model was chosen as in Fig 1 with a centered DFOV of size |DFOV|=0.55*|FOV| and max frequency of 3.1 Hz. The SB-PDISE *k-t* lattice design algorithm [1] was run to find the *SNR-optimal lattice*. The search space for (TR, Δk) was limited according to the acquisition and SSFP pulse sequence specifications (resolution, min TR, etc). Computation time was < 30s and each *k-t* lattice was immediately preceded by a conventional coil-profile calibration acquisition to provide the final *k-t* schedule.

Discussion Figs 3 and 4 shows the reconstructed end-systolic (ES) and end-diastolic (ED) frames, respectively. Reconstructed frames for the SB-PDISE scheme acquired according to the SNR-optimized *k-t* lattice are shown in Figs 3B, 3C, 4B, and 4C. Using breath-held gated cine results (Fig 3A,4A) as a reference for shallow-breathing real-time scans, it is seen that SB-PDISE images are visually artifact-free. The two respiratory phases clearly demonstrate through-plane motion as noted by the arrows in Figs 3C,4C. Figs 3D,4D show SB-PDISE reconstruction using a *x-f* support model with a maximum frequency of 4 Hz. These frames are very noisy because the acquired *k-t* lattice was optimized using a different support information (noted above), hence the reconstruction SNR with the modified support is suboptimal. It can be shown that the SB-PDISE image sequences capture beat-to-beat motion variability and the rapid ES wall thickening (i.e., have high temporal resolution). This is expected and matches previous studies [2]. In conclusion, the *in-vivo* results demonstrate the effectiveness and feasibility of the SB-PDISE scheme. In contrast to the previously proposed technique [1,2], the support model is not subject-adapted; hence the optimum sampling pattern can be fully pre-computed. In order to provide artifact-free images, the breathing pattern has to be: (1) shallow (for the support model to be accurate) and (2) slow (so the coil sensitivities during scan time are close to the calibration).



3A Gated Cine ES Frame; 256x224 matrix;
4A Gated Cine ED Frame; 256x224 matrix;
3B PARADISE ES Frame; Respiratory Phase 1; 192x154 matrix;
4B PARADISE ED Frame; Respiratory Phase 1; 192x154 matrix;
3C PARADISE ES Frame; Respiratory Phase 2; 192x154 matrix;
4C PARADISE ED Frame; Respiratory Phase 2; 192x154 matrix;
3D Non-optimized PARADISE ES Frame;
4D Non-optimized PARADISE ED Frame;

References [1] Sharif, Bresler, Proc. IEEE ISBI 2007, pp. 1020-23. [2] Sharif *et al*, ISMRM 2007(15), p.151. [3] Liang, Lauterbur, IEEE Trans. Med. Imaging 13:677-86, 1994. [4] Aggarwal, Bresler, Inverse Problems vol. 24, pp. 045015, 2008. [5] Tsao, MRM 47:202-07, 2002. [6] Elstad *et al*, J. Physiol. 2001;536;251-59.

Acknowledgements: Dr. P. Kellman (coil profile estimation), Drs. R. J. Lederman, A. Faranesh, A. George (experiment support). Financial support: Beckman Institute, UIUC.