K-t Radial SPARSE-SENSE: Combination of Compressed Sensing and Parallel Imaging with Golden Angle Radial Sampling for Highly Accelerated Volumetric Dynamic MRI

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Introduction: Compressed sensing (CS) [1] has recently emerged as a valuable technique for speeding up data acquisition in MRI by exploiting the sparse representation of MR images. However, the performance of CS using a conventional Cartesian trajectory is limited because undersampling can only be employed along the phase-encoding dimensions, thus both sparsity and incoherence cannot be exploited along readout dimensions. Radial sampling is an attractive alternative for CS due to its unique properties such as dense oversampling in the center of k-space and high incoherence in multiple directions, which enables exploitation of sparsity and incoherence along frequency encoding dimension [2-3]. Additionally, it is known that radial trajectories are less sensitive to motion, allowing for better performance in capturing dynamic information. The use of the golden angle approach in dynamic radial MRI, where uniform coverage of k-space is obtained by grouping a number of consecutive spokes, allows for continuous data acquisition and retrospective reconstruction with arbitrary temporal resolution. K-t SPARSE-SENSE is a technique recently developed CS and parallel imaging to highly accelerate dynamic imaging by jointly exploiting temporal sparsity and coil sensitivity encoding [4-5]. In this work, we propose k-t RAdial SParse-Sense (k-t RASPS) for accelerated dynamic MRI using stack-of-stars 3D golden-angle radial trajectories. The performance of this technique is demonstrated for highly accelerated 3D free breathing liver perfusion imaging with both high spatial and temporal resolutions.

Methods: ***Golden Angle Radial Trajectories:*** The angular increment of 2 adjacent spokes in a GA radial trajectory is 111.25° [6-7]. Fig 1a shows the acquisition of m spokes, which provides flexibility in reconstruction of dynamic images with arbitrary temporal resolution, and enables a continuous acquisition without considering explicit time frames. Specifically, when the number of spokes selected at each time point is a Fibonacci number (defined as $F_n = F_{n-1}+F_{n-2}$ with initial values $F_0=0$ and $F_1=1$), a relatively uniform coverage of k-space is ensured. Fig 1b is an example grouping 8 spokes as one time point, generating an N=m/8 dynamic series. Extension of GA radial trajectory into a 3D stack-of-stars format is straightforward, as in Fig 1c.

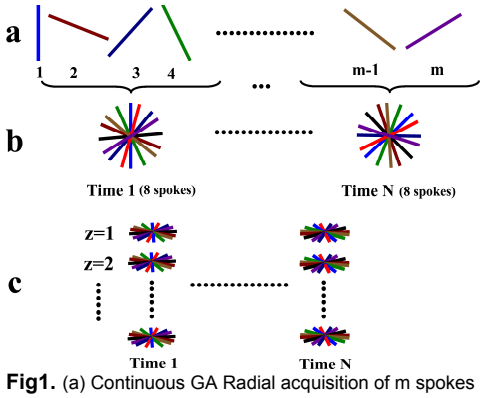

Fig1. (a) Continuous GA Radial acquisition of m spokes (b) A dynamic series (N=m/8) generated by grouping 8 adjacent spokes as one time point

K-t RASPS: K-t SPARSE-SENSE reconstruction is previously developed by minimizing $||E\bullet x-y||_2+\lambda||T\bullet x||_1$, where y is the undersampled k-space data, x is the image to be reconstructed, T is the sparsifying transform, E is an undersampling Fourier transformation operator that incorporates multiplication with coil sensitivities, and λ is a regularization parameter that controls the tradeoff between data consistency and sparsity [4]. The k-t RASPS reconstruction was extended from k-t SPARSE-SENSE by including a radial gridding operation, which is formulated as $||R^{-1}E\bullet x-y||_2+\lambda||T\bullet x||_1$ where R is a gridding operator (NUFFT [8]) that interpolates the Fourier data onto spokes to ensure data consistency in radial k-space.

Liver Perfusion Imaging: Liver perfusion MRI was performed on a healthy volunteer with 0.05 mmol/kg of Gd-DTPA (Magnevist). A 3D stack-of-stars radial FLASH pulse sequence with GA trajectory was employed on a whole-body 3T scanner (Siemens, Verio) equipped with a 12-coil body matrix array. The relevant image parameters include: FOV=380 x 380 mm, base resolution = 384 for each radial spoke, slice thickness = 3 mm, FA = 12°, TE/TR = 1.7/3.9 ms and BW = 620 Hz/pixel. 600 spokes were continuously acquired for each of 30 partitions (60 partitions were prescribed and 50% slice resolution reduction was used in z direction) during free breathing to cover the entire liver and the total acquisition time was 77 seconds. Dynamic series with different temporal resolutions were generated by grouping different (Fibonacci) numbers of adjacent spokes into one time point, which is then gridded into volumes with size 384 x 384 x 30 x N. Spatial resolution is 1mm x 1mm and N is number of dynamic points correlating with the number chosen for each time point. An adaptive compensation method [9] was performed to correct system-dependent gradient-delay errors. Coil-sensitivity maps were calculated using the adaptive coil combination method [10] with the regridded images using all the 600 spokes as coil-calibration reference. The reconstruction was implemented off-line in MATLAB (MathWorks, MA) using a non-linear conjugate gradient algorithm and a total variation along temporal direction was chosen as the sparsifying transformation. The regularization parameter λ was set to be a relatively high value in the first iteration in order to first capture the significant coefficients and then was decreased in each of the following iterations to recover more high frequency coefficients.

Results: Fig 2a shows images of arterial and portal vein peak enhancement in two selected partitions, 13 adjacent spokes was grouped together for each time point (matrix size=384x384, temporal resolution=1.5s). Fig 2b is the plot of signal-time curve of the contrast enhancement in partition 2. Fig 3 shows the same images and curves, but with 8 spokes grouped for each time point (matrix size 384x384, temporal resolution=0.94s).

Discussion: The proposed k-t RASPS reconstruction demonstrates the feasibility of accelerating free breathing 3D liver perfusion MRI while achieving high spatial and temporal resolutions. Results show that k-t RASPS could reconstruct images successfully in high acceleration. This technique can also be applied in other perfusion MRI studies, such as cardiac or breast imaging. Additionally, the technique can be further improved by shifting the trajectory in the slice direction, which will increase the incoherence in that direction.

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Reference: [1] Lustig M, et al. MRM 2007; 58:1182-1195. [2] Block KT. et al.; MRM V. 57, pp. 1086 (2007). [3] Zhang S, et al. JCMR. 2010; 12(1): 39. [4] Otazo R, et al. MRM 2010; 64:767-776. [5] Feng L, et al. MRM 2011; 65: 1661–1669. [6] Koehler T, et al. *IEEE Nucl. Sci. Med. Imag. Conf*, 2004, M10-200. [7] Kasantsev IG, et al. Electr. Notes Discrete Math. Vol 20, pp. 205-216, 2005. [8] Fessler. IEEE T-SP 2003 51(2):560-74. [9] Block KT et al. ISMRM 2011 3026. [10] Walsh et al. MRM. 2000; 43(5):682-90.

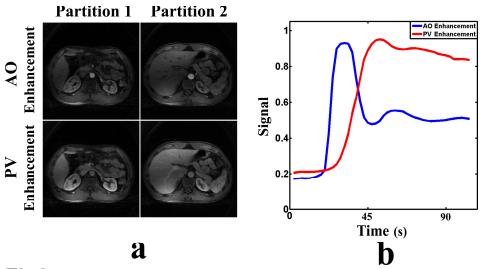

Fig2. (a) Arterial and portal vein peak enhancement in two partitions (13 spokes in each time point) (b) Signal-time curve of the contrast enhancement for AO (blue) and PV (red) in partition 2 (right)

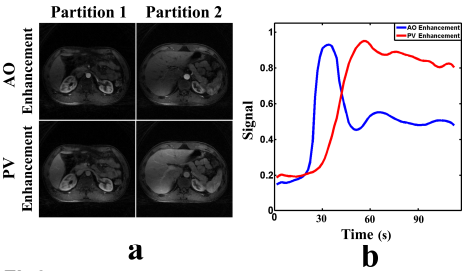

Fig3. **(a)** Arterial and portal vein peak enhancement in two partitions (8 spokes in each time point) (b) Signal-time curve of the contrast enhancement for AO (blue) and PV (red) in partition 2 (right)