# Wave-CAIPI Enables Highly Accelerated 3D MRI

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*Abstract*— **We introduce Wave-CAIPI acquisition-reconstruction technique to accelerate 3D MR Imaging by an order of magnitude, with negligible noise amplification and image artifact penalties. Wave-CAIPI involves playing sinusoidal $G_y$ and $G_z$ gradients during the readout of each phase encoding line while modifying the phase encoding strategy to incur slice shifts as in 2D-CAIPI. This combination spreads out aliasing due to data undersampling evenly in all spatial directions, thereby taking full advantage of 3D coil sensitivity distribution of the receiver coil. By expressing the voxel spreading effect as a convolution in image space, an efficient reconstruction scheme that recovers the undersampled data without data gridding is proposed. Wave-CAIPI enables full-brain gradient echo (GRE) acquisition in 2.3 minutes with 1 mm isotropic voxel size and R=3×3 acceleration, and yields maximum g-factors (noise amplification) of 1.08 at 3T, and 1.05 at 7T. Relative to state of the art accelerated imaging methods, this is a factor of 2 reduction in maximum g-factor at 3T and 7T.**

***Keywords—Accelerated MRI; Parallel Imaging; Image Reconstruction.***

## I. INTRODUCTION

Modern MRI scanners are equipped with multiple receiver coils that are sensitive to signals arising from their vicinity. The spatial encoding information provided by the various receivers can be used to help reconstruct images from sub-sampled data acquisitions, and thereby reduce scan time. The fundamental tradeoff incurred from sub sampling the data is a reduction in the image field-of-view, resulting in signal from multiple locations collapsing/aliasing into each voxel of the image. To create an alias-free image, a parallel imaging algorithm can be used to “unalias” the signals by using the spatial encoding profiles from the multiple receivers. Due to physical limits of multi-channel receivers, pushing the acceleration factor beyond 3 usually degrades the image quality with current techniques, e.g. SENSE [1], GRAPPA [2].

Parallel imaging capability of 3-D imaging can be improved by data reduction in the phase ($k_y$) and partition ($k_z$) encoding directions by shaping the appearance of aliasing artifacts. A possible way to achieve this is to modify the phase encoding strategy to incur interslice shifts in phase encoding direction. This idea forms the basis of 2D-CAIPI (Controlled Aliasing in Parallel Imaging) [3], wherein the variation in the coil sensitivity profiles across the aliasing slices is improved to reduce aliasing artifacts. Alternative approaches for accelerated volumetric imaging include Bunched Phase Encoding (BPE) [4], where a $G_y$ gradient is applied during the readout of each phase encoding line to create a zigzag trajectory that can be reconstructed using Papoulis’s generalized sampling theory to give an alias-free image. BPE has also been combined with parallel imaging [5]–[7] whereby the zigzag trajectory allows utilization of the coil sensitivity variation in the readout direction to improve the reconstruction.

Herein, we introduce Wave-CAIPI, which combines and expands 2D-CAIPI and BPE strategies by playing sinusoidal $G_y$ and $G_z$ gradients during the readout of each phase encoding line. This results in a highly efficient k-space sampling strategy that spreads the aliasing evenly in all spatial dimensions. Since this scheme takes full advantage of the variation in the 3D coil sensitivity profiles, it enables highly accelerated volumetric imaging with low artifact and signal-to-noise ratio (SNR) penalties. Unlike BPE that requires data gridding, we propose an efficient reconstruction scheme that recovers the undersampled data by expressing the voxel spreading effect as a convolution in image space. This formulation is a generalization of the SENSE forward model [1] and can admit additional image regularization. At field strengths of 3T and 7T with 1 mm isotropic resolution, Wave-CAIPI offers up to 2-fold improvement in retained signal to noise ratio (SNR) and image artifact level for in vivo acquisitions when compared to BPE and 2D-CAIPI approaches.

## II. THEORY

Starting from the signal equation, the effect of playing additional gradients $G_y$ and $G_z$ (Fig.1) during each readout line can be formulated as

$$\mathrm{wave}(x,y,z) = \sum_k \mathrm{e}^{i2\pi kx/N} \left(\mathrm{e}^{-i2\pi(W_y(k)y+W_z(k)z)} \sum_x \mathrm{m}(x,y,z)\mathrm{e}^{-i2\pi kx/N}\right) \tag{1}$$

Here, $\mathrm{wave}(x,y,z)$ is the image acquired with the Wave gradients, $\mathrm{m}(x,y,z)$ is the underlying magnetization, $N$ is the number of readout samples, and $k$ is the k-space index along $x$ axis. The variables $W_y(k)$ and $W_z(k)$ denote the k-space trajectory traversed by the $G_y$ and $G_z$ gradients, and are computed by integrating these waveforms (corkscrew trajectory in Fig.1). Recognizing the inner and outer summations as forward and inverse Discrete Fourier Transforms (DFT) in the readout direction, the signal equation simplifies to

$$\mathrm{wave}(x,y,z) = \mathbf{F}_x^{-1} \cdot \mathrm{Psf}(k,y,z) \cdot \mathbf{F}_x \cdot \mathrm{m}(x,y,z) \qquad (2)$$

Here, $\mathbf{F}_x$ represents the DFT operator in the $x$ axis, and $\mathrm{Psf}(k,y,z) = \mathrm{e}^{-i2\pi(W_y(k)y+W_z(k)z)}$ is the point spread function (psf) that explains the effect of the Wave gradients. Viewed from this perspective, the forward model for Wave-CAIPI is a simple convolution that does not require data gridding as in the preceding methods [4], [6]. This expression suggests a simple explanation for the effect of the Wave gradients: Each readout line in the underlying image m is convolved with a psf that depends on the spatial location (y,z) to yield the acquired wave image. The effect of the spatially varying psf is demonstrated in Fig.2, where the combination of the Wave gradients along y and z with interslice shifts gives rise to spreading in all spatial directions.

For accelerated acquisitions, the forward model in Eq.2 is augmented with coil sensitivity information to unfold aliasing sets of voxels. Wave-CAIPI modifies the aliasing pattern of signals in the accelerated acquisition, such that they are from spatial locations that are as far apart as possible. This makes the spatial sensitivity of the receivers in the collapsed voxels maximally different, thereby substantially improving the parallel imaging capability.

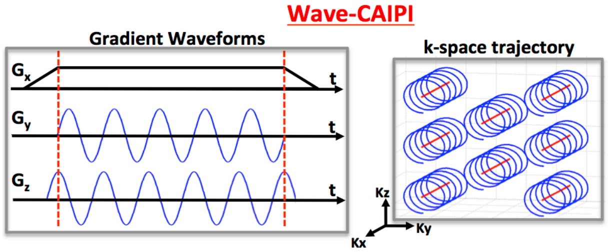

Fig.1 Wave-CAIPI gradient waveforms yield a corkscrew trajectory in k-space

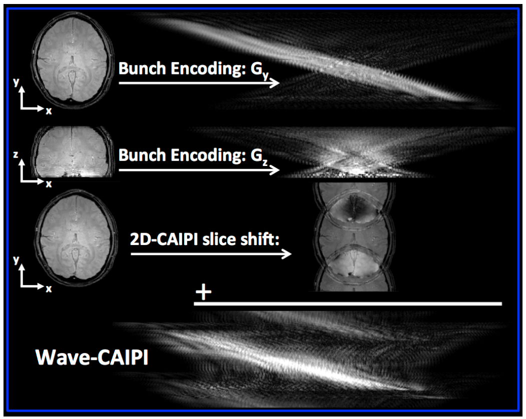

Fig.2 Sinusoidal gradient waveforms $G_y$ and $G_z$ lead to spatially varying spreading in the readout (x) direction. Combination of spreading and inter-slice shifts yields the Wave-CAIPI acquisition.

## III. METHODS

To assess the improvement in parallel imaging capability relative to existing techniques, Wave-CAIPI was compared to CAIPI-2D [3] and BPE [4]–[7]. Two healthy subjects were scanned at 3T and 7T with parameters FOV=240×240×120 mm$^3$, 1 mm$^3$ isotropic voxel size, TR=40 ms, BW=70 Hz/pixel, 32 receive coils. The echo times were 17 ms at 3T and 20 ms at 7T. At each field strength, R=3×3 accelerated gradient echo (GRE) data were acquired using Wave-CAIPI, 2D-CAIPI and BPE sampling, where each acquisition took 2.3 min. Psfs for Wave-CAIPI and BPE were estimated using rapid calibration scans which entail single-slice data acquisition with and without the Wave gradients.

## IV. RESULTS AND CONCLUSION

**At 3T**, Fig.3 depicts the reconstructions and 1/g-factor maps obtained by the three methods. The 1/g-factor maps represent the retained SNR, where the best possible 1/g-factor is 1. Residual aliasing artifacts are visible for 2D-CAIPI and BPE, while Wave-CAIPI produces a clean image. From g-factor analysis, maximum g-factors $g_{max}$ for Wave, 2D-CAIPI and BPE were found to be 1.08, 1.82, and 1.93, while the average g-factors $g_{mean}$ were 1.03, 1.22, and 1.38.

**At 7T**, $g_{max}$ for Wave, 2D-CAIPI and BPE were, 1.05, 1.74 and 2.12, and $g_{mean}$ were 1.02, 1.19, 1.30 (Fig.4).

Wave-CAIPI offers an order of magnitude speed-up in MR scan time with negligible noise amplification and close to 2-fold reduction in maximum g-factor relative to competing methods at 3T and 7T.

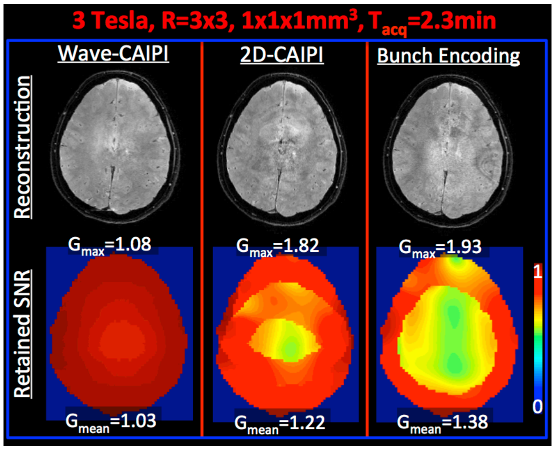

Fig.3 Comparison of methods at 3T with 9-fold acceleration.

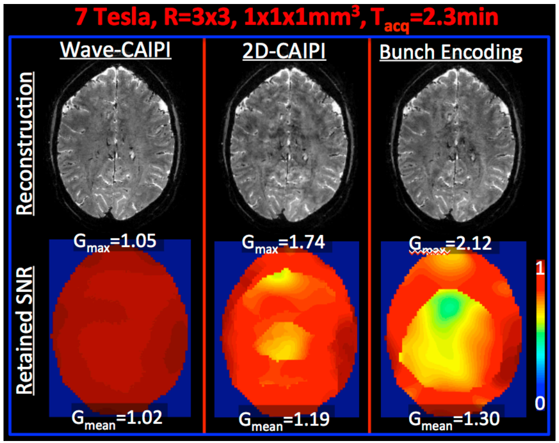

Fig.4 Comparison of methods at 7T with 9-fold acceleration.

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