Accelerated 3D radial imaging with 3D variational regularization

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Introduction: Iterative parallel-imaging methods are highly promising for MR image reconstruction from undersampled data, due to their flexibility to incorporate a priori knowledge using regularization. In particular, when incorporating non-linear L1-based penalty terms, they enable exploiting the compressed-sensing principle to obtain images from highly incomplete data [1]. However, due to the iterative nature, these methods are computationally very expensive and memory demanding. Consequently, most implementations so far used acquisition schemes that allow separating the reconstruction into smaller sub-problems, either by sampling along parallel lines or by stacking identical geometries along one direction, as the case for stack-of-stars sampling. By performing an initial FFT along this direction, images can then be reconstructed slice-by-slice. However, this comes at the expense of losing acceleration capability in this direction, which limits the achievable overall scan efficiency. Furthermore, for certain imaging techniques, e.g., 3D radial ultra-short TE (UTE) imaging, separation of the reconstruction is not feasible. In this work, we present a novel implementation, supported by graphics-processing hardware (GPUs), that treats the whole 3D imaging volume as a single data set and, thus, enables completely arbitrary 3D trajectories, which may use acceleration in any dimension. Further, it enables incorporating fully 3D regularization functionals.

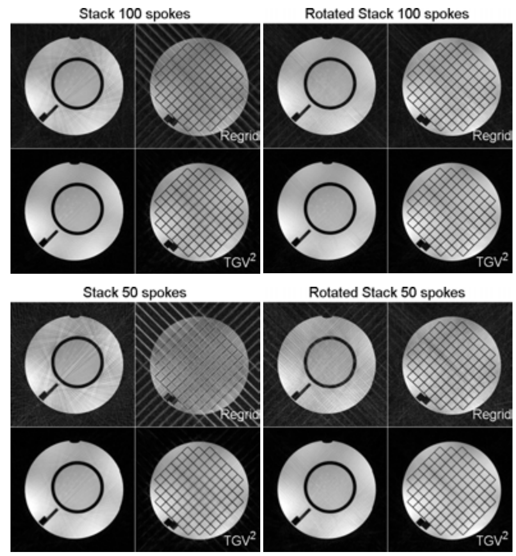

Fig. 1: Reconstructions of a quality phantom using stack of stars and rotated stack of stars with 100 and 50 radial spokes. Both conventional gridding reconstructions and iterative PI reconstructions with TGV^2 constraint are displayed.

Methods: To demonstrate the capabilities of the reconstruction algorithm, phantom and volunteer data were acquired using 3 different sequences: a conventional 3D radial stack-of-stars GRE sequence [2], a 3D radial stack-of-stars sequence where individual discs are rotated relative to each other, and a true 3D radial balanced UTE (bUTE) sequence [3].

Both stack-of-stars sequences used the golden-angle ordering scheme. Sequence parameters for the phantom measurements were: TR=4.2ms, TE=1.9ms, FOV=240mm, matrix=256×256, inplane res=0.9mm, slice thickness=3mm. Parameters for the in-vivo scans were TR=13.2ms, TE=6.1ms, FOV 208mm, matrix=192×192, inplane res=1.1mm, slice thickness 2mm. In the rotated version, each partition used a unique angle offset, which precludes running an initial FFT to separate slices but spreads aliasing artifacts more incoherently across the volume.

The radial bUTE sequence used a continuous sample path through a nearly uniform distribution of points on the surface of a sphere [4]. Sequence parameters were TR=1.62ms, TE=0.1ms, FOV=210mm, matrix=160×160×160, FA=18°. A fully sampled scan (25582 radial 3D projections, scan time 47s) as well as undersampled scans with acceleration factor R=4 (6395 radial 3D projections, scan time 12s) and R=8 (3197 radial 3D projections, scan time 6s) were acquired. Measurements were performed on clinical 3T MR systems (MAGNETOM Trio / Skyra, Siemens AG, Erlangen) using a 12-channel head coil in CP mode and using a 32-channel head coil. To reduce the overall data size for the bUTE data, SVD-based coil compression to 10 virtual channels was done as an initial step of the image reconstruction. Afterwards, coil sensitivities were estimated as described in [5]. In the subsequent parallel-imaging reconstruction, second-order Total Generalized Variation (TGV^2) was used as a variational penalty [5], which inherently adds compressed sensing for incoherent trajectories. A solution is found using a primal-dual extragradient algorithm [6] by solving the optimization problem:

$$\min_u \frac{1}{2}\|F(u)-k\|_2^2 + \alpha \cdot TGV^2(u).$$

F is a subsampled Fourier operator which includes modulation of the object with the sensitivity profiles, *u* is a full 3D volume of images, *k* is the measured *k*-space data, $TGV^2()$ is the penalty term and *α* is a regularization parameter.

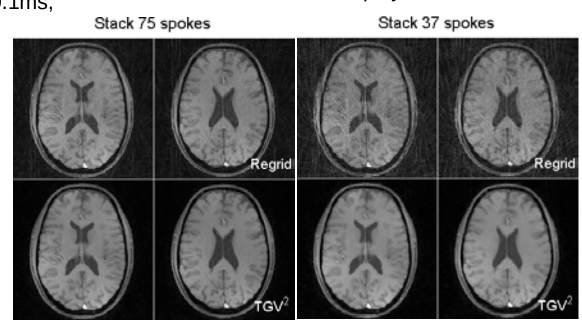

Fig. 2: Reconstructions of the brain of a healthy volunteer using stack of stars with 75 and 37 radial spokes. Both conventional gridding reconstructions and iterative PI reconstructions with TGV^2 constraint are displayed.

Results: **Fig. 1** shows reconstructions of a quality phantom from stack-of-stars data and rotated stack-of-stars data with 50 and 100 spokes per slice using conventional gridding and TGV^2. **Fig. 2** shows stack-of-stars measurements of human brain data from 75 and 37 spokes; the results for 3D radial bUTE are shown in **Fig. 3**.

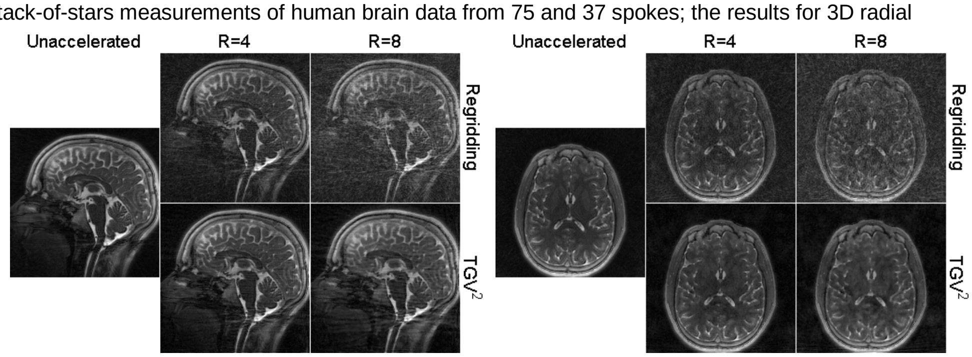

Fig. 3: Sagittal and transversal images from 3D radial bUTE imaging. A fully sampled unaccelerated and accelerated acquisitions with R=4 and R=8 (relative to a fully sampled Cartesian 3D acquisition) are shown for both conventional gridding and TGV^2.

Discussion and Conclusion: This work demonstrates that it is technically feasible to perform iterative parallel-imaging reconstructions that handle complete 3D volumes at once and, thus, enable regularization along all dimensions to improve conditioning of the problem. Due to the flexibility to process any 3D trajectory, higher incoherence of the aliasing patterns can be achieved as verified by phantom scans for rotated stack-of-stars trajectories, which for iterative TGV^2-regularized reconstructions translates into lower residual artifacts or, respectively, into higher achievable acceleration. This effect was previously observed for post-processing based removal of streaking artifacts [7] and is confirmed experimentally here, while a detailed investigation of the potential acceleration gain is still pending and will be presented in future work. Finally, in the case of 3D bUTE imaging, which requires processing of the whole imaging volume at once, the approach enabled reconstruction of high resolution isotropic full brain images from data acquired in only 6 seconds.

References: [1] Block et al., MRM 57: 1086-98 (2007), [2] Chandarana et al. Invest Radiol 46: 648-53 (2011), [3] Diwoky et al. p327, ISMRM 2011, [4] Sam et al., MRM 32: 778-84 (1994), [5] Knoll et al., MRM 65: 480-91 (2011), [6] Chambolle & Pock, J. Math. Imag. Vis. 40: 120-145 (2010), [7] Knoll et al., MAGMA 23: 103-14 (2010).