Reference-Free Parallel Imaging with Phase Scrambling (PIPS)

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INTRODUCTION: A failure to fulfil sampling requirements in MRI results in aliasing. Receiver coil arrays allow accelerating Fourier encoding by relaxing k-space sampling rules. Parallel imaging either interpolates missing k-space lines [1] or undoes aliasing in image domain [2]. Coil sensitivity data are acquired either prior to the actual scan or as a fully sampled part of k-space (auto-calibrating scans, ACS). The need for calibration substantially degrades the time gained, e.g. 4x-accelerated 256-line scan with 32 ACS results in a net gain of 2.67. Furthermore, oft the sampling schemes or the nature of the imaged object do not allow for coil sensitivity calibration.

Since the early days of MRI it is know, that quadratic phase modulation inside the object prior to the data acquisition results in a dramatic change of the k-space appearance, [3,4]. As shown recently, for phase-scrambled acquisitions it is possible to suppress aliasing with a simultaneous reduction of resolution, such that the data amount is preserved [5]. In this work we propose to combine traditional parallel imaging with phase scrambling (PIPS) to enable a reference-free reconstruction.

$$I(x)=g_\alpha(x)\int s(k)\,g_\alpha(k)\,g^*_\alpha(k-x)\,dk. \quad (1)$$

$$I(x)=g_\alpha(x)\,IDFT\left\{DFT\left[s(k)\,g_\alpha(k)\right]\cdot G^*_\alpha(x)\right\}, \quad (2)$$

where $G^*_\alpha(x)=FT\left[g^*_\alpha(k)\right]$

METHODS: PS-MRI reconstruction can be formulated in terms of a modulation and convolution of acquired MR signals with a *chirp* function $g_\alpha(\vartheta)=\exp\left(i\pi\alpha\vartheta^2\right)$ as : (Eq. 1). For $\alpha=1$ Eq. 1 is equivalent to an inverse Fourier transform (the proof is outside of the scope of the abstract). If $\alpha\neq 1$ a scaled image will result. Eq. 1 can be efficiently calculated in Fourier space. If following [5] a *discrete* Fourier transform is applied with the *original* matrix sizes and an analytic Fourier transform of the *chirp* function is used reconstruction becomes: (Eq. 2). It is the combination of analytic and discrete Fourier transforms being the key of enabling the alias-free reconstruction of undersampled PS-MRI datasets. It may be shown that $IDFT\left[G^*_\alpha(x)\right]=g^*_\alpha(k)\cdot w_\alpha(k)$, where $w_\alpha(k)$ is an implicit window function. In our reconstruction an explicit window is used instead, based on a symmetric Fermi filter, to allow flexible control of residual ghost and ringing artefacts.
Simulations were performed in Matlab (The Mathworks, USA) using a modified Shepp-Logan phantom. MRI scans were performed on a Siemens TIM-Trio 3T system (Siemens Healthcare, Germany) in a healthy volunteer using a 3D gradient echo sequence to acquire fully encoded raw data with 256^2 matrix, FOV=256mm, 16 2mm thick partitions, TR=150ms, FA=15°. With the A22 shim current offset to the maximum TE=25ms was required to achieve a sufficient quadratic phase modulation.

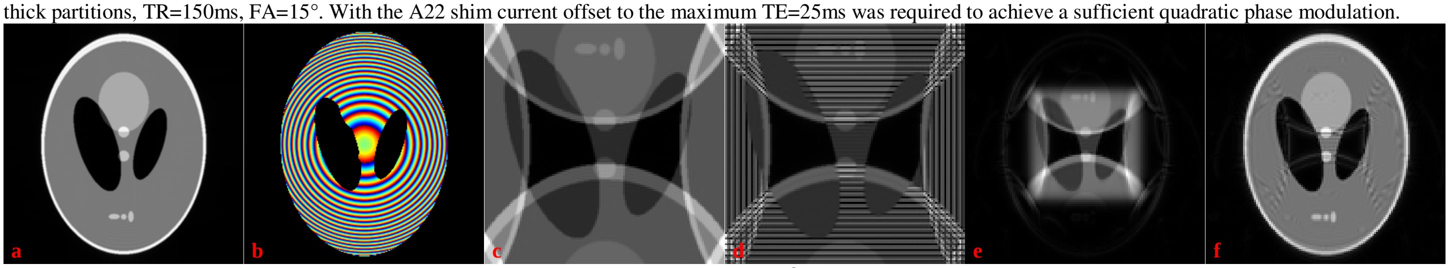

***Fig. 1.** (a) Magnitude and (b) phase of the Fourier transform of the full PS-MRI data (256^2 matrix). Fourier transform of the data 2x undersampled in both directions results in differently appearing aliasing patterns for (c) no phase modulation and (d) PS-MRI with quadratic modulation. (e) Zoom-out reconstruction fails to undo aliasing in data with no phase modulation. (f) Alias-free reconstruction with minor high-frequency ghosts results for PS-MRI. Images (c-f) are on a 128^2 matrix.*

RESULTS AND DISCUSSION: In Fig. 1 simulation results are shown. Given a quite substantial quadratic phase modulation (Fig. 1b), yet not producing any noticeable intravoxel dephasing (Fig 1a), it is possible to unwrap the aliased image (Fig 1d) into a lower resolution image with a minor high-frequency ghosting (Fig. 1f).

In Fig. 2 in vivo images are shown. A centre partition of a single fully-encoded 3D raw dataset was retrospectively undersampled by a factor of 2 to produce all figures. Figs. 2a and 2b are traditional SENSE and GRAPPA reconstructions with coil sensitivities and ACS lines extracted from the original data. In GRAPPA reconstruction the ACS were not included into the final k-space. PS-MRI reconstruction (Fig. 2c) suffers from a very minor high-frequency ghosting and, as expected, is of lower resolution. Single coil images in Fig. 2c were used directly to calculate coil sensitivities for SENSE resulting in Fig. 2d. However, due to the high-frequency ghosts and less apparent local phase errors in the calculated sensitivities, image quality in Fig 2d is degraded. Advanced coil sensitivity extraction algorithms shall be able to tackle this problem. Images in Fig. 2c can also be used to simulate a central part of the fully acquired k-space to serve as ACS for GRAPPA. The resulting images (Fig. 2e) are indistinguishable from the GRAPPA reconstruction in Fig 2b. Apparently GRAPPA weights calculation procedure is less sensitive to the residual artefacts in Fig. 2c.

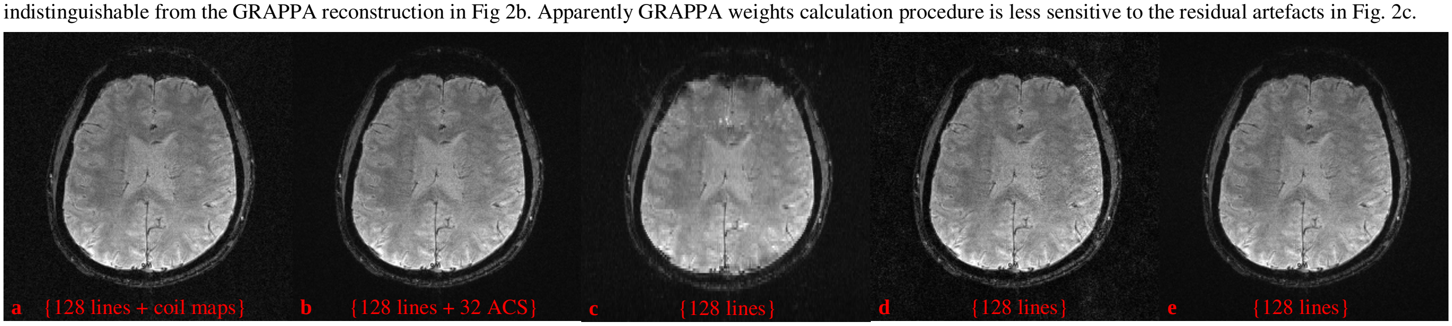

***Fig. 2.** In vivo experiment results. (a) SENSE reconstruction based on 128 k-space lines and coil sensitivity profiles. (b) GRAPPA reconstruction based on 128 k-space and 32 ACS lines. (c) PS-MRI reconstruction based on 128 k-space lines. Note reduced spatial resolution and residual high-frequency ghosting. (d) SENSE reconstruction with coil sensitivities calculated from PS-MR images. (e) GRAPPA reconstruction with 32 ASCs simulated by Fourier-transforming PS-MR images.*

There are numerous ways to induce quadratic phase modulations. The approach used here of offsetting the second order shim currents shows the limitations of the presently available hardware, as (a) pulse sequence has no control of the modulation and (b) available shim coils are not powerful enough requiring the echo time to be prolonged to ~25ms. At this juncture further options based on using tailored RF pulses [6] or 2D/3D selective excitation seem attractive.

PS-FT allows to reconstruct lower resolution images from the undersampled data or in other words to interpolate k-space lines close to k=0. Parallel imaging retains high resolution images based on the same amount of data, but requires coil calibration. It is the combination of both, which appears to be the most advantageous. The immunity of PS-MRI to undersampling comes at a price of locally k-space asymmetries resulting in an increase of ringing artefacts, especially on the FOV periphery. If an excess quadratic phase modulation is applied signal dropouts result. On the other hand, for higher reduction factors stronger modulation is required.

Presented here is a proof-of-concept study aiming to show, that undersampled PS-MRI data from coil arrays contain enough information for the artefact-free full resolution image recovery. A grand unifying reconstruction approach, avoiding the intermediate step of explicit low resolution PS-MRI reconstruction implies further advantages in terms of SNR improvements, as PS-MRI reconstruction affords to recover low resolution fraction of the data with no g-factor penalty.

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