Partial Fourier techniques in single-shot cross-term spatiotemporal encoded MRI

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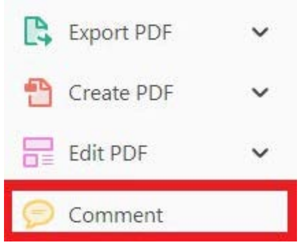

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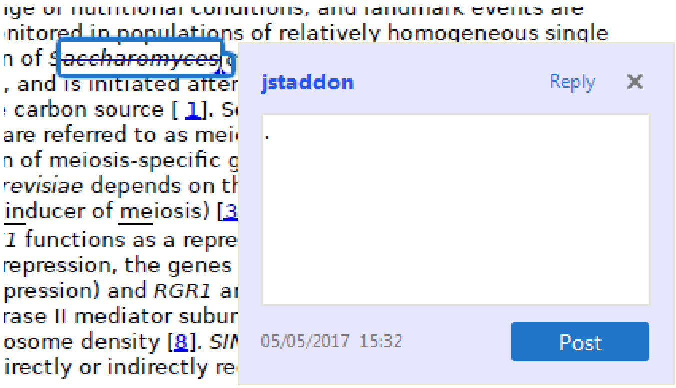

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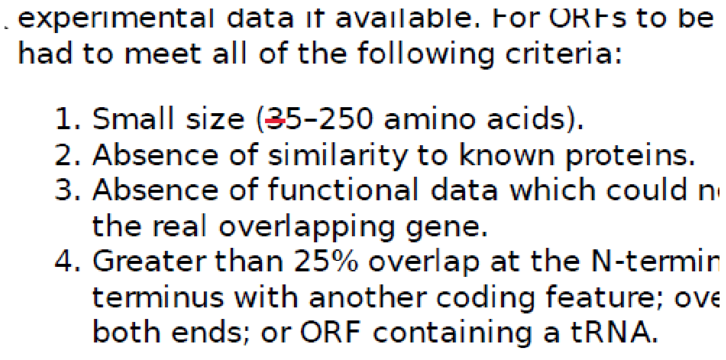

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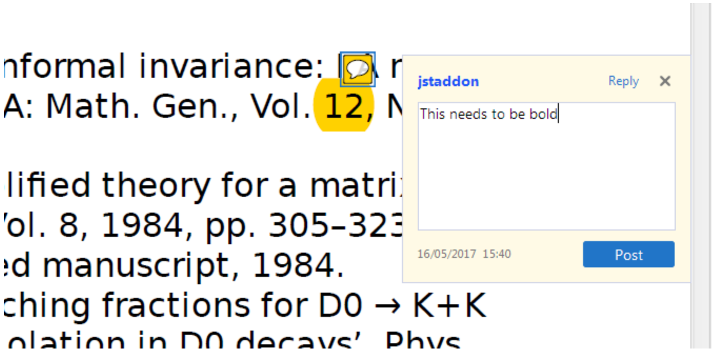

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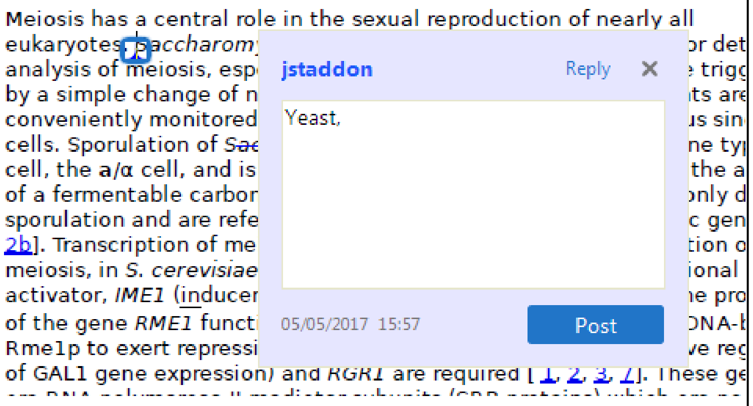

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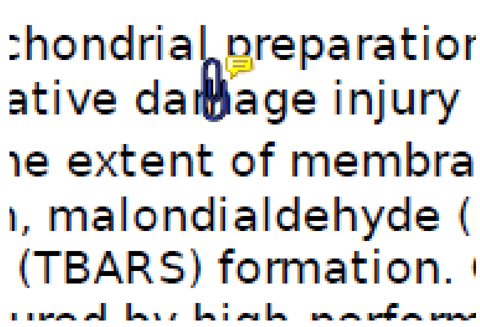

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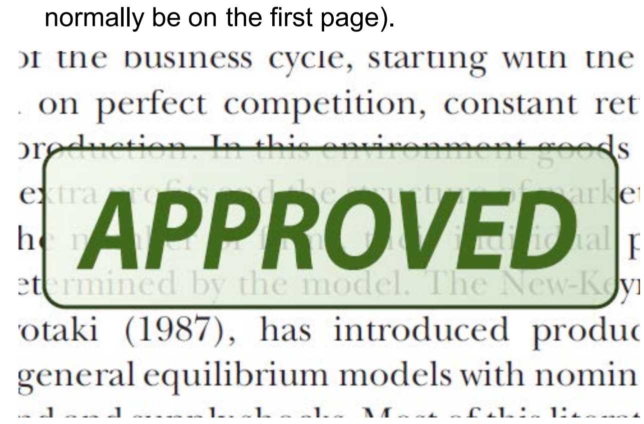

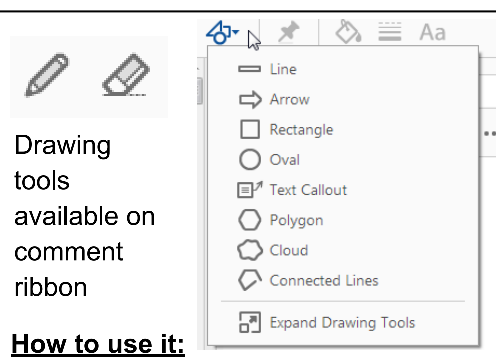

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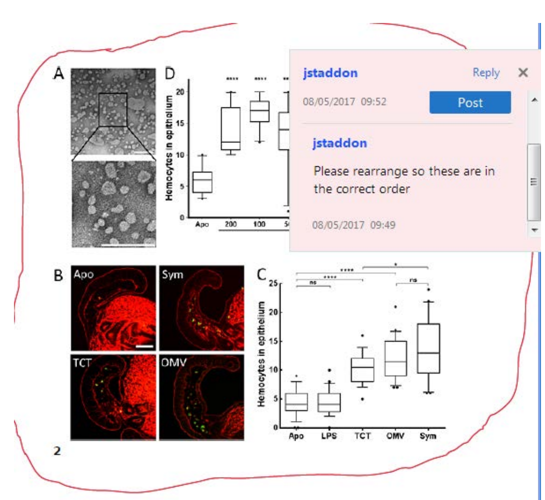

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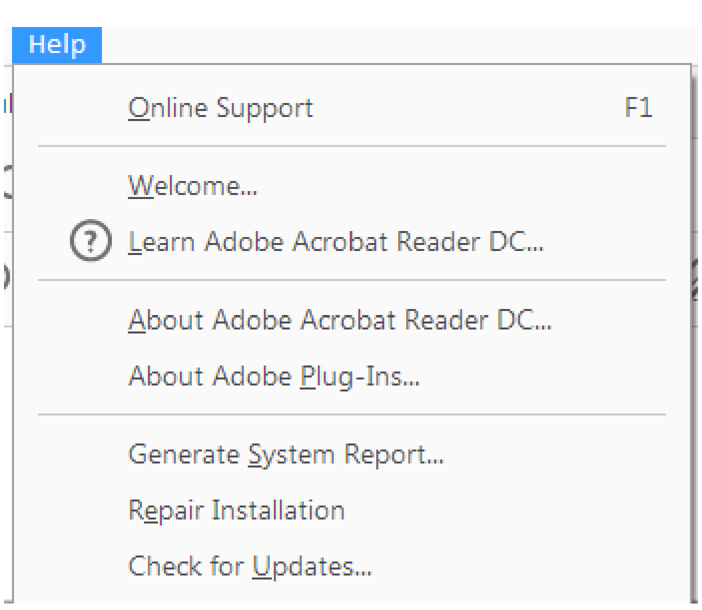

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Partial Fourier Techniques in Single-Shot Cross-Term Spatiotemporal Encoded MRI

Zhiyong Zhang and Lucio Frydman*

Purpose: Cross-term spatiotemporal encoding (xSPEN) is a single-shot approach with exceptional immunity to field heterogeneities, the images of which faithfully deliver 2D spatial distributions without requiring a priori information or using postacquisition corrections. xSPEN, however, suffers from signal-to-noise ratio penalties due to its non-Fourier nature and due to diffusion losses—especially when seeking high resolution. This study explores partial Fourier transform approaches that, acting along either the readout or the spatiotemporally encoded dimensions, reduce these penalties.

Methods: xSPEN uses an orthogonal (e.g., *z*) gradient to read, in direct space, the low-bandwidth (e.g., *y*) dimension. This substantially changes the nature of partial Fourier acquisitions vis-à-vis conventional imaging counterparts. A suitable theoretical analysis is derived to implement these procedures, along either the spatiotemporally or readout axes.

Results: Partial Fourier single-shot xSPEN images were recorded on preclinical and human scanners. Owing to their reduction in the experiments' acquisition times, this approach provided substantial sensitivity gains vis-à-vis previous implementations for a given targeted in-plane resolution. The physical origins of these gains are explained.

Conclusion: Partial Fourier approaches, particularly when implemented along the low-bandwidth spatiotemporal dimension, provide several-fold sensitivity advantages at minimal costs to the execution and processing of the single-shot experiments. **Magn Reson Med 000:000–000, 2017. © 2017 International Society for Magnetic Resonance in Medicine.**

Key words: single-shot MRI; spatiotemporal encoding; xSPEN; resolution enhancement; sensitivity enhancement; partial Fourier transform

INTRODUCTION

Cross-term spatiotemporal encoding (xSPEN) is a novel approach delivering single-scan NMR images with unprecedented resilience to field inhomogeneities (1). Like its spatiotemporally encoded (SPEN) predecessors (2–11), xSPEN relies on imprinting a shaped phase during an initial encoding process, which then serves as the focal point for a subsequent, gradient-driven image readout. In both experiments, this nonlinear phase encoding $\Phi_e(r)$ leads to destructive interferences among signals emitted from neighboring spins, except for those positioned close to positions fulfilling the stationary-phase condition $(\nabla\varphi_e)_{r=r_o}=0$. The action of an acquisition gradient, G_a, which provides to this initial encoding profile an additional evolution phase, $\varphi_a = k_a \cdot r$, with $k_a=\gamma G_a t$ displaces this stationary phase point throughout the targeted field of view (FOV). If properly steered, this will eventually reveal the full $\rho(r)$ spin density over the targeted FOV during a time-domain acquisition. Both SPEN and xSPEN thus differ from echo planar imaging (EPI, (12,13)) in that their image readout occurs in direct, physical space.

SPEN imparts its encoding as a quadratic, y^2, phase modulation, whereas xSPEN relies on a $y.\ z$-type phase (1,14). When applying such hyperbolic encoding, the option arises of activating either the G_a^y or G_a^z acquisition gradients to unravel, respectively, either the $\rho(z)$ or the $\rho(y)$ spatial profiles. The physical basis of how these acquisition gradients allow one to read, in direct space, the spins' profile along an orthogonal axis has been explained elsewhere (1). Such analysis also reveals that utilizing a *z*-axis gradient to both encode and unravel a $\rho(y)$ image enables one to perform an acquisition that can be entirely free from chemical shift or field inhomogeneity effects. This reflects that, rather than viewing frequency dispersions as artifacts that need to be overcome by application of an overpowering external field gradient, this approach to MRI incorporates any disturbing frequency broadening as part of both the initial phase encoding and the subsequent image decoding processes. This capability is particularly valuable when considering single-shot 2D acquisitions, experiments that although central in numerous diffusion- and functional-oriented applications are known to be particularly sensitive to field inhomogeneity distortions (15–17). Figure 1a illustrates one method whereby the xSPEN strategy was adapted for the realization of such single-shot 2D acquisitions. This sequence imparts its hyperbolic phase encoding by turning on a G_z along the slice-selection axis; this is used for exciting a slice of width L_z and is kept on throughout the rest of the scan. In combination with two linearly swept adiabatic inversion pulses (5,18) applied in the presence of a bipolar gradient $\pm G_y$, this results in xSPEN's characteristic $\Phi_e = -Cy.\ z$ phase profile, in which C is a spatiotemporal encoding constant under the experimentalist's control and y,z are positions in the $-FOV_y/2 \le y \le FOV_y/2,\ -L_z/2 \le z \le L_z/2$ ranges. Then, over the course of the acquisition, the continued action of the constant G_z displaces Φ_e's saddle-shaped profile along the *y*-axis. In synchrony with this, an

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oscillating $\pm G_x$ gradient applied along an orthogonal readout dimension explores the k_x-axis in a conventional, EPI-like manner. The mechanism by which the constant application of a G_z gradient delivers an image free from offset-derived in-plane distortions has been discussed in detail elsewhere (1,19). Basically, even in the presence of a shift or inhomogeneity $\delta\omega(r)$, the xSPEN signal collected as a function of the oscillating wavenumber k_x and the acquisition time t can be expressed as

$$S(k_x,t) = \int_X dx \cdot e^{ik_x x} \int_Y dy \cdot \rho(x,y) \cdot \frac{L_z}{1+f[\delta\omega]} \cdot \mathrm{sinc}\left[(-Cy+\gamma G_z t)\frac{L_z}{2}\right]. \quad [1]$$

AQ16 Therefore, rearrangement of these data and 1D Fourier transformation (FT) along k_x leads, apart from potential distortions related to the slice selection and represented by the function $f[\delta\omega]$, to a 2D $\rho(x,y)$ image as a function of t that will be devoid from all offset-derived misregistrations.

Although delivering single-shot images devoid from in-plane distortions, xSPEN's lack of FT along the low bandwidth dimension carries substantial signal-to-noise ratio (SNR) penalties. These penalties are compounded by the constant G_z gradient required by xSPEN, which being AQ17 larger than a usual EPI phase-encoding gradient by a ratio $\approx\frac{FOV_y}{L_z}$, usually will be responsible for the diffusion-related losses of this technique. In the absence of inhomogeneities, this makes single-shot xSPEN less sensitive than methods such as EPI or even its SPEN predecessors—particularly if using the long acquisition times T_a required for achieving high in-plane resolutions. A well-known route to alleviate such effects is the partial FT (pFT, (20,21)), an approach that leverages the properties of the image being sought to reduce the acquisition coverage along one of the k-domains. Indeed, given the real nature of the NMR spectral correlations, in principle it is possible to sample only half the extent of the full k-space and still achieve the same levels of spatial resolution that would arise from AQ18 sampling a full $-k^{max} \le k \le k^{max}$ range of values (22,23). In practice, such maximal reduction in the sampled data AQ19 rarely is achieved, and partial sampling factors $0.6 \le p$ AQ20 ≤ 0.8 are more common. The $T_a \rightarrow p$.). T_a shortening of the overall acquisition times associated to this partial sampling can lead to a considerable reduction in relaxation and in diffusion-driven losses—particularly for constant-gradient sequences such as xSPEN. The question then arises of how to exploit these k-based phase-conjugation arguments in sequences that, like SPEN or xSPEN, are AQ21 based on the hybrid sampling of k_x and of y -domains. The physical basis of pFT experiments along the readout and low-bandwidth dimensions, and demonstrations of pFT's usefulness to achieve resolutions that so far have been out of xSPEN's experimental reach, are presented below.

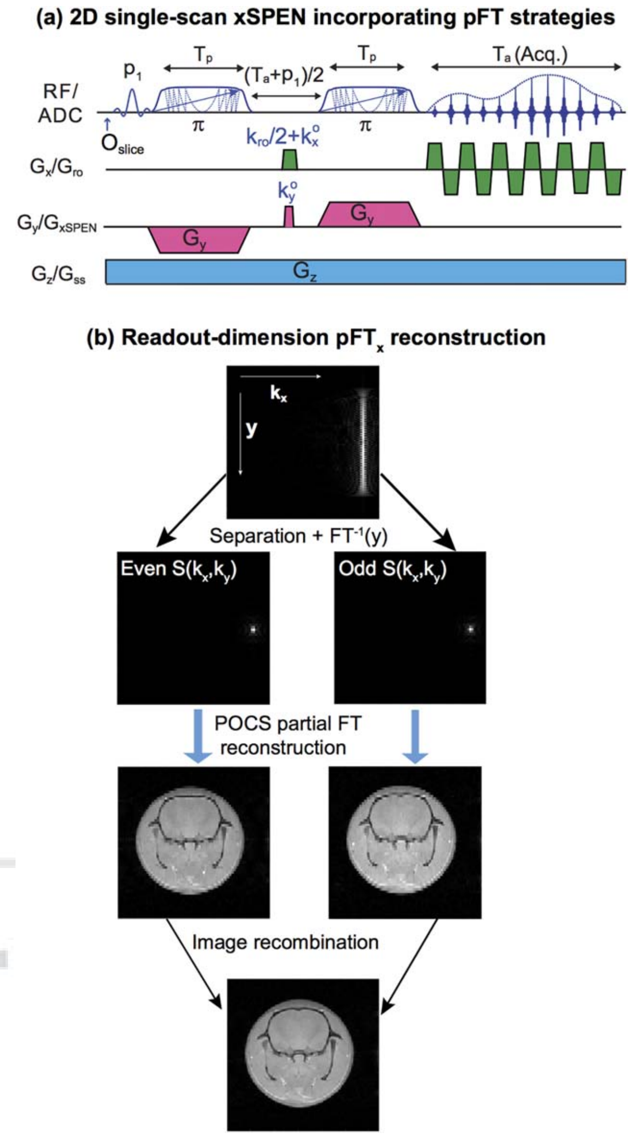

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FIG. 1. (**a**) Single-shot xSPEN sequence incorporating partial Fourier acquisitions by adding short prephasing pulsed gradients along the readout (pFT_x) or xSPEN (pFT_y) axes. (**b**) pFT_x reconstruction involving the addition of a k_y^0 gradient pulse that displaces the $S(k_x,y)$ interferogram (top), separate processing of even/odd $S(k_x,k_y)$ datasets via POCS reconstruction, and subsequent combination (interleaving of magnitude data in image space to avoid phase problems) of the two sets. FT, Fourier transform; pFT, partial Fourier transform; POCS, projection onto convex sets; RF, radiofrequency; xSPEN, cross-term spatiotemporal encoding. AQ43

METHODS

Theoretical background

pFT seeks to retain spatial resolution, while reducing MRI's acquisition times, by estimating part of the k-space data using complex conjugation. Thus, although 1D MRI's inherent resolution depends on the maximal sampled wavenumber $|k^{max}|$, blurring will characterize magnitude images unless a symmetric $-k^{max} \le k \le +k^{max}$ region is sampled. pFT relies on the fact that k-domain data have to fulfill $S(-k) = [S(+k)]^*$ order to calculate the images that would arise from the full $-k^{max} \le k \le k^{max}$ support, while limiting actual samples to a $-(2p-1).k^{max} \le k \le k^{max}$, $0.5 \le p \le 1$ fraction (20,24). When extending these considerations from a 1D axis to a 2D plane, two potential strategies emerge. One is to exploit the $S(-k_x, -k_y) = [S(k_x,k_y)]^*$ symmetry along the directly detected readout domain; the other is to apply it along the phase- AQ22

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encoded dimension. In conventional multi-shot MRI, the latter is the preferred option because it may shorten by a factor p the duration of the experiment. Single-shot techniques such as EPI generally also apply pFT only along the phase-encoded domain because doing so along the readout axis tends to complicate even/odd artifact corrections. In single-shot xSPEN, the readout (x) dimension is k-based, and these even/odd complications are absent because there is no FT along the low bandwidth (y) dimension. Consequently, pFT_x in xSPEN is to some extent simpler than what generally is the case in EPI: xSPEN's pFT_x simply
AQ23 does a 1D phase conjugate reconstruction separately on positive and negative k_x-axis acquisitions and then recombines the two datasets in image space without phase problems to deliver its image (Fig. 1b).

Less straightforward is envisioning how pFT could be exploited along the xSPEN y-dimension. As mentioned, single-shot xSPEN imparts a preacquisition hyperbolic
AQ24 phase-encoding e^{-iCyz}, the stationary point of which is shifted over the course of the acquisition by a constant z-gradient. Such gradient in essence performs an analog Fourier analysis of the encoded data, delivering a y-axis image while simultaneously removing all ΔB_o inhomogeneity effects. This in turn means that an inverse FT of the data collected while under the action of the G_z gradient will be equivalent to a conventionally k_y-encoded MRI acquisition, with $k_y = -Cz$ being the Fourier-conjugate to the y-position. Therefore, in the same way that conventional pFT relies on breaking the echo symmetry of a k_y-domain acquisition by applying a prewinding G_y gradient, performing an asymmetric encoding of the xSPEN image would demand the introduction of a prewinding G_y pulse, even if the image is subsequently unraveled by the action of a G_z. Figure 1a highlights how this route to performing pFT_y along the low-bandwidth dimension can be included in the original 2D sequence by introducing a short prephasing gradient pulse k_y^0. Such prephasing effectively shifts xSPEN's virtual k_y encoding, thereby opening a route by enhancing the y-axis resolution via pFT. To see how this arises, we revisit Equation [1] in the absence of inhomogeneities for a 1D case that for simplicity ignores the k_x readout dimension. Approximating the *sinc* function in that formula as

$$L_z \mathrm{sinc}\left[(-Cy + \gamma G_z t)\frac{L_z}{2}\right] \approx \int_{-\frac{L_z}{2}}^{+\frac{L_z}{2}} dz \cdot e^{i(-Cy+\gamma G_x t)z} \quad [2]$$

enables us to describe the effect of the prephasing pulsed gradient K_y^0 on the detected signal as

$$S[k_z(t)] = \int_Y dy \int_Z \rho(y) e^{-ik_y^0 y} e^{i(-Cy+k_z)z} dz \approx e^{-ik_y^0 y'} r(y') \quad [3]$$

where $y' = k_z/C$ is the coordinate decoded by the action of the acquisition wavenumber $k_z = \gamma G_z t$, and $r(y')$ is a function representing the xSPEN image, given by a convolution of the $\rho(y)$ spin density with the *sinc*-based sampling point spread function. The $e^{-ik_y^0 y'}$ prefactor clearly represents a shift in the k_y-space origin associated with $r(y')$'s inverse Fourier transform signal $S(k_y) \int_Y r(y') e^{ik_y y'} dy'$. In other words, if in conventional xSPEN the maximum y-axis spatial resolution is given by the *sinc*'s width $\frac{2}{CL_z}$, the equivalent k_y sampling associated to the prefactor in Equation [3] will be shifted from $-CL_z/2 \leq k_y \leq CL_z/2$ to an interval of $-CL_z/2 + k_y^0 \leq k_y \leq CL_z/2 + k_y^0$. Hence, an inverse FT of the acquired xSPEN image, a suitable phase-conjugation processing, and a forward FT should yield images arising from an extended k_y support and thus possessing an enhanced y-axis resolution.

Similar pFT considerations would apply to single-shot 2D experiments if the imaging processes along xSPEN and readout axes were fully decoupled; this would be the case if the G_z acquisition gradient would be pulsed in between the bipolar readout gradients. In practice, however, it often is convenient to leave on G_z continuously because this frees not only the low bandwidth but also the readout dimension from field inhomogeneity distortions. The simultaneous action associated with the oscillating G_x readout and the constant G_z gradients acting during xSPEN's 2D single-shot acquisition bring about new features that need to be corrected before attempting a pFT_y. For deriving these features and their corrections, we consider for simplicity an xSPEN evolution that is free from relaxation, diffusion, or field inhomogeneities. The full 2D signal observed in such experiments can be expressed as

$$S(k_x, k_z) = \begin{cases} S^{odd}(k_x, k_z) = \int_X \int_Y \int_Z \rho(x,y) e^{o\phi^{odd}} e^{-i(Cz\cdot y + k_y^0 y)} e^{ik_x x} e^{i(k_z z + \beta k_x z)} dx\, dy\, dz & \text{if } G_x \geq 0 \\ S^{even}(k_x, k_z) = \int_X \int_Y \int_Z \rho(x,y) e^{o\phi^{even}} e^{-i(Cz\cdot y + k_y^0 y)} e^{ik_x x} e^{i(k_z z + \beta k_x z)} dx\, dy\, dz & \text{if } G_x < 0. \end{cases} \quad [4]$$

Here, the integrals extend over the targeted slice and FOVs; k_z and k_x are the acquisition wavenumbers along the low-bandwidth and readout axes; β is a zigzag factor (23,25) reflecting the fact that the k_x wavenumber advances/recedes in conjunction with k_z over the course of the readout oscillation; and ϕ^{odd}, ϕ^{even} are unknown phase terms associated with imperfections in the readout gradients. To adapt the $s(k_y) = \int_Y r(y') e^{ik_y y'} dy'$ notation introduced above to this 2D sampling case, we introduce functions related to what would be the conventional k-space signal associated to this acquisition; that is, AQ25

$$\begin{aligned} S^{odd}(k_x, k_z) &= \int_X \int_Y \rho(x,y) e^{o\phi^{odd}} e^{ik_x x} e^{ik_y y} dx\, dy \\ S^{even}(k_x, k_z) &= \int_X \int_Y \rho(x,y) e^{o\phi^{even}} e^{ik_x x} e^{ik_y y} dx\, dy\,. \end{aligned} \quad [5]$$

These virtual signals arising from positive and negative readout gradients can be used to rewrite Equation [4] as

$$S(k_x,k_z)=\begin{cases} S^{odd}(k_x,k_z)=\int\limits_Z S^{odd}(k_x,k_y+k_y^0)e^{i(k_z z+\beta k_x z)}dz \\ S^{even}(k_x,k_z)=\int\limits_Z S^{even}(k_x,k_y+k_y^0)e^{i(k_z z+\beta k_x z)}dz \end{cases}. \quad [6]$$

Furthermore, because k_z rasterizes the y-axis, this is equivalent to the mixed-domain interferogram

$$S(k_x,y')=\begin{cases} S^{odd}(k_x,y')=\int\limits_{k_y} S^{odd}(k_x,k_y+k_y^0)e^{i(k_z y'+\beta k_y k_x/C)}dk_y \\ S^{even}(k_x,y')=\int\limits_{k_y} S^{even}(k_x,k_y+k_y^0)e^{i(k_z y'+\beta k_y k_x/C)}dk_y. \end{cases} \quad [7]$$

where $y'=k_z/C$.

If not for the β-related terms, one could apply the same arguments that followed Equation [3] to justify the extraction from these data of a pFT_y-enhanced resolution. To appreciate the effects associated to the β-terms, we perform on Equation [7] a final change of variables $k_y'=k_y+k_y^0$:

$$S(k_x,y')=\begin{cases} S^{odd}(k_x,y')=e^{-ik_y^0y'}e^{-i\beta k_x k_y^0/C}\int\limits_{k_y'} S^{odd}(k_x,k_y')e^{ik_y'(y'+\beta k_x/C)}dk_y' \\ S^{even}(k_x,y')=e^{-ik_y^0y'}e^{-i\beta k_x k_y^0/C}\int\limits_{k_y'} S^{even}(k_x,k_y')e^{ik_y'(y'+\beta k_x/C)}dk_y' \end{cases} \quad [8]$$

AQ26 The $e^{-ik_y^0y'}$ phase-modulation term here is, as in conventional pFT_y; however, the new phase terms $e^{i\beta k_x k_y^0/C}$ and $e^{i\beta k_x k_y^0/C}$, affecting the S^{even} and S^{odd} interferograms, evidence a coupling between the k_y^0 echo shifts and the k_x sampling that needs to be removed from even and odd
AQ27 data sets before performing a pFT_y. In practice, we apply this *zigzag correction*, involving a row-by-row multiplication of these a priori known β-phase terms, in conjunction with a removal of the $e^{i\phi^{odd}}$ and $e^{i\phi^{even}}$ phase imperfections that may affect signals collected under $\pm G_x$ gradients (23,26,27). The full procedure is summa-
F2 rized and exemplified in Figure 2. In the present study, the POCS (projection onto convex sets) partial Fourier reconstruction (28,29) was the pFT algorithm chosen to enhance resolution along either the readout or low-bandwidth axes.

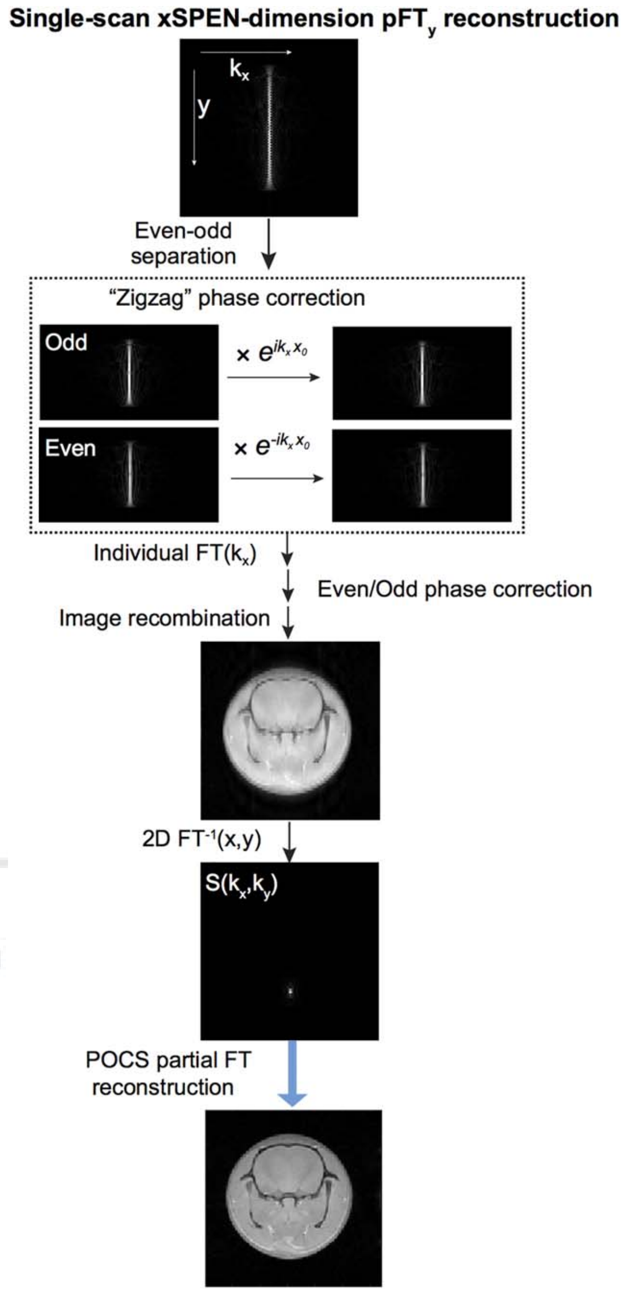

FIG. 2. pFT_y reconstruction involving the addition of a k_y^0 gradient pulse that modulates the xSPEN y-image, separation of even/odd data sets, phase correction by a priori known zigzag effects $k_x x_o$ with $x_o=\beta k_y^0/C$, subsequent correction of residual even/odd phase problems, and final POCS-based partial FT reconstruction of the effective k-domain $S(k_x,k_y)$ data. Notice that whereas fixing even/odd phase problems was not essential in the original xSPEN experiment, if solely a 1D FT along the readout axis was involved, it becomes necessary when implementing the additional manipulations involved in the pFT. FT, Fourier transform; pFT, partial Fourier transform; POCS, projection onto convex sets; xSPEN, cross-term spatiotemporal encoding.

Experimental

Phantom and animal-based acquisitions were carried out on a 7T/120 mm horizontal magnet using a quadrature volume coil and a DD2 Agilent console (Agilent Technologies, Santa Clara, California, USA). Animal protocols and maintenance were done in accordance with guidelines of the Institutional Committee on Animals of the Weizmann Institute of Science (protocol 10790514). AQ28 Spin-echo multi-shot (SEMS) images and SE-EPI experiments were carried out using sequences taken from the scanner's library; all SE-EPI acquisitions required reference "navigator" scans to correct for ghosting along the

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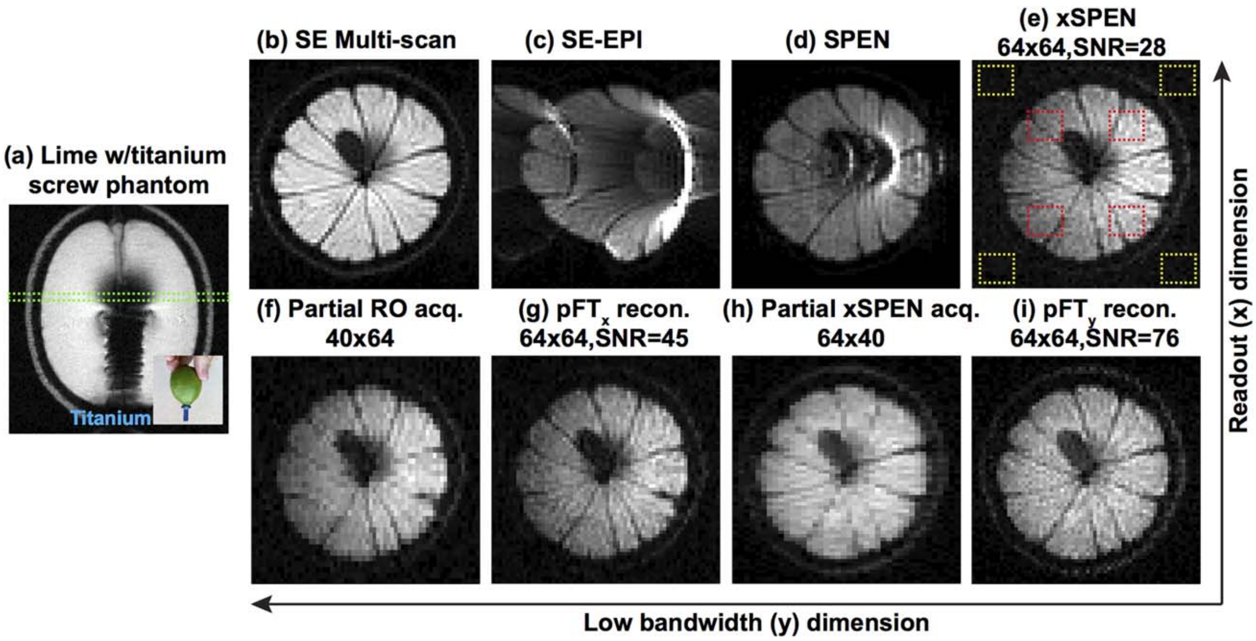

FIG. 3. Representative results arising from a lime phantom incorporating a titanium screw (**a**). (**b**) Spin-echo multi-shot image arising from the green axial slice indicated in (**a**). (**c–e**) 2D imaging results delivered for the same slice by different single-shot sequences with identical FOV and resolution settings. (**f**,**g**) Images from a same acquisition involving partial sampling of the readout dimension, processed with and without POCS reconstruction. AQ44 (**h**,**i**) Idembut with and without pFT$_y$ reconstruction along the xSPEN dimension. Both (**g**) and (**i**) have the same resolution as (**e**) but higher SNR, as evaluated from averaged ratios of the yellow/red squares denoting noise/signal regions shown in panel (**e**). Acquisition parameters: FOV = 40 × 40 mm^2; thickness = 4 mm; repetition time = 2 s; T_a = 22.02, 15.88, AQ45 and 13.76 ms for (**e**), (**f**), and (**h**); time-bandwidth products 2.G_eT_e = 64, 64 and 40 for (**e**), (**f**), and (**h**); chirp bandwidths = 5.8, 8.0, and 5.8 kHz for (**e**), (**f**), and (**h**), respectively. Matrix sizes for images in (**b–d**) were 64 × 64; xSPEN image sizes were as indicated. EPI, echo-planar imaging; FOV, field of view; pFT, partial Fourier transform; RO, readout; SE spin echo; SNR, signal-to-noise-ratio; xSPEN, cross-term spatiotemporal encoding.

phase-encoded dimensions. SPEN and xSPEN imaging experiments were run in this preclinical scanner using custom-written pulse sequences and processing macros that were integrated into Agilent/Varian's VNMRJ (Agi- AQ29 lent Technologies) imaging software; these are available upon request. Human volunteers were scanned on a 3T AQ30 Siemens TIM TRIO platform (Siemens Healthcare, Erlangen, Germany) using a 32-channels head coil. Compared in these scans were SE-EPI sequences taken from the scanner's library against custom-written xSPEN acquisition/processing programs. These experiments were approved by the internal review board WOMC-0091-11 of the Wolfson Medical Center (Holon, Israel) and collected after obtaining informed suitable consents. Main parameters used for setting up the various experiments are detailed in the corresponding figure captions.

RESULTS

F3 Figure 3 illustrates the advantages resulting from the pFT procedures just discussed, when performed on a 7T AQ31 preclinical scanner. In these experiments, a lime was analyzed, onto which a nonferromagnetic titanium screw of a kind usually employed in orthopedic prostheses was inserted axially for exacerbating the field inhomogenei- AQ32 ties. Figure 3a shows a photograph of the screw plus AQ33 fruit, together with a SEMS sagittal image showing the effects of the screw as well as a challenging slice on which further axial analyses where implemented. These compared a SEMS image (usually used as our gold standard) (Fig. 3b) and images collected with SE-EPI with fully refocused SPEN and with the xSPEN sequence introduced in (1). This progression clearly shows the latter's higher robustness and faithfulness (Figs. 3c–3e). Using this single-shot xSPEN image collected with the original sequence as starting point, Figures 3f through 3i illustrate the kind of improvements that can be achieved by implementing pFT procedures. Figures 3f and 3g show images obtained upon reducing the number of points collected along xSPEN's readout segments from 64 to 40. Although a simple FT$_x$ procedure yields a lower resolution vis-à-vis the original 64-points xSPEN acquisition, the pFT$_x$ processing clearly restores this resolution. At the same time, the shortened echo times brought about by the $p = 0.625$ reduction in readout points clearly improves the sensitivity. An even larger sensitivity improvement is observed for identical P values if the pFT is implemented along the low-bandwidth dimension. Indeed, although Figure 3h once again shows that resolution is sacrificed upon reducing the sampled xSPEN lines from 64→40, the procedure in Figure 2 can restore the lost resolution while nearly tripling SNR vis-à-vis the original single-shot xSPEN image (Fig. 3i vs. 3e).

Figure 4 demonstrates another aspect of pFT's sensitiv- F4 ity improvements, this time focusing on tradeoffs between resolution and SNR. Shown in the first row are images recorded for the phantom and slice introduced in

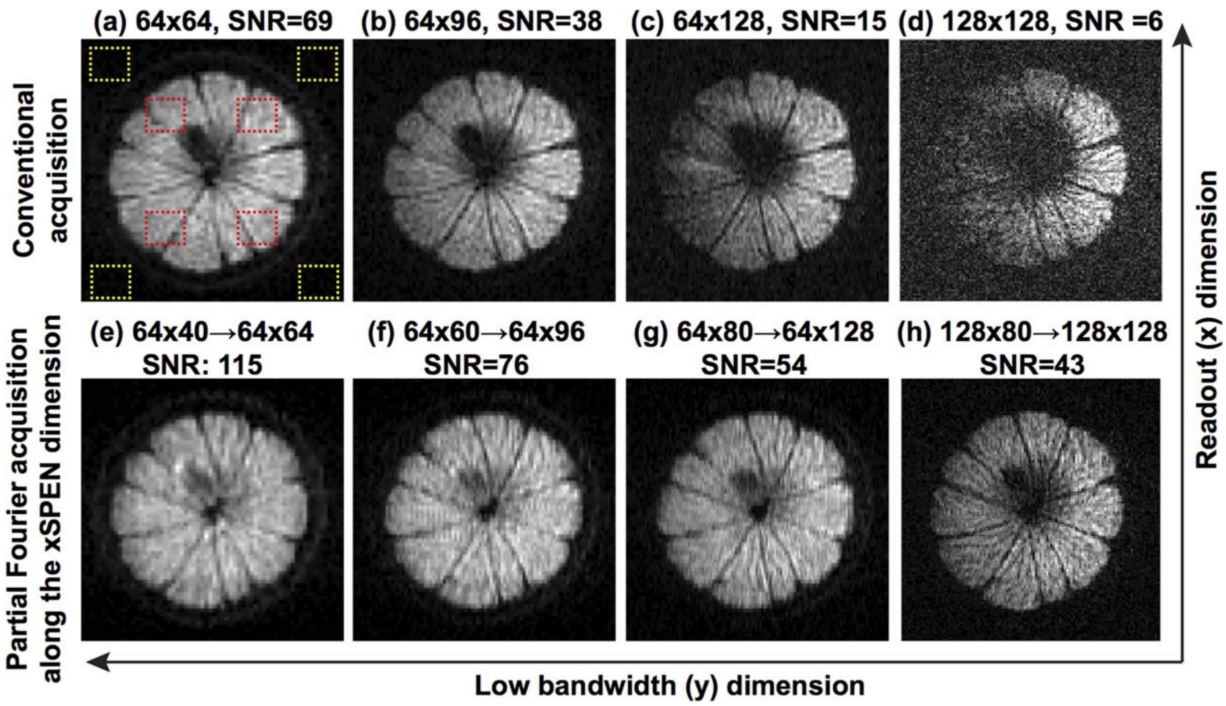

FIG. 4. Sensitivity benefits arising from partial Fourier processing along the xSPEN dimension (pFT_y), as judged by the SNR figures arising from the indicated yellow/red squares on the phantom introduced in Figure 3. (**a–d**) Images acquired with conventional xSPEN, showing how SNR degrades with increasing image resolution due to longer T_as and associated diffusion losses. (**e–h**) pFT_y reconstructed counterparts showing how SNR gains improve with resolution. Acquisition parameters field of view = 40 × 40 mm^2; thickness = 4 mm; repetition time = 2 s; T_a = 22.02, 33.02, 44.03, 76.8, 13.76, 20.64, 27.52, and 48 ms for (**a–h**); time-bandwidth products = 64, 96, 128, 128, 40, 60, 80, and 80 for (**a–h**); and chirp bandwidths = 5.8, 5.8, 5.8, 3.3, 5.8, 5.8, 5.8, and 3.3 kHz for (**a–h**), respectively. Matrix sizes were as indicated (arrows indicate the extent of the augmentation brought about by the pFT procedure). pFT, partial Fourier transform; SNR, signal-to-noise-ratio; xSPEN, cross-term spatiotemporal encoding.

Figure 3b, using the original xSPEN sequence as function of increasing matrix size. This quickly trades SNR for resolution (Figs. 4a–4d), reflecting in part the decreasing voxel sizes, but foremost the diffusion and relaxation penalties incurred upon seeking to increase resolution along the low bandwidth dimension. Images reconstructed using pFT_y clearly can increase SNR vis-a-vis conventionally acquired xSPEN counterparts (Figs. 4e–4h). Moreover, the higher the resolution desired, the larger the SNR benefits arising from relying on a pFT.

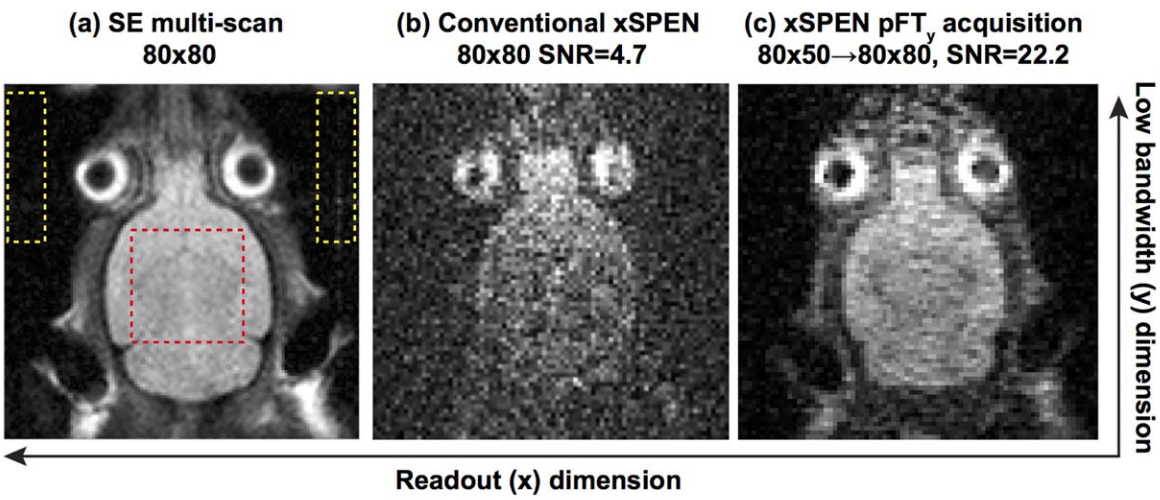

FIG. 5. Sensitivity benefits arising from xSPEN's pFT_y, illustrated with in vivo mouse head scans. (**a**) Reference spin-echo multi-shot image acquired in 2 min 40 s without respiration trigger, and indicating the regions used to evaluate signal (yellow) and noise (red). Images with lower and with improved SNR acquired by single-shot xSPEN MRI without (**b**) and with (**c**) pFT to deliver the same resolution. Field of view = 24 × 24 mm^2; slice thickness = 2.5 mm; repetition time = 2 s; T_a = 32.6 and 20.4 ms; time-bandwidth products = 80, 50 and chirp bandwidths 4.8, 4.8 kHz for (**b**) and (**c**), respectively. Matrix sizes as indicated. pFT, partial Fourier transform; SE spin echo; SNR, signal-to-noise-ratio; xSPEN, cross-term spatiotemporal encoding.

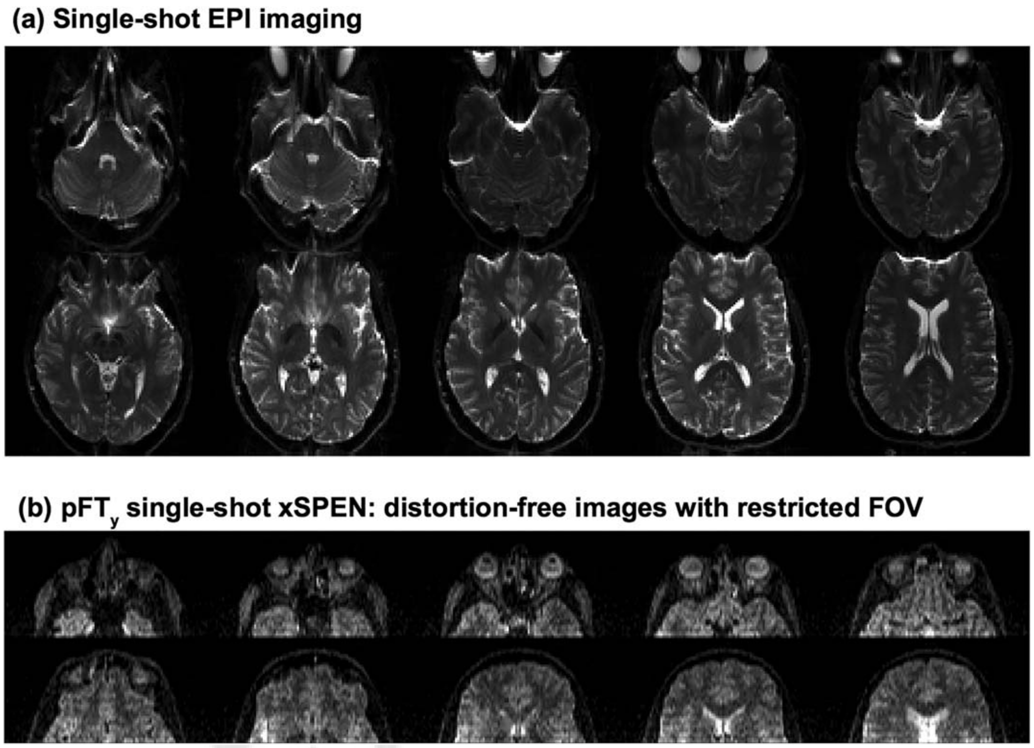

FIG. 6. (**a**) Multislice single-shot EPI images (TR = 2 s) collected on a human volunteer at 3T. FOV = 192 × 192 mm^2, matrix size = 96 × 96, echo time = 77 ms, and T_a = 72.96 ms. (**b**) Corresponding single-shot xSPEN images arising from the same volunteer upon performing a partial FT scan (TR = 4 s) along the spatiotemporal dimension. FOV = 192(RO) × 96(xSPEN) mm^2, matrix size = 96 × 30 reconstructed into a 96 × 48 array by pFT$_y$, T_a = 22.08 ms, time-bandwidth product = 30, and chirp bandwidth = 2.7 kHz. All images possess identical 2 × 2 mm^2 in-plane resolutions. EPI, echoplanar imaging; FOV, field of view; pFT, partial Fourier transform; RO, readout; TR, repetition time; xSPEN, cross-term spatiotemporal encoding.

F5 These advantages are recapitulated in Figure 5, with in vivo experiments comparing SEMS data against single-shot xSPEN images targeting a mouse head. Notice the absence of distortions in regions that typically challenge single-shot applications, for example, near eyeballs and in the ears. Notice as well the large ($\geq 5\times$) SNR improvements brought about by the pFT$_y$ procedure for the $p =$ 0.625 and $300 \times 300\,\mu\text{m}^2$ in-plane resolution targeted
F6 here. Figure 6 illustrates a similar advantage, but for a series of scans collected at 3T on a human volunteer and focusing on the frontal orbital cortex. Due to the susceptibility gradients introduced by the sinuses and eye sockets, single-shot EPI exhibits substantial distortions over various head regions (Fig. 6a). xSPEN yields distortionless images for these regions, but the strong diffusion-driven losses arising when seeking in-plane resolutions better than $2 \times 2\,\text{mm}^2$ render this approach of limited value—even if restricting the FOV to limit the overall acquisition times (data not shown). By contrast, pFT$_y$ enables xSPEN to successfully target this resolution: by sampling only 62.5% of the readout lines, this procedure achieves acceptable SNR and yields undistorted, single-shot zoomed images, free from folding and/or susceptibility artifacts (Fig. 6b).

DISCUSSION

Single-scan xSPEN MRI shows remarkable resilience to field inhomogeneities yet suffers from SNR penalties due to its non-Fourier nature and diffusion and T_2 losses. These losses can be taxing when seeking improvements along the spatiotemporally encoded dimension, for which resolution is given by $\delta y = \frac{2\pi FOV}{T_a \cdot \gamma G_z L_z}$ (1). δy can thus be improved by restricting FOV_y or by increasing the slice thickness L_z, albeit at the expense of losing in- or AQ34
out-of-plane information. Additional parameters available for increasing resolution are G_z, the gradient that in xSPEN stays on for the course of the scan, and the acqui- AQ35
sition time T_a. Owing to xSPEN's refocusing demands, T_a will be proportional to each voxel's position-dependent echo time TE, and hence impart an $e^{-TE/T_2} = e^{-(\alpha T_a)/T_2}$ attenuation for which α is a factor ranging between 2 and 3, and depending on the voxel's y-position. Improving resolution by increasing either G_z or AQ36
T_a will incur in diffusion losses. Based on the Bloch-Torrey model (30,31), these losses can be approximated by an exponential attenuation varying as the square of the gradient and the cube of the free evolution time. On the basis of this, and disregarding for simplicity the effects of the refocusing pulses or of the $\pm G_y$ encoding and $\pm G_x$ readout gradients, xSPEN's diffusion-driven attenuation will be proportional to $e^{-D\gamma^2 G_z^2 (\alpha \cdot T_a)^3/12}$, with D the diffusion coefficient.

pFT decreases these sensitivity losses without sacrificing resolution by collecting a fraction $p < 1$ of the points that would normally be required. This will result in shortened acquisition times that can be implemented by partially sampling either the readout (x) or the

spatiotemporally encoded (y) dimensions. The first of these options restores the original x-axis resolution while acquiring a fraction of the original readout points. Disregarding for simplicity complications associated to ramp sampling or finite gradient slew rates, reducing the number of sampled readout points by a factor of $p<1$ will shorten accordingly the associated acquisition time T_a, leading to a reduction of the T_2-driven relaxation losses by $e^{-p(\alpha\cdot T_a)/T_2}$. However, if this is to be done without a concomitant loss in the y-axis resolution, the relation given earlier for δy implies that G_z will have to increase by a factor $1/p$. The ensuing diffusion-related attenuation factor will therefore be reduced to $e^{-p\cdot D\gamma^2 G_z^2(\alpha\cdot T_a)^3/12}$; because $p<1$, this is clearly an improvement over the original attenuation. Compare this with the case of pFT_y, in which the T_a reduction is achieved by sampling fewer points along y-axis—that is, by applying fewer $\pm G_x$ readout oscillations. The reduction in T_2-driven attenuation losses will remain as for pFT_x; however, the fact that the G_z can now be kept at its original strength without incurring in a δy-degradation means that the diffusion-driven attenuation factor will be reduced to $e^{-p^3\cdot D\gamma^2 G_z^2(\alpha\cdot T_a)^3/12}$. Therefore, although both pFT_x and pFT_y will improve SNR over xSPEN's original realization, pFT_y will lead to a larger improvement due to the $p^3<p<1$ factor arising in the diffusion-weighting exponent. This advantage of pFT_y over pFT_x is compounded by xSPEN's lack of Fourier transform along the xSPEN dimension, which makes
AQ37 the sensitivity of the overall method drop as [# sampled points]$^{1/2}$. By reducing this number by a factor p, adopting the pFTy procedure enhances sensitivity by another factor $1/\sqrt{p}$. All these expectations are confirmed by the data in Figure 3. They also explain the observations in Figure 4 whereby the higher the resolution being sought, the more there is to be gained by adopting the pFT_y procedure. Indeed, in the latter case the increases in resolution called for the use of longer encoding and acquisition times that rapidly increased the diffusion-related attenuation exponent; the larger this exponent, the more remarkable are the benefits of the p^3 pFT_y scaling in the final image SNR. Although an exact quantification of the SNR enhancement introduced by the pFT might benefit from synthetic replica procedures, the large factors evidenced by the experimental data demonstrate the method's usefulness.

In addition to pure SNR considerations, a number of technical factors point to the convenience of choosing partial Fourier sampling along the spatiotemporal rather than the readout axis, particularly when considering
AQ38 xSPEN realizations on humans. One of these pertains to the limited p reductions that can be achieved in clinical scanners along the readout axis, where minimum readout times already are constrained by the maximal slew rates that physiological considerations allow one to achieve. Another limitation derives from the aforementioned need to increase the value of G_z by $1/p$ upon performing pFT_x without decreasing the y-axis resolution. This gradient increase means that chirped pulses with larger bandwidths are needed to cover the original FOV_y and L_z dimensions, resulting in concomitant increases in xSPEN's SAR values. In terms of data postprocessing, however, the reverse considerations apply: pFT_x will barely change the original simplicity of the xSPEN processing, whereas pFT_y requires both even/odd and zigzag phase corrections. Furthermore, to some extent there is an approximation in the assumption made in Figure 2 that these two corrections can be treated independently: a more rigorous analysis of even/odd mismatch problems incorporating the zigzag effect suggests that it may not always be feasible to factor out the phase terms $e^{\pm i\beta k_x k_y^0/C}$ from the integrals introduced in Equation [8]. When this is the case—and this naturally will depend on the nature of the even/odd mismatches—artifacts may arise in images processed, as described above. A general solution to this problem consists of replacing the continuous G_z-driven xSPEN decoding by equivalent gradient blips, acting during the ramp times of the oscillating readout gradient train.

CONCLUSION AQ39

In summary, partial FT approaches acting along either the readout or the spatiotemporally encoded dimensions were introduced and shown to significantly improve the tradeoffs between resolution and SNR in single-scan xSPEN MRI. Details on how to implement these approaches were derived, and associated data processing considerations were introduced. In all cases, examples collected on preclinical and clinical scanners unambiguously demonstrate the advantages of the method without affecting xSPEN's unique resilience to field inhomogeneities. From a practical standpoint, this should readily benefit the potential applications of this new single-scan technique. From a conceptual standpoint, new physical insight had to be introduced in connection to the pFT_y, dealing with the application of orthogonal gradients to k-shift and to acquire a given imaging axis. These insights can in fact be extended to derive altogether new sampling schemes for single- and multi-shot xSPEN, as will be further detailed in upcoming studies.

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AQ6: AU: Review article carefully. As needed, abbreviations deleted at only mention and expanded at first mention.

AQ7: AU: Abbreviation SNR spelled out at only mention in Abstract.

AQ8: AU: Note that single-shot MRI, resolution enhancement, and sensitivity enhancement do not appear in this article.

AQ9: AU: Check for accuracy. There is a superscript period here. Should it be deleted or revised?

AQ10: AU: Meaning of original unclear. Check revised for accuracy and modify as needed.

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AQ17: AU: Larger than a typical?

AQ18: AU: En dash in –kmax revised with minus symbol ($-k^{max}$) throughout. OK?

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AQ21: AU: Should this be revised as k_y?

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AQ23: AU: Rephrase "simply does"

AQ24: AU: Confirm punctuation in e^{-icyz}.

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AQ26: Here is what? Please clarify.

AQ27: This sentence does not make sense. Could it be revised as "The $e^{-ik_y^0 y'}$ phase-modulation term here is [content appears to be missing here], as in conventional $\mathrm{pFT_y}$; however, the new phase terms $e^{i\beta k_x k_y^0/C}$ and $e^{i\beta k_x k_y^0/C}$, affecting the Seven and Sodd interferograms evidence a coupling between the k_y^0 echo shifts and the kx sampling that must be removed from even and odd data sets before performing a pFTy"? See revised punctuation and note comment about missing text.

AQ28: AU: Abbreviation IACUC deleted at only mention in article.

AQ29: AU: Confirm accuracy.

AQ30: AU: Confirm accuracy.

AQ31: AU: Perhaps clarify as "one lime" or "an average-sized lime."

AQ32: AU: OK as revised?

AQ33: AU: Perhaps revise as "screw plus fruit (lime)."

AQ34: AU: Confirm that you meant losing.

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AQ36: AU: Check for consistency. Proportional to each voxel's position-dependent echo time TE ... and based on each voxel's *y*-position?

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AQ44: AU: Check "idem" for accuracy.

AQ45: AU: Check for accuracy. There is a superscript period here. Should it be deleted or revised?

AQ46: Please confirm that given names (red) and surnames/family names (green) have been identified correctly.

Funding Info Query Form

Please confirm that the funding sponsor list below was correctly extracted from your article: that it includes all funders and that the text has been matched to the correct FundRef Registry organization names. If a name was not found in the FundRef registry, it may be not the canonical name form or it may be a program name rather than an organization name, or it may be an organization not yet included in FundRef Registry. If you know of another name form or a parent organization name for a "not found" item on this list below, please share that information.

FundRef name	FundRef Organization Name
Israel Science Foundation	[NOT FOUND IN FUNDREF REGISTRY]
ERC-2016-PoC	[NOT FOUND IN FUNDREF REGISTRY]
Minerva funding	[NOT FOUND IN FUNDREF REGISTRY]
Federal German Ministry for Education and Research	[NOT FOUND IN FUNDREF REGISTRY]
Kimmel Institute for Magnetic Resonance	[NOT FOUND IN FUNDREF REGISTRY]
Perlman Family Foundation.	[NOT FOUND IN FUNDREF REGISTRY]