# Multi-slice ultrafast spatiotemporal encoding (SPEN) MRI by new two dimensional excitation pulses

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**Introduction.** Two-dimensional (2D) excitation pulses are often used for localization in spectroscopic imaging and for in-plane region-of-interest delineation in MRI [1,2]. Recent research has shown that these RF manipulations can also be based on spatiotemporal encoding (SPEN) principles [3,4]. SPEN-based 2D RF pulses show higher robustness against magnetic field inhomogeneities and chemical shift offset effects than traditional *k*-space based counterparts [3]. SPEN is a joint temporal and spatial manipulation that has also been used for single-shot ultrafast MRI –including functional MRI, diffusion and hyperpolarized spectroscopic imaging [5-7]. Fast volumetric SPEN MRI acquisitions, however, are still challenged: usually these are executed using schemes incorporating 180° chirp pulses [8,9] leading to relatively high Specific Absorption Rate (SAR) values if targeting multiple slices in short scan times. The present work merges the benefits of both 2D SPEN-based excitation and 2D SPEN single-shot acquisitions, as it demonstrates a multi-slice ultrafast sequence incorporating a discrete chirp 2D pulse to generate a single-slice quadratic encoding, which is unraveled by a SPEN acquisition. Experiments testing these ideas were demonstrated on phantom as well as on brain volunteer imaging experiments at 3 T. The SAR resulting upon extending these schemes to volumetric multi-slice acquisitions was ca. 60% lower than SPEN scans based on 180°-based counterparts.

**Fig.1 – Multi-slice SPEN using hybrid 2D reciprocal and direct space excitation pulse**
**Slice select along "fast" dimension and discrete chirp pulse in the "slow" dimensions**

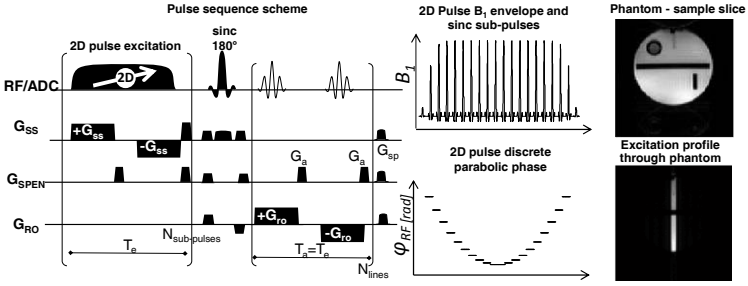
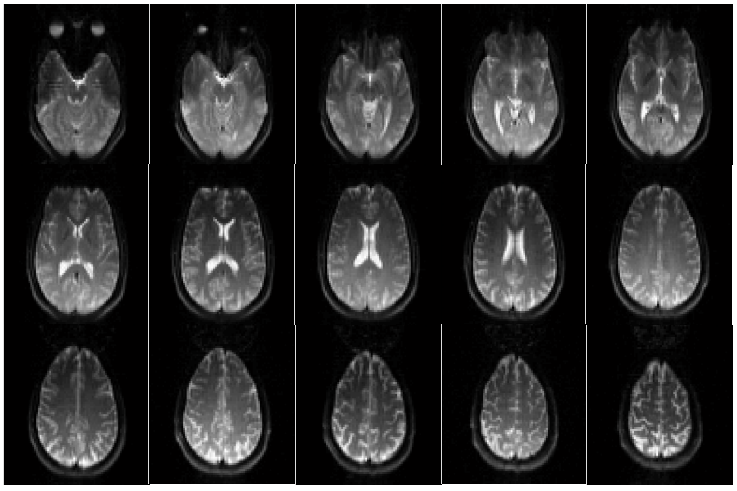



**Fig.2 – Brain volunteer imaging with 2D hybrid SPEN excitation – 15 slices (5mm thickness)**



**Methods.** A 2D pulse comprising *sinc* sub-pulses in the "fast" (slice-selective) dimension and a discrete parabolic phase enveloped by a *WURST* amplitude modulation for the SPEN –or low bandwidth dimension– were synthesized. This spin excitation 90° pulse was combined with a set of suitable oscillating gradients (see Figure 1). Discrete steps of a quadratic phase $\varphi_{RF} = ay^2 + by + c$ (with a,b,c defined by $G_e$, $T_e$ and *FOV*: the gradient magnitude and duration in the SPEN direction and the excited field-of-view) were added to each of the sub-pulses to generate an effective discrete chirp pulse, with a bandwidth of $BW = \gamma \tilde{G}_e FOV$ ($\tilde{G}_e$ being the mean gradient applied in the "slow" SPEN dimension) [5]. In such implementation excitation replicas will be generated with a parabolic phase similar to the main excited area, but shifted in space. The ratio between the bandwidth-time (Q) product and the number of sub-pulses ($N_{sub\text{-}pulses}$) will define the overlap of these excitation replicas with the main excited region [3]: if $Q=N_{sub\text{-}pulses}$ the replicas will be adjacent to the excited FOV, but if $Q>N_{sub\text{-}pulses}$ they will partially overlap the latter. For spatial encoding we are interested in high Q values, since this improves the level of immunity to magnetic field inhomogeneity. To increase Q, the RF was applied over both "even" and "odd" sub-pulse lobes (i.e., during positive and negative gradients $\pm G_{ss}$). In order to use these even/odd sub-pulse alternations while avoiding ghosts, a phase correction of the even sub-pulses (dependent on slice position) had to be used to account for imperfect delays between even/odd sub-pulses; this correction was found using a preliminary scan. The parameters of the SPEN acquisition are then defined in the same way as in a continuous 1D 90° pulse excitation [7], including the accounting for the $\int_0^{T_e} G_e(t)dt = \int_0^{T_a} G_a(t)dt$ and for the fully refocusing $T_a=T_e$ conditions. To verify the correctness of the slice profile excitation, the same scan was performed with the "fast" dimension gradient applied in the readout (in-plane) dimension. Preliminary reconstruction in acquisitions with $Q>N_{sub\text{-}pulses}$ values were performed using Super Resolution reconstruction combined with sensitivity maps information similar to shown in [10]. Finally, phantom and brain volunteer imaging experiments were performed using these new 2D pulses on 3 T Siemens TIM TRIO clinical platform using a 4-channels brain coil.

**Results.** The right-hand panels in Figure 1 illustrate, with a sample slice extracted from out of 15 slices used to image a phantom, the characteristics of the 2D SPEN RF excitation profile. This profile was verified to supply the requested slice thickness, and a quadratic phase yielding good 2D Hybrid SPEN images. Figure 2 shows another multi-slice application of these pulses, with human brain SPEN imaging results obtained with the new sequence (Q=120, $N_{sub\text{-}pulses}$=120, $T_e$=$T_a$=56 ms, 80x80 acquired pixels), showing well-resolved slices and no significant distortions. The scan was compared to EPI with the same acquisition parameters (not shown) and showed the expected increased robustness at no cost in performance. The SAR of the ensuing sequence was ca. 3x higher than that of its EPI counterpart, but ca. 1/3 the magnitude of a comparable multislice SPEN acquisition based on 180° encoding pulses [9].

**Conclusions.** A 2D RF excitation pulse based on SPEN shows promising results towards facilitating multi-slice SPEN imaging with reduced SAR. The hybrid SPEN pulse has reduced sensitivity vis-à-vis $B_0$ inhomogeneities, and hence is not subject to limitations we have encountered using comparable 2D RF pulses based on *k*-space encoding concepts [8]. The ensuing approach should proof beneficial in functional, diffusional and other ultrafast MRI applications. Using parallel transmit methods the robustness of these novel 2D pulses should be further improved.

**Acknowledgments.** We are grateful to Mr. A. Seginer for helpful discussions and assistance in the even/odd sub-pulses calibration procedure. Additional thanks to Dr. E. Haran and the Weizmann MRI technician team, and to Dr. S. Shushan (Wolfson Medical Center) for assistance in the human imaging scans. **Financial support:** ERC Advanced Grant #246754, a Helen Kimmel Award for Innovative Investigation, Kamin-Yeda Grant #711237 (Israel).

**References.** [1] Meyer C.H., et al. Magn.Reson.Med.1990;15:287-304, [2] Weber-Fahr W., et al. Magn.Reson.Imag. 2009;27:664-671 [3] Dumez J-N., et al. J. Magn. Reson. 2013;226:22-34. [4] Shulte R., et al., J. Magn. Reson. 2013; 235: 115–120. [5] Goerke U., et al. Neuroimage. 2011; 54(1):350-60. [6] Solomon E., et al. .Magn. Reson. 2013; 232:76-86 [7] Schmidt R., et al., Proc. Intl. Soc. Mag. Reson. Med. 2013;21:0654. [8] Ben-Eliezer et al., NMR Biomed. 2011; 24: 1191–1201. [9] Schmidt R, et al. Magn. Reson. Med. 2013;doi: 10.1002/mrm.24714. [10] Schmidt R, et al. Magn. Reson. Imag. 2013; doi:10.1016/j.mri.2013.07.007.