Variable Density Single-Shot Fast Spin Echo with Auto-Calibrated Wave Encoding

Feiyu Chen[1], Valentina Taviani[2], Joseph Y. Cheng[3], Tao Zhang[4], Brian A. Hargreaves[3], John M. Pauly[1], and Shreyas S. Vasanawala[3]

[1]*Electrical Engineering, Stanford University, Stanford, CA, United States*

[2]*Global MR Applications and Workflow, GE Healthcare, Menlo Park, CA, United States*

[3]*Radiology, Stanford University, Stanford, CA, United States*

[4]*Global MR Applications and Workflow, GE Healthcare, Houston, TX, United States*

Synopsis

Wave encoding was implemented in a variable-density single-shot fast spin echo (VD-SSFSE) pulse sequence. Auto-calibrated estimation of the wave-encoding point-spread function (PSF) and coil sensitivity maps was used. Images were reconstructed with parallel imaging and compressed sensing reconstruction. Compared to non-wave-encoded Cartesian imaging, wave-encoded VD-SSFSE achieves improved image quality with reduced aliasing artifacts at higher acceleration factors and with full k-space coverage, providing fast acquisitions and clinically relevant echo times.

Purpose

Single-shot fast spin echo (SSFSE) can provide excellent T2 contrast and good motion-robustness for abdominal and pelvic MRI. Variable density (VD) sampling and variable refocusing flip (VRF) angles along the echo train[1] have been shown to reduce scan time, enable compressed sensing reconstruction, and allow full k-space coverage with clinically relevant echo times. However, the maximum acceleration factor of variable density sampling is usually limited by the number of available coil elements in the phase-encoding (PE) direction, and therefore, residual aliasing and noise amplification can be observed in highly accelerated VD-SSFSE scans. In this work, we combine wave encoding[2] with VD-SSFSE to enable higher acceleration factors. By applying sinusoidal modulation in the PE direction during readout, the proposed wave-encoded VD-SSFSE can reduce noise amplification and residual aliasing artifacts without decreasing the scanning efficiency of conventional VD-SSFSE scans.

Methods

A sinusoidal wave-encoding gradient (5.5 cycles, 12 mT/m amplitude), played out during the readout of each k_x encoding line, was added to a VD-SSFSE pulse sequence (Fig. 1a). VD under-sampling patterns with 20 central PE lines of auto-calibration signals (ACS), 50-70 k_y views, and an effective acceleration factor of about 5 were used (Fig. 1b). Over-sampling of 1.6-2.0 in the frequency-encoding (FE) direction was used to account for voxel spreading effects due to wave encoding[2]. The VRF schedule was controlled by prescribing first, minimum, and last flip angles as well as the flip angle corresponding to the center of k-space. A 90° minimum flip angle was used to minimize signal loss due to cardiac pulsation in the left lobe of the liver.

The under-sampled wave-encoded k-space was used to estimate the wave-PSF[3]. Auto-calibrated estimation of coil sensitivity maps[4] (ESPIRiT[5]), and CS-SENSE image reconstruction[6] with L1-

wavelet regularization were performed in MATLAB and C (the BART toolbox[7]). Simulated g-factor maps of wave-encoded and non-wave-encoded acquisitions with uniform under-sampling were compared using estimated coil sensitivity maps of a 32-channel torso coil (NeoCoil, Pewaukee, WI) for uniform sampling patterns.

Phantom and volunteer scans were performed with Institutional Review Board approval and informed consent at 3T (GE MR750, Waukesha, WI) using a 32-channel receive-only torso coil (NeoCoil, Pewaukee, WI) with FE along R/L and PE along A/P. Conventional Cartesian acquisitions were performed for comparison using the same sampling pattern and reconstruction framework.

Results

With an acceleration factor of 4 (Fig. 2a), wave encoding achieved an average g-factor of 1.2 and a peak g-factor of 1.5, while the corresponding Cartesian acquisition resulted in a g-factor of 1.4 and a peak g-factor of 2.5. Wave encoding also reduced the average g-factor from 1.9 to 1.5 and reduced the peak g-factor from 3.6 to 2.1 when a 5x acceleration was used (Fig. 2b). Phantom (Fig. 3) and in-vivo images (Fig. 4 and 5) of wave-encoded acquisitions showed reduced residual aliasing artifacts, less blurring, and more structural details than conventional Cartesian acquisitions. Similar T2 contrast was observed in Cartesian and wave-encoded images. The reconstruction time per slice was 10s for Cartesian and 20s for wave encoding, and the calibration of wave-PSF took about 60 s prior to reconstruction.

Discussion

Wave encoding creates voxel spreading in the FE direction[2]. Therefore, it makes use of coil sensitivities more efficiently and achieves lower g-factors than Cartesian acquisitions. When incorporated in VD-SSFSE readouts, wave encoding significantly improves the image quality at an effective acceleration factors of about 5. The readout duration, echo-spacing, and TE of wave-encoded acquisitions are within 3 ms of those of conventional Cartesian acquisitions, which helps maintain T2 contrast. Auto-calibration of the wave-PSF and the coil sensitivity maps using the under-sampled wave-encoded k-space is intrinsically more robust to motion, which is critical for body applications, and reduces the complexity of translating the proposed approach to clinical practice.

Conclusion

Compared to conventional non-wave-encoded Cartesian imaging, the proposed wave-encoded sampling achieves improved image quality with reduced aliasing artifacts at higher acceleration factors and with full k-space coverage, thus providing fast acquisitions and clinically relevant echo times. Combined with variable-density sampling and compressed sensing, wave-encoded sampling enables acceleration factors of close to 5 for 2D SSFSE scans.

Acknowledgements

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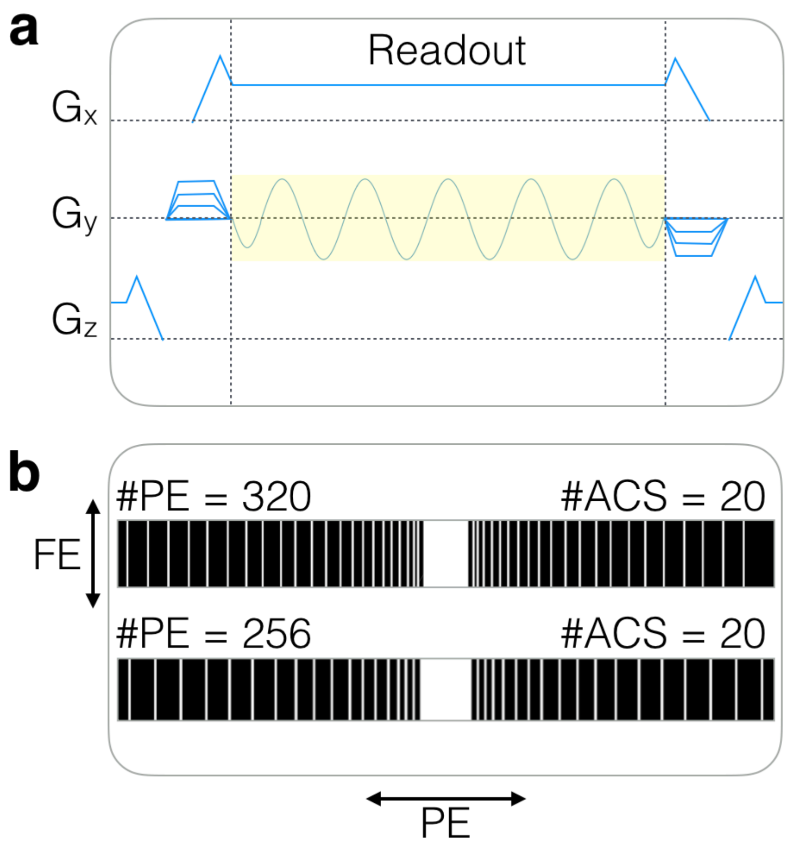

Fig. 1 (a) Illustration of the wave-encoding gradient waveform for VD-SSFSE. The area of the first and the last sinusoidal lobes is reduced by 50% to center the trajectory on the current phase-encode (PE) line. (b) The two variable-density sampling patterns used consisted of 320 and 256 PE lines. The number of effective PE lines were 65 for the first pattern and 54 for the second pattern. 20 central phase-encodes were used for parallel imaging calibration.

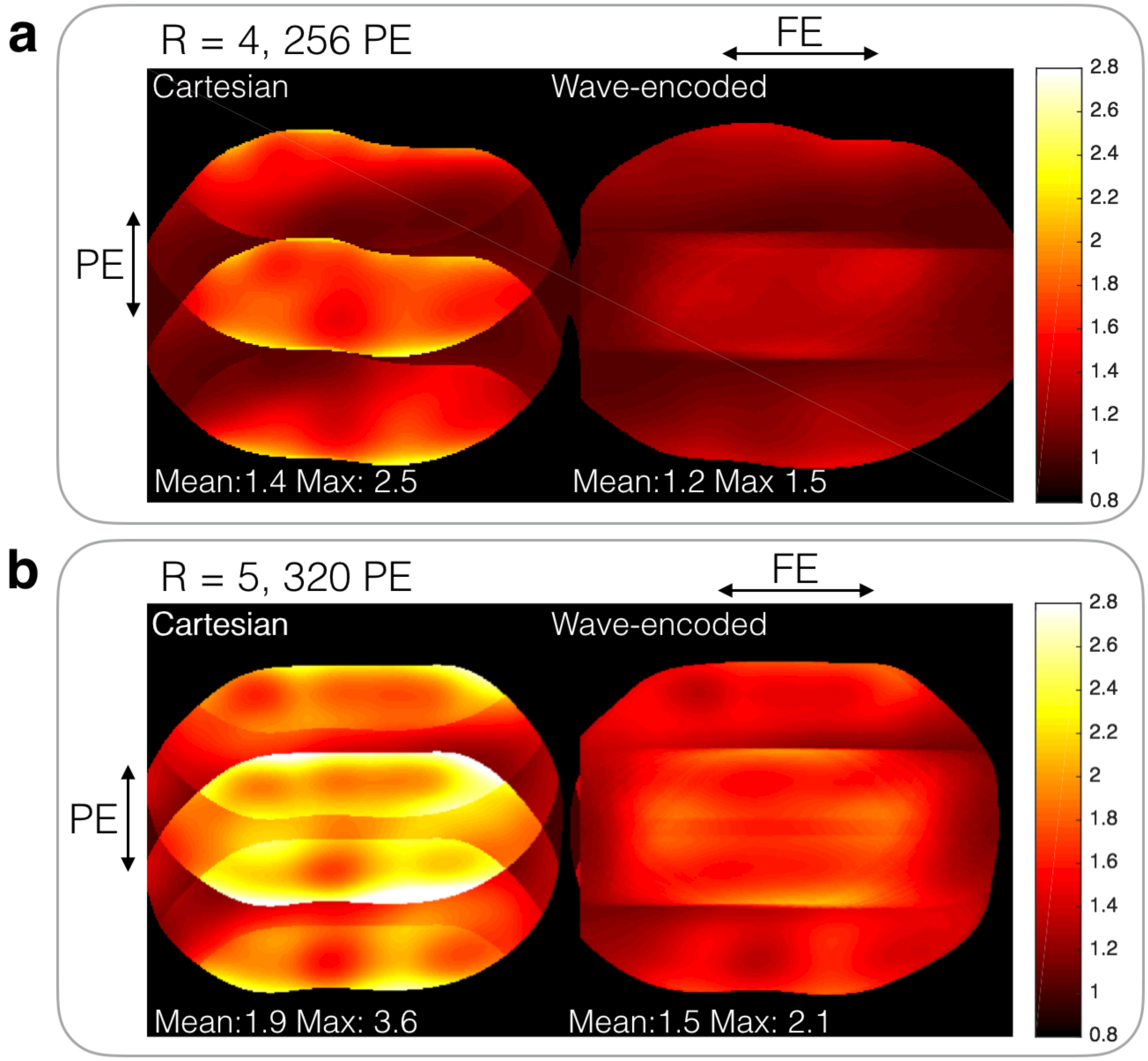

Fig. 2 Comparison of g-factor maps between uniformly under-sampled 2D Cartesian and 2D wave-encoded acquisitions with (a) an acceleration factor of 4 and 256 phase-encodes, and (b) an acceleration factor of 5 and 320 phase-encodes.

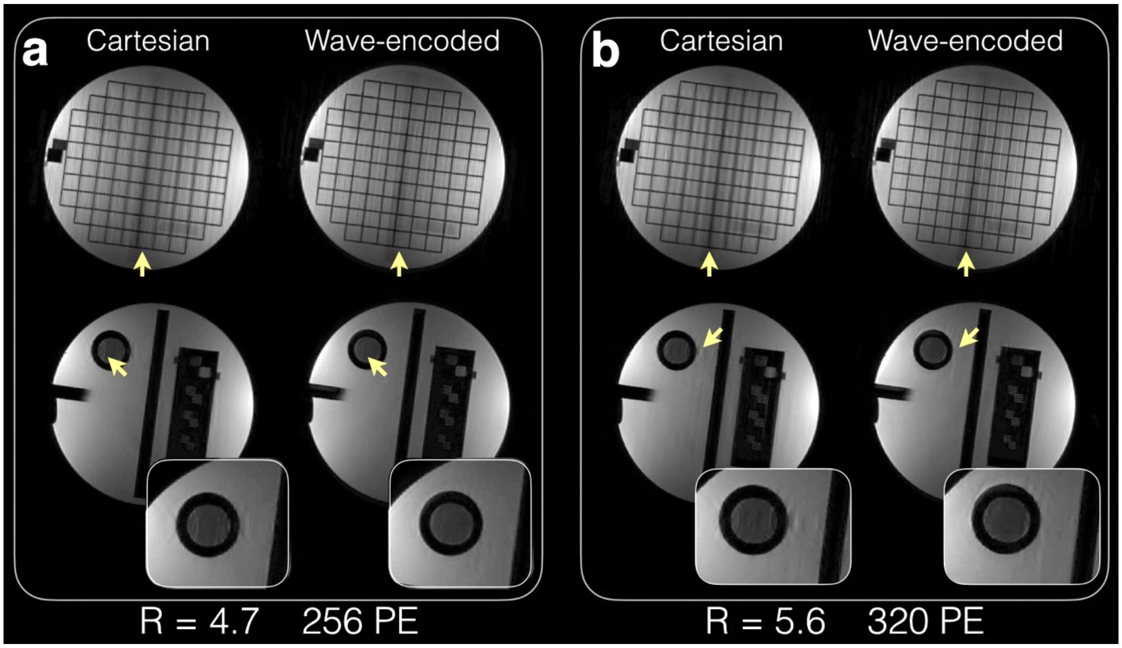

Fig. 3 Comparison of reconstructed phantom images between variable-density under-sampled Cartesian and wave-encoded acquisitions with (a) an acceleration factor of 4.7, 256 phase-encodes, and 20 ACS lines, and (b) an acceleration factor of 5.6, 320 phase-encodes, and 20 ACS lines. Arrows highlight artifacts or increased noise in Cartesian that are reduced with the wave-encoding method.

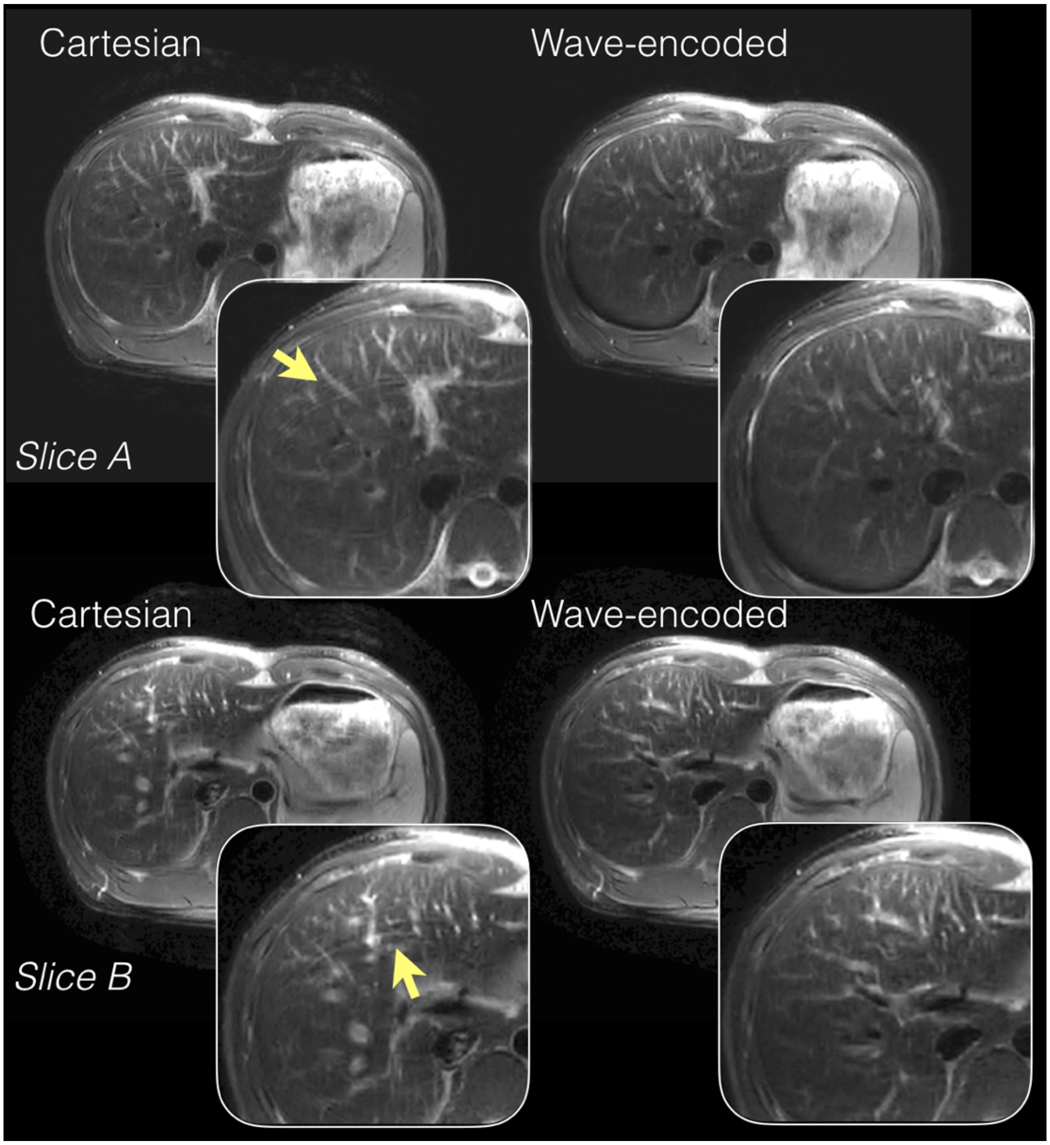

Fig. 4 Volunteer scans with an effective acceleration factor of 4.7 and 20 ACS lines. The scan parameters for both Cartesian and wave-encoded acquisitions were: TE 94ms, TR 479ms, BW +/- 244Hz/pixel, #FE 256, #PE 256, slice thickness 8mm, NEX 1, echo-train length 54, echo-spacing 4.2ms, and FOV 34cm×34cm. An over-sampling factor of 2 in the FE direction was used for the wave-encoded scans. Note how the residual high-frequency artifacts typical of VD (arrows) were resolved with wave encoding.

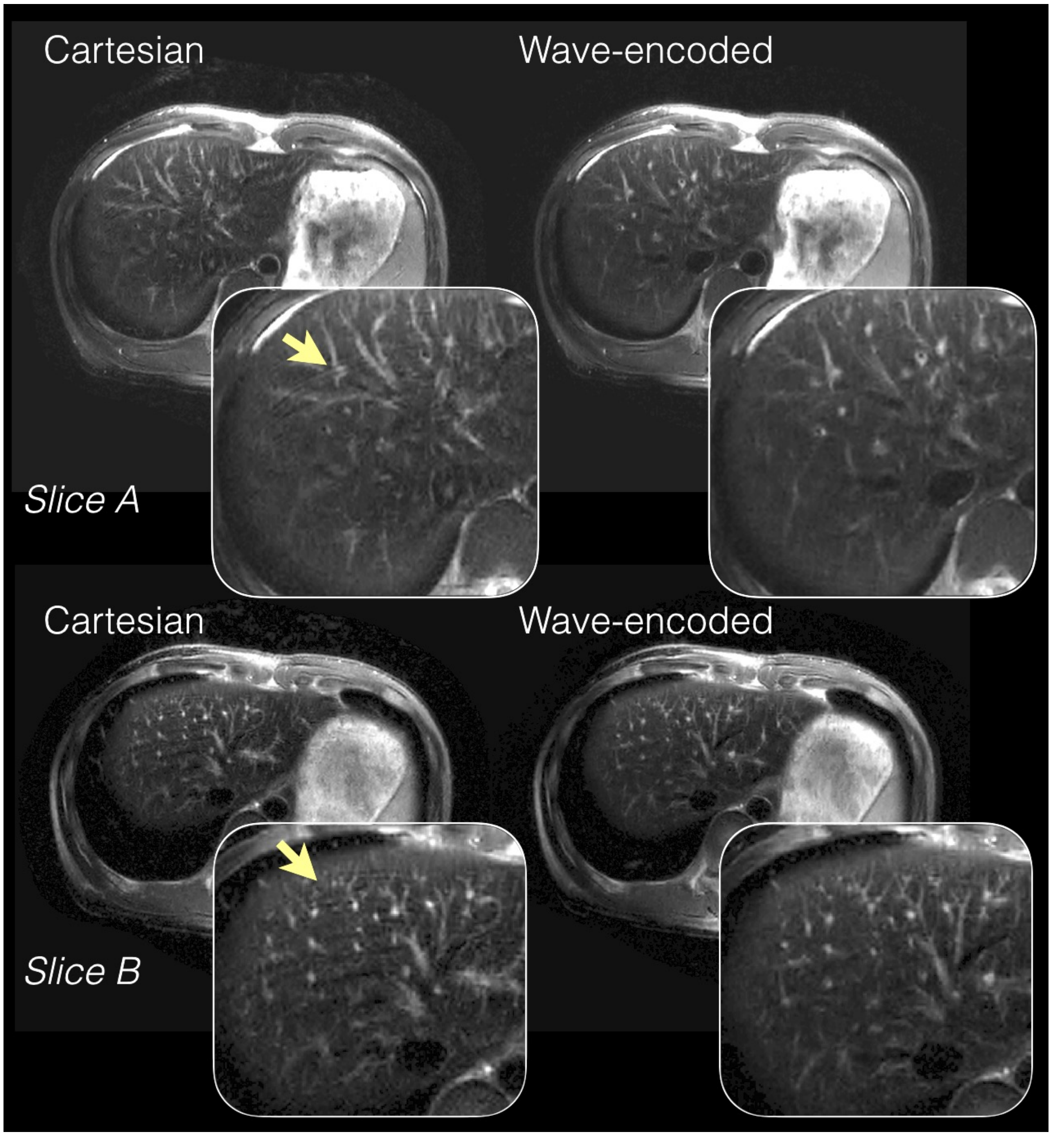

Fig. 5 Volunteer scans with an effective acceleration factor of 4.9 and 20 ACS lines. The scan parameters were: TE 104ms for Cartesian acquisitions and 107ms for wave-encoded acquisitions, TR 569ms, BW +/-260Hz/pixel for Cartesian acquisitions and +/-244Hz/pixel for wave-encoded acquisitions, #FE 320, #PE 320, slice thickness 8mm, NEX 1, echo-train length 65, echo spacing 4.1ms for Cartesian acquisitions and 4.3ms for wave-encoded acquisitions, and FOV 34cm×34cm. An over-sampling factor of 1.6 was used in the FE direction for the wave-encoded scans. Note how the residual high-frequency artifacts typical of VD (arrows) were resolved with wave encoding.