Comparison of Wideband Steady-State Free Precession and $T_2$-Weighted Fast Spin Echo in Spine Disorder Assessment at 1.5 and 3 T

Giovanna S. Danagoulian, Lei Qin, Krishna S. Nayak, Rivka R. Colen, Srinivasan Mukundan, Jr., Mitchell B. Harris, Ferenc A. Jolesz, Ajit Shankaranarayanan, William A. Copen, and Ehud J. Schmidt

INTRODUCTION

In spinal magnetic resonance imaging, degenerative changes are often seen in asymptomatic patients, making a diagnosis of clinically significant nerve root compression or spinal stenosis difficult (1). More than 20% of adults over 60 years of age with no symptoms or functional limitations may have findings of spinal stenosis on imaging studies (2–5). Furthermore, in patients who do have symptoms, localizing the anatomic source is frequently difficult using MRI, especially in cases of radiculopathy of extradural origin, or in patients with multiple level disc protrusions (6–8).

The most common MRI sequences for imaging the spine are sagittal and axial 2D $T_1$-weighted and $T_2$-weighted images. In $T_2$-weighted images, the cerebrospinal fluid (CSF) appears bright, and intradural nerves are seen as darker structures. For $T_2$-weighted imaging, fast spin echo (FSE) is usually used in sagittal scans, whereas for axial scans, 2D FSE or gradient echo are used. The recently introduced multiple echo recombined gradient echo (MERGE and MEDIC) sequence, which utilizes multiple gradient echoes, provides improved $T_2$-weighted contrast between nerve roots and foraminal stenosis (9,10). In addition, 3D FSE (SPACE, CUBE, and VISTA) is used to image the postganglionic nerve roots at high resolution (9,11), although its long acquisition times (8–10 min), make the sequence less suitable for spinal cord imaging and traumatic injury, such as nerve root avulsion (12).

Fat-suppressed diffusion-weighted unbalanced 3D SSFP was proposed (13) for imaging the small peripheral nerves. It provided isotropic spatial resolution, but due to a low signal-to-noise ratio (SNR), required a long (11 min) acquisition time. Diffusion tensor imaging with echo planar readouts, have also been evaluated for the assessment of tumoral pathologies (14). Diffusion tensor imaging tractography has delineated anatomical changes, translations, and deformations of fibers around tumors in the cord and the separation between the neoplastic tissue and nerve roots (14).

Balanced steady-state free precession (balanced-SSFP) was used for spinal imaging in the 1980s and 1990s but was thereafter rejected in favor of FSE, due to severe susceptibility artifacts at the bone-tissue and bone-fluid interfaces in the spine, which restricted its use to low spatial–resolution applications (primarily rapid localizers).
Wideband steady-state free precession (WB-SSFP) was developed (15) to reduce susceptibility artifacts found in balanced-SSFP. WB-SSFP increased distances between nulls in the balanced-SSFP frequency profile and enabled high-resolution 3-T cardiac ventricular function and coronary artery imaging (16–18). WB-SSFP possesses a $T_2/T_1$-weighted contrast similar to that found in balanced-SSFP, albeit with a slight reduction in SNR (16).

We hypothesized that WB-SSFP’s expanded possibilities might allow for higher spatial-resolution spine imaging and potentially reduce the large radio frequency energy deposition, which currently limits FSE slice coverage at 3 T.

We optimized WB-SSFP spatial resolution and contrast-to-noise for cervical and lumbar spine imaging at both 3 and 1.5 T. A key goal was delineating the nerve roots outside the dural sac, as they pass through bone marrow, fat, and muscle, which is a challenge with FSE.

The WB-SSFP sequence was compared with the clinical spine $T_2$-weighted FSE sequences used at Brigham and Women’s Hospital and Massachusetts General Hospital. The sequences were compared at two field strengths in normal subjects and in patients with various spinal disorders. Image SNR and contrast-to-noise ratio (CNR) between several spinal tissues were measured. Neuroradiologists at Brigham and Women’s Hospital and Massachusetts General Hospital qualitatively assessed WB-SSFP’s ability to separate pathology from normal anatomy. In this publication, a limited number of pathological cases are shown, demonstrating WB-SSFP advantages and disadvantages.

METHODS

Subjects

All subjects were studied with institutional review board approval at both Massachusetts General Hospital and Brigham and Women’s Hospital and provided informed consent. Three Tesla studies were performed on a GE (Waukesha, WI) 3 T Twin-speed, and 1.5 T studies were performed on a GE 1.5 T Twin-speed, utilizing software version 15X (HDx) and using the Head-Neck or the CTL-Spine receiver-coil arrays.

Normal Subjects

For the purpose of evaluating SNR and CNR, we scanned six healthy subjects, without spinal symptoms, with a median age of 38-years old.

Patients

Ten patients with degenerative or traumatic spine disorders were imaged in the Massachusetts General Hospital Emergency Room 1.5 T MRI, whereas 50 degenerative spine patients were imaged on two Brigham and Women’s Hospital 3 T MRIs. Selected patient cases are used to demonstrate advantages and disadvantages of WB-SSFP, while a quantitative analysis of the complete data will follow in a subsequent report.

The Wideband Steady-State Free Precession Sequence

The balanced-SSFP sequence (True-FISP, FIESTA, and balanced-FFE) reaches a steady state in both the longitudinal and transverse magnetization after multiple repetition times (TRs) of the basic waveform are executed (19,20). Balanced-SSFP differs from spoiled gradient-echo pulse sequences, wherein the contribution of transverse magnetization from a given RF pulse to the signal measured in subsequent repetitions is reduced or suppressed (21) as compared to the balanced-SSFP approach. In balanced-SSFP, all gradient directions possess a zero net area over the course of a TR, and frequently, also a zero-gradient first moment, which reduces flow sensitivity. As compared to a spoiled gradient-echo sequence, balanced-SSFP provides a higher SNR efficiency. Balanced-SSFP has a $T_2/T_1$ contrast (19,20), which differs from the contrast provided by $T_2$-weighted FSE. Balanced-SSFP sequences are characterized by a useable band of frequencies (the passband), whose width $D_\delta$ is the frequency difference in hertz (Hz) between two successive frequencies of maximal signal interference, where the signal amplitude is minimal.

WB-SSFP uses alternating TR lengths to widen the passband of SSFP and minimize undesired signal nulling, as illustrated in Fig. 1 (Matlab code from http://mrel.usc.edu/share.html was used for this simulation). The two alternating TRs, the short TR (TRs) and the long TR (TRl), delivered with an alternating RF phase, produce an oscillating steady state (15). The ratio of the short and long TRs, $a = TRs/TRl$, determines the extent of the broadening of the SSFP passband, as at low flip angles (15)
\[
\Delta f = 2/[TR(1 + a)]
\]  

providing the contrast and resolution advantages of WB-SSFP, as shown in Fig. 1, where four values of \(a\) (1, 0.75, 0.58, and 0.41) are displayed. When \(a = 1\), WB-SSFP becomes balanced-SSFP (TR\(_L\) = TR), while when \(a = 0\), TR\(_L\) is twice the length of the maximal TR possible with balanced-SSFP. The intensity of the steady-state signal is a function of \(a\), and therefore, the SNR and the CNR, as a function of flip angle (\(\alpha\)), vary with \(a\) (15). There is a loss of \(\sim30\%\) in the passband amplitude as \(a\) is reduced. WB-SSFP is preferably used with \(a < 1\), as then it is possible to substantially prolong TR\(_L\), which, in turn, allows performing scans with narrower receiver bandwidths, which recovers the WB-SSFP SNR loss, and is required for obtaining sufficient SNR and the required increased readout time for high spatial resolution scans. The advantages of WB-SSFP increase at higher field, as off-resonance effects due to magnetic susceptibility scale linearly with field strength, restricting the usable TR of balanced-SSFP to yet shorter values (e.g., 3.5 ms at 3 T).

The WB-SSFP fat signal in the presence of susceptibility is problematic, because it is shifted by \(-220\) (1.5 T) or \(-440\) Hz (3 T) relative to water. As a result, spinal fat experiences the SSFP-side-lobes (Fig. 1) and may experience amplitude fluctuations, especially off center, where the spectra can be shifted further from the central frequency due to local field gradients.

The magnitude of the WB-SSFP signal at the passband center depends on the flip angle \(\alpha\) according to the Eq. 16

\[
M_{SSFP}(\alpha, \beta) = [(a + \cos \alpha)\beta + (1 - a \cos \alpha)\sin \alpha] / [\sin^2 \alpha(1 - \beta)^2 + \beta(a^{0.5} + a^{-0.5})^2]
\]  

where \(\beta = (T_1/T_2)\). For spinal tissues, such as muscle \(\beta(1.5, 3\) T) = 32, 45 and nerves \(\beta(1.5, 3\) T) = 10, 12, the magnetization peaks at \(\alpha = 20\)–25 and \(30\)–\(35\)°, respectively, for \(a = 0.3\)–1.0. The magnetization transfer effect in balanced-SSFP (22) can be quite strong. It can contribute to a large attenuation of the muscle signal, an effect that grows with \(a\) and decreases with TR\(_L\). This effect further increases the desirability of imaging on the lower bounds of the flip-angle ranges determined from Eq. 2, because at higher flip angles, the skeletal muscle will appear darker.

When there is a mechanism that affects the transverse magnetization phase over a TR, for example, gradients induced by susceptibility differences (23), the longer TR\(_L\) of WB-SSFP (>6 ms), relative to balanced-SSFP, is expected to produce a larger phase accumulation over TR. Flow-induced phase accumulation occurs in balanced-SSFP and WB-SSFP (24), even when the pulse sequences are balanced to first order, as flow moves spins between regions with differing magnetic fields, corresponding to different points in the balanced-SSFP frequency response. In the case of pulsatile flow, where the flow velocity is time varying, each TR\(_L\) during image acquisition sees a different phase accumulation (24) so that a successive TR cannot completely compensate for the phase dispersal of its predecessor, which can lead to a net loss in the signal intensity (SI) established over multiple TRs. As a result, when an image is acquired, we expect that vessels with pulsatile flow will appear darker than those with constant (or no) flow, which may lead to the appearance of dark blood vessels (e.g., a “black-blood” effect) in long-TR balanced-SSFP and in WB-SSFP.

The MRI Spine Imaging Protocol

The imaging protocol utilized for all subjects was based on the standard clinical lumbar spine protocol in use at both institutions, which included T\(_2\)-weighted FSE, supplemented by WB-SSFP. To reduce 3-T magnetic susceptibility artifacts, second-order magnetic-field shimming was performed before WB-SSFP, using a 240-mm field-of-view (FOV) and the system’s “Zoom” gradients (peak: 50 mT m\(^{-1}\), slew rate: 150 T m\(^{-1}\) s\(^{-1}\)).

WB-SSFP parameters were optimized for cervical and lumbar spine imaging at both 3 and 1.5 T, to provide higher spatial resolution and contrast-to-noise resolution than FSE. A key goal was delineating the nerve roots outside the dural sac, as they pass through bone marrow, fat, and muscle, which is a challenge with FSE.

Three-dimensional WB-SSFP’s parameters include (3 T) TR\(_L\)/TR\(_R\)/TE/\(\alpha\) = 5.1–6.0/2.5/2.1 ms/25–30°, 320 × 256, FOV = 180–200 mm, 0.56 × 0.69 mm\(^2\)–0.63 × 0.89 mm\(^2\) in-plane resolution, slice thickness 1.6–2.0 mm, interpolated to 0.8–1.0 mm, 1 average; (1.5 T) TR\(_L\)/TR\(_R\)/TE/\(\alpha\) = 6.9–7.1/2.5/4.2 ms/30°, 256 × 256, FOV = 180–200 mm, 0.7 × 0.7 mm\(^2\) in-plane resolution, slice thickness 2.6 mm, interpolated to 1.3 mm, 1 average. Edge slices (5%) were removed during reconstruction.

WB-SSFP receiver bandwidths were ±41.67 kHz, far narrower than the ±125 kHz ordinarily used with balanced-SSFP, allowing extension of the long TR (TR\(_L\)) to 6 or 7.2 ms (3 and 1.5 T), from the maximum TR possible with balanced-SSFP, 4 or 5 ms (3 and 1.5 T). Acquisition was preceded by 10 s of unacquired repetitions to insure achievement of magnetization steady state. The WB-SSFP flip angles (25–30°) were chosen to achieve the highest SNR without the muscle tissue appearing very dark. The slab-select gradients were required to be >0.15 G/cm, preventing slice folding due to local-field distortions, but limiting the slab thickness to 8 cm. Scans were performed in sagittal, coronal, or axial slice orientations, with single-scan full-spine coverage possible for the coronal or sagittal scans.

Two-dimensional FSE parameters include (3 T) TR/TE 1767–2317/110–126 ms, echo train length 16, 416 × 224, FOV = 200 mm, slice width 2 or 4 mm, bandwidth = ±32 kHz, 2 averages; (1.5 T) TR/TE 2117–2317/110–126 ms, echo train length 16, 288 × 224, FOV = 230 mm, slice width 4 mm, bandwidth ±32 kHz, 2 averages.

Data Analysis

Image reconstruction utilized the sum-of-squares approach for adding the eight coil-array elements, and no image-shading correction was performed.

SNR was defined as the ratio between the SI and the noise, SI/\(\sigma\), where SI is the mean of a distribution of
pixel intensities in a region-of-interest belonging to a given tissue, and \( \sigma \) is the standard deviation of the distribution of pixel intensities in a region in the background. Tissue regions-of-interest were chosen to be close to the FOV center, so as to reduce flip-angle and field in-homogeneity issues, with their size (Table 1), carefully selected to prevent partial volume effects. Such an SNR measurement for an array coil is not strictly valid but is used here only for intersequence comparison.

The SNR efficiency \( \mu \), the SNR per unit time and unit volume, was defined as the ratio between the SNR and the square root of the scan duration time (SNR/\( \sqrt{t} \)), normalized to the voxel volume.

\[
\mu = \frac{\text{SNR}}{(s^{0.5} \text{mm}^3)}
\]

The CNR was defined as the difference in SNR efficiency between two adjacent tissues, A and B.

\[
\text{CNR}_{A:B} = \mu_A - \mu_B
\]

A larger CNR allowed better intertissue differentiation. We measured SNR and calculated the CNR between four pairs of neighboring tissues (Fig. 3b): nerve-ganglia:fat, nerve-roots:muscle, intradural-nerves:CSF, and discs:fat.

### Image Postprocessing

The GE advantage Windows workstation’s reformatting function was used to reformat 3D WB-SSFP images into orthogonal planes.

### RESULTS

#### Quantitative Comparison of WB-SSFP and FSE

Table 1 shows the SNR efficiency \( \mu \), which were measured for various spinal tissues in six normal subjects (Table 1). At 3 T, \( \mu \) of most spinal tissues is two to four times that of FSE, excluding the nerve ganglia, which has 1.5 times the \( \mu \) of FSE. At 1.5 T, the WB-SSFP \( \mu \) of most tissues is two to three times that of FSE, excluding disk and nerve ganglia, where the signals are essentially equivalent.

Figure 2 is a comparison of intertissue CNR at both 3 and 1.5 T. At 3 T, WB-SSFP CNR was \(~3.7–5.2\) times that of FSE, excluding CNR_{ganglia:fat}, which was approximately twice as great for WB-SSFP relative to FSE. At 1.5 T, the WB-SSFP CNR was \(~3–3.5\) that of FSE, excluding CNR_{nerve:CSF}, where the WB-SSFP was \(~1.7\) times that of FSE. The observed magnetic-field differences in CNR
are primarily due to the large changes in tissue $T_1$ values that occur between the two field strengths (25,26).

Table 2 summarizes the scan time and SAR comparisons. At 1.5 T, achieving equivalent spatial resolutions with both WB-SSFP and FSE is impossible, a result of the lower CNR of FSE, so FSE acquisitions acquired at similar scan times to WB-SSFP are of far lower resolution. WB-SSFP’s average SAR is also smaller, a result of use of lower flip angles. At 3 T, 3D WB-SSFP requires slightly longer acquisitions than 2D FSE, due primarily to the use of thinner WB-SSFP slices at 3 T, with more slices required to cover the entire spine. The WB-SSFP SAR is far lower than that of FSE, a result of use of lower flip angles.

Figure 3 compares FSE and WB-SSFP acquired at identical in-plane resolution and slice-width. The CSF signal is brighter and less uniform on FSE, a result of the dephasing of pulsatile signal in WB-SSFP (24). Muscle signal is also lower in WB-SSFP, which may be a result of the magnetization transfer effect (22), which is larger in WB-SSFP (27), relative to balanced-SSFP. Intradural nerves are seen equally well with FSE and WB-SSFP, but WB-SSFP is superior in delineating extradural structures such as the nerve ganglia and nerve roots, which can be tracked with WB-SSFP to distances of 15 cm from the dural sac, a result of the large $\text{CNR}_{\text{nerve:fat}}$ or $\text{CNR}_{\text{nerve:muscle}}$.

Clinical Examples of WB-SSFP

Figure 4 shows 3 T axial images of a patient with left-sided pain imaged at 3 T with both FSE and WB-SSFP. There is better conspicuity of the nerve roots in the WB-SSFP images, especially seen in the sharper depiction of intraforaminal segments. Inferior to the discs, the WB-SSFP images are more motion blurred than the 2D-FSE images.

Figure 5 shows 3-T coronal WB-SSFP images that depict impingent upon the left-sided S1 nerve. The difference in the sizes of the nerve ganglia can also be assessed by comparison with the contralateral nerves. The images also demonstrate the wealth of bone-marrow contrast in WB-SSFP, which is sensitive to the water and fat content in a matter similar to other gradient echo sequences. Despite our fear that the fat signal would spatially vary, the large FOV images demonstrate a homogeneous fat signal over large regions (the body right–left variation is also seen in FSE, and is due to coil sensitivity variations at 3 T).

Figure 6 demonstrates the high resolution achieved with WB-SSFP in the cervical spine. The coronal acquisition was reformatted into orthogonal planes, and a curved plane reconstruction was then performed by a neuroradiologist following the course of the C2 nerve, which is nicely depicted (Fig. 6d). The dark carotid arteries appearance demonstrates WB-SSFP’s black-blood contrast. Cervical spine imaging was affected by higher magnetic-field inhomogeneity, so the maximum coverage attained without off-resonance artifacts was 80 ± 10 mm in the superior–inferior direction (Fig. 6e,f).

Figure 7 compares FSE and WB-SSFP at 1.5 T in a cervical spine trauma patient. The central diagnostic question was damage to the intradural left C1 nerve roots. WB-SSFP answered this question, while the larger

![FIG. 3. Axial FSE (a) and WB-SSFP (b) of a healthy volunteer at 1.5 T obtained at the same resolution. Fat (plus), disc (ellipse), CSF (star), intradural nerves (circle), and nerve ganglia (minus) used for SNR and CNR calculations are shown. The dorsal root ganglia and cauda equina (arrow heads) are better seen with WB-SSFP, whereas WB-SSFP’s CSF signal is less hyperintense and more spatially uniform than that of FSE.](image)
slice-width FSE image was harder to read. The deformation of the vertebrae posterior to the cord is also nicely depicted on the 3D WB-SSFP image. Since the carotid and vertebral arteries appear dark in WB-SSFP, the images lack pulsation artifacts, substantially different from conventional balanced-SSFP, in which blood vessels appear bright. A WB-SSFP deficiency relative to FSE is FSE’s stronger edema signal (compare Fig. 7b,c).

Figure 8 shows a high-resolution WB-SSFP acquisition in a patient with spinal stenosis. This patient had multilevel disc protrusions, so the coronal acquisition was reformatted into sagittal and axial planes to visualize cord compression at several levels. The severe compression more inferiorly (Fig. 8c) is clearly visible, as is the milder compression presents more superiorly (Fig. 8f). The reformatted images are of diagnostic quality, because of the high CNR and spatial resolutions. The extradural nerves and the intradural roots are also well resolved.

DISCUSSION
WB-SSFP reduces susceptibility artifacts by utilizing alternating long and short TRs. The longer TRL of WB-SSFP permits imaging with lower receiver bandwidth, relative to balanced-SSFP. Depending on tissue relaxation times and other pulse-sequence parameter values, this may improve SNR, which can be traded for increased spatial resolution. We demonstrate that these properties allow for high spatial resolution and distortion-free spinal imaging at both 1.5 and 3 T. Spinal coverage remains somewhat restricted; limited to a superior–inferior coverage of 80 ± 10 mm in the C-spine and 200 ± 20 mm in the L-spine.

WB-SSFP’s black-blood contrast contributes to reducing CSF-pulsation artifacts. The dark-lumen of blood vessels, which differs from the bright lumens found in shorter-TR balanced-SSFP, is particularly useful in the cervical spine, where there are major blood vessels close

FIG. 5. a–f: Three-Tesla coronal WBSSFP images of a patient with left-sided radicular pain. Bone impingement upon left S1 nerve (d–f, orange arrows) and the ensuing deformation of the nerve is clear. The sciatic nerves can be easily followed on both sides (yellow arrowheads).
to the spinal cord. In balanced-SSFP, such vessels frequently contribute to pulsation artifacts that mask spinal anatomy.

The CNR of WB-SSFP at 3 and 1.5 T was higher than that of FSE for most spinal tissues. These CNR differences between WB-SSFP and FSE are larger at 3 T than at 1.5 T, a result of the longer $T_1$ at higher field strength. At 3 T, WB-SSFP images improved visualization of the intraspinal and foraminal nerve root segments, with a CNR three times higher than that of FSE. At 1.5 T, the CNR of WB-SSFP is approximately double that of FSE, excluding the CNR$_{spinal-nerve:CSF}$, which is comparable to that of FSE.

Important clinical deficiencies of WB-SSFP relative to FSE are its lower hyperintense edema signal, and its lower muscle signal. Three-dimensional WB-SSFP currently also has greater motional blurring relative to 2D-FSE in the anterior spine, although we found it to be far less motion sensitive than 3D gradient echo or multiple echo recombined gradient echo. Implementation of

![FIG. 6. Three-Tesla C-spine 3D WB-SSFP. A coronal acquisition (a and b) was reformatted into axial (c). d: Curved-plane reformatting was then performed along white line, tracing the C2 nerve (orange arrows). Dark blood vessels (yellow arrowhead) demonstrate the “black blood” contrast. Images (e) and (f) demonstrate larger coverage, showing spinal vein (e) and brachial plexus (f), although off-resonance artifacts (large white arrows) create darker bands, limiting C-spine superior–inferior coverage to 70–90 mm.]

![FIG. 7. Cervical spine trauma patient scanned at 1.5 T. Sagittal FSE image (a) shows an acute dens fracture (yellow arrowhead). Impingement upon the left C2 nerve roots (yellow arrow) is better seen in the 2.6-mm axial 3D WB-SSFP image (c) than in the 4-mm axial FSE image (b). A fracture of the right posterolateral ring of the C2 vertebra is well demonstrated by all three images, while edema (orange arrow) is better seen with FSE.]

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spatial presaturation or motion compensation may minimize this effect.

Three-dimensional WB-SSFP’s high CNR made it an excellent sequence for detecting anatomic changes, and its higher resolution permitted multiplanar reformating without image degradation. As the intradural nerve roots and spinal nerves are nicely depicted with WB-SSFP, semi-automatic or fully automatic nerve bundle segmentation may be possible, as has recently been demonstrated by our collaborators (28). These nerve maps can be used for preprocedural interventional or surgical planning, and for intraprocedural image guidance. The improved planning and guidance may contribute to reduced damage to the dural sac or to extradural nerves during the procedure, reducing complications, and improving outcomes.

At 3 T, the larger SNR efficiency of WB-SSFP, relative to FSE, can be traded for higher resolution imaging or shorter scan times. As a result, it is possible to perform full-spine (i.e., 18 cm FOV) coverage high-resolution 3D scans with acquisition times of 4–6 min, whereas an equivalent 3D-FSE study (12) might require 8–10 min.

Three-dimensional WB-SSFP, due to the absence of FSE’s 180° and 90° pulses, has a lower average and peak SAR. This difference plays a key role at 3 T, where FSE is limited in slice coverage due to its large SAR. WB-SSFP has a higher dB/dt than FSE, because all gradient waveforms use the maximal scanner slew rate to shorten TR, and the receiver bandwidth is relatively high. However, we did not receive any complaints of neurological stimulation due to the sequence.

There are limitations to our comparisons between FSE and WB-SSFP, since we utilized FSE with standard clinical parameters. Different FSE imaging parameters would have somewhat changed this comparison, especially for long $T_1$ tissues such as CSF, where the FSE SNR efficiency and CNR might have been 20–30% higher than the values we received.

A current WB-SSFP limitation is the shortest achievable TRS (2.5 ms), a result of the RF and gradient waveform requirements. With this TRS, the minimum spatial resolution possible is $0.25 \times 0.25 \times 1$ mm$^3$ and the maximal spinal coverage in a single 3D scan, based on our experience in 60 subjects, free of both off-resonance and slice-folding artifacts, is $180 \times 180 \times 80$ mm$^3$ (lumbar spine) or $180 \times 180 \times 70$ mm$^3$ (cervical spine). With this coverage restriction, complete spinal acquisition is preferably performed in a coronal or sagittal orientation, with axial acquisition requiring multislab acquisition. Use of a shorter TRS, possible with stronger gradients, might have achieved a wider “passband” and a longer TR, providing slightly higher spatial resolution or wider coverage, although there would be an associated SNR loss.

As a supplementary note, WB-SSFP’s utility can be expanded by placing diffusion gradients in the currently unused TRS segment. We demonstrated (29,30), in a small patient cohort, high-resolution diffusion-weighted spinal imaging with this approach, which may also provide for better edema visualization.

In conclusion, 3D WB-SSFP complements FSE in spinal imaging by providing improved CNR for many spinal...
REFERENCES