Removing Silicone Artifacts in Diffusion-Weighted Breast MRI by Means of Shift-Resolved Spatiotemporally Encoding

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Purpose: Evaluate the usefulness of diffusion-weighted spatiotemporally encoded (SPEN) methods to obtain apparent diffusion coefficient (ADC) maps of fibroglandular human breast tissue, in the presence of silicone implants.

Methods: Seven healthy volunteers with breast augmentation were scanned at 3 Tesla (T) using customized SPEN sequences yielding separate silicone and water ¹H images in one scan, together with their corresponding diffusion-weightings.

Results: SPEN's ability to deliver multiple spectrally resolved images in a single scan, coupled to the method's substantial robustness to magnetic field heterogeneities, served to acquire ADC maps that could be freed from contributions that did not belong to fibroglandular tissue.

Conclusion: SPEN-based sequences incorporating spectral discrimination and diffusion-weighting enable the acquisition of reliable ADC maps despite the presence of dominant signals from silicone implants, thereby opening new screening possibilities for the identification of malignancies in breast augmented patients. Magn Reson Med 000:000–000, 2015. © 2015 Wiley Periodicals, Inc.

Key words: diffusion-weighted imaging – DWI; spatiotemporal encoding – SPEN; breast MRI; silicone implants

INTRODUCTION

There is a growing need for imaging breasts with silicone implants, driven by the extended use of augmentation procedures for cosmetic and cancer-related reconstruction purposes (1). A main radiological concern in these analyses resides in evaluating the implant’s integrity (2–5), with MRI being the modality of choice to monitor the potential rupture of an implant (6). Most MRI protocols used for such evaluations rely on multishot T₁ and T₂-weighted techniques (4,5,7), favored due to the detailed anatomical information that they provide about a silicone implant’s integrity. Although silicone implants are not associated with an increased risk of breast cancer, augmented subjects may eventually develop the disease as they share similar lifetime risks as the general population. Even cancer patients that underwent surgery followed by silicone implant reconstruction, exhibit a risk—although greatly reduced—of cancer recurrence (8). Given the concerns that have been raised about the accuracy of mammography as a suitable approach to screen these cases (9,10), there is an increased interest in using MRI for detecting breast cancer in the presence of silicone implants (11). High-risk subjects will be candidates for dynamic contrast enhanced studies, a protocol involving the injection of contrast material that reveals the location of potential cancerous tissue by T₁-weighted MRI scans (12–14). A promising alternative to diagnose breast cancer without injecting external contrasts, is diffusion-weighted MRI (DWI) (15). While still in a process of validation, the main goal of these tests is to monitor water’s microdiffusion by means of the calculation of apparent diffusion coefficient (ADC) maps. Cancerous tissues exhibit lower ADCs than normal fibroglandular tissue or benign lesions (16–18), reflecting the restricted diffusivity brought about by the high cellular density of proliferating cancer regions (14,19). Performing such cancer diagnoses based on diffusion measurements requires a fast technique, exhibiting reduced sensitivity to motional effects. ADC map measurements are thus based on techniques like spin-echo echo planar imaging (SE-EPI), which can provide the desired image with good quality in a single scan (20). Despite its speed and sensitivity SE-EPI is still prone to display artifacts, particularly when applied to the examination of challenging organs like silicone-implanted breasts. Indeed the presence of water-, silicone-, and fat-rich regions in an off-axis organ results in chemical shift artifacts and field heterogeneities, leading to spurious replicas that may confound the radiological interpretation (21–23). Complex suppression schemes of the nontargeted resonances coupled to field-compensating navigator scans and image postprocessing, are thus usually needed to enable a suitable EPI-based diagnosis of silicone-implanted breasts (24).

Over the last years, single-shot imaging alternatives operating on the basis of spatiotemporal encoding (SPEN) methods, have succeeded to reduce the distortions incurred by field inhomogeneities and/or chemical shift artifacts (25–30). These advantages are a consequence of...
SPEN’s non-Fourier nature, enabling its use of stronger average acquisition gradients than those dictated by EPI’s Nyquist criteria. Initial SPEN implementations exhibited SNR limitations, yet the use of super-resolved reconstruction techniques (31–35) have alleviated these. Furthermore, a suitable theoretical framework (36) enables the retrieval of ADC maps from a variety of SPEN implementations. These combined improvements have been recently brought to bear in a series of breast ADC map measurements in both healthy volunteers and cancer patients, that revealed systematically better b-zero images and cleaner ADC maps, than their SE-EPI counterparts (37).

A particular advantage of SPEN rests in its ability to resolve—and thereby suppress without any changes in the basic imaging-oriented pulse sequence—the contributions of different chemical sites (38). This ability to deliver one-dimensional (1D) spectral / 2D spatial correlations in a single scan has been demonstrated in both animal experiments on thermal and hyperpolarized substrates, as well as in human water/fat separations (39,40). The present Note explores yet another possible application of SPEN’s spectroscopic imaging abilities with a SPEN-based study focusing on the acquisition of reliable ADC maps of fibroglandular tissue in silicone-implanted human breasts, whose images have been freed from the dominant silicone signal contributions.

METHODS

Phantoms and Patients

This study was approved by the Institutional Review Board of the Wolfson Medical Center (Holon, Israel), and informed consents were obtained from all participants. Seven healthy volunteers (mean age, 34 years; range, 24–54 years) were recruited for this comparative study, with different sizes of silicone implants and of fibroglandular tissue. Also examined was a round plastic phantom containing tap water, oil, and silicone.

MRI Protocols

Axial images of one or both breasts were acquired at 3 Tesla (T) on a Siemens TrioTIM scanner (Erlangen, Germany) using a four-channel bilateral breast coil (Siemens). Phantom images were collected on the same scanner using a focus-channel head coil. For all seven volunteers the MRI protocol included a $T_2$-weighted turbo spin-echo (TSE) sequence covering the whole breast for anatomical reference. This was followed by DWI acquisitions of fat-suppressed twice-refocused SE-EPI (41) and by SPEN DWI of five central slices in the region of the nipple, with the diffusion-weighting gradients repeatedly applied along three orthogonal directions. Common over all parameters of these experiments included fat suppression, δ = 17 ms pulse-diffusion-weighting gradients, intergradient delays $\Delta = 35$ ms, and $n=7$ nominal $b$ values of 0, 50, 150, 300, 450, 600, and 750 s/mm$^2$. These $b$ values were subsequently corrected by imaging cross-tern gradient effects for the SPEN experiments (36,42) before using them to calculate the actual ADC maps. The spatial resolutions of both SE-EPI and SPEN experiments were $2.0 \times 2.0 \times 2.5$ mm$^3$; FOVs were set to $320 \times 320$ mm$^2$ and $320 \times 160$ mm$^2$, respectively, exploiting in the latter case SPEN’s ability to “zoom” without suffering from folding effects. In both SE-EPI and SPEN experiments the low bandwidth (phase- or SPEN-) directions were scanned in an anterior—posterior geometry. Although not ideally suited for implementing parallel MRI acquisitions, this geometry was chosen over its R–L low-bandwidth counterpart to minimize lateral ghosting artifacts; otherwise, ghost images arising from the strong silicone residuals in EPI acquisitions would often interfere with the retrieval of clear images for the two breasts. The total scan durations (per slice) were 180 ms and 277 ms for DWI SPEN and SE-EPI, respectively. Each scan was preceded by a scanner-provided, selectively fat saturation procedure. These procedures were optimized based on repeated scans on volunteers, which revealed that SPAIR (fat saturation including adiabatic pulses (43), performed best for SE-EPI, while FatSat (37) was the most efficient suppression protocol for SPEN. To shorten the SE-EPI echo times, a partial Fourier k-scan was set to 6/8; even with this provision, Nyquist criteria demanded that the average acquisition gradients along the low-bandwidth dimension of these experiments be ca. six times weaker than their SE-EPI counterparts: 0.01 G/cm for EPI, ~0.06 G/cm for SPEN. The three main frequency peaks present in these experiments (water, fat and silicone) were detected automatically by the scanner, and a frequency adjustment was manually made for both SE-EPI and SPEN whereby the central carrier frequency was positioned on the water peak before each acquisition. Because using the scanner-supplied four-channel bilateral breast coil enables parallel imaging only in the R–L direction, no parallel imaging was used for either method. Minimum echo times for the SE-EPI and SPEN acquisitions were 101 ms and 124 ms, respectively. The sequence bandwidth was 1736 Hz/pixel with an echo spacing of 0.72 ms and 2232 Hz/pixel with an echo spacing of 0.69 ms, for the SE-EPI and SPEN experiments respectively.

Single-Scan DWI in the Presence of Multiple Chemical Shifts

Figure 1 illustrates the SPEN sequence used in this study. It includes a FatSat pulse; an initial slice-selective $\pi/2$ excitation; a diffusion weighting block where bipolar gradient pulses of duration $\delta$ are separated by a diffusion time $\Delta$; a SPEN module including an adiabatic $\pi$ sweep in the presence of a gradient $G_0$, a FOV defined by the adiabatic frequency range $\gamma G_0$FOV (and thereby introducing a quadratic evolution phase $\phi_r(t) \propto r^2$ along the encoding gradient’s direction) (29,44,45); and an acquisition including an oscillating $\pm G_\alpha$ gradient providing a modulation that when subject to Fourier Transform (FT) gives the spatial profile along the readout axis, interspersed with blipped $G_\alpha$ gradients resolving the position of each SPEN-encoded spin packet. Focusing on the acquisition along the latter axis only, the ensuing signal is of the form
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\[ S(t) \propto \Delta r \times \sum_{\text{sites}} \int_{\text{FOV}} \rho_{\text{cs}}(\Omega_{\text{cs}}, r) \cdot \exp\left[i\phi_{\text{cs}}(r) + k_d(t) + \Omega_{\text{cs}} t\right] dr \]

where \( \Delta r = \frac{\text{FOV}}{r_{\text{cent}}} \) is a nominal spatial resolution defined by the encoding characteristics, \( \rho_{\text{cs}}(r) \) is the spin density for a particular chemical site of shift \( \Omega_{\text{cs}} \), and \( k_d(t) = \gamma \int_{0}^{\text{FOV}} G(t) dt \). The linear spatial nature of \( k_d \) means that SPEN’s highest-sensitivity point, contributing at an instant \( t \) to the signal \( S(t) \), will be unraveled on a \( t \)-dependent manner. This in turn suggests that each image pixel will be modulated along the SPEN dimension by a phase evolution that is proportional to \( V \), pixel will be modulated along the SPEN dimension by a dependen manner. This in turn suggests that each image pixel will be modulated along the SPEN dimension by a phase evolution that is proportional to \( \Omega_{\text{cs}} t \), reflecting the chemical shift(s) of the emitting species. While the SPEN image retrieval process does not necessitate Fourier processing, it follows that FT along the SPEN/time acquisition axis, will yield the sample’s chemical shift spectrum (38). The SPEN-specific portion of the sequence in Figure 1 can, therefore, provide a spatial image discriminated according to chemically shifted resonances, if the following procedure is adopted: (i) A modified signal \( S_d(t) \) is calculated, given by the acquired \( S(t) \) in Eq. (1) demodulated by the \( i^2 \) quadratic contribution associated to \( \phi_{\text{cs}}(r) \) i.e., multiplied by \( \exp[-i\phi_{\text{cs}}(r)] \); this is carried out in a procedure that already includes a super-resolution image reconstruction along the SPEN axis (31,33). (ii) An FT is applied to the ensuing arrays to resolve the magnetic resonance spectrum of every chemical shifted peak along the SPEN-encoded axis. (iii) Individual contributions arising from each chemical-resolved site are isolated (or removed) by applying a suitable spectral filtering—in our particular case aiming at erasing the contributions of unwanted residual silicone peaks. (iv) An inverse FT is applied to reconstruct the raw 2D fat-suppressed data (Fig. 2a), and apply- ing fat) and a medical silicone breast implant (Fig. 2e). Phantom Imaging Example

RESULTS

Figure 2 illustrates the main steps adopted for the SPEN-based acquisition of breast images free from silicone signal interferences, illustrated with a single-slice experiment collected on a 12 cm diameter round plastic phantom containing tap water, a tube with oil (mimicking fat) and a medical silicone breast implant (Fig. 2e). As a first step in the experiment a chemical shift resolved spectrum of the specimen was obtained by taking the raw 2D fat-suppressed data (Fig. 2a), and applying on it a FT along the readout axis and a super-resolved reconstruction eliminating from the signal the parabolic \( \frac{\gamma^2}{2T_2^2} \) phase contribution along the SPEN dimension. Applying then a FT along the latter axis reveals the frequencies of the silicone signal to be removed, as well as of the water signal to be targeted (Fig. 2b). With this separation, the silicone (and if need be the residual fat) contribution to the total image was eliminated by suitably masking a region in the ensuing...
spectral/spatial 2D plane (Fig. 2c). Replacing in this manner the silicone peak by zeroes and suitably re-processing the data, leads to a clean 2D image of the targeted phantom containing mostly water and collected in a single scan (Fig. 2d). This is devoid from the interferences that would otherwise arise from a regular processing of the SPEN-collected data (Fig. 2f). This masking procedure leading to the elimination of the silicone signal will impose a partial loss in the resolution of the water-derived signal along the SPEN-encoded direction. The resolution penalty will depend on the spatial overlap between the nulled and the targeted resonances; for the case of well-resolved resonances like water and silicone, a common loss will entail a factor of two [39]. Still, scan parameters can be planned so as to take this penalty into account beforehand, and still achieve a desired pre-targeted resolution. The Supporting Information, which is available online, illustrates this case-by-case dependence as well as the efficiency of the silicone suppression, with the aid of numerically synthesized examples.

**Separated Water and Silicone Images in a Single Scan**

Figure 3 illustrates the silicone filtering procedure, with the single-shot acquisition of separate silicone- and water-based images from a single breast anatomical scan. Used in this evaluation was SPEN’s capability to evaluate tissues using a reduced FOV, while avoiding folding. Figure 3a shows a high-definition TSE multishot scan showing the boundaries of the silicone implant, the fibroglandular tissue, and the fatty tissue. Applying an acquisition scheme similar to that described for the phantom in Figure 2 yields solely the silicone and water contributions; a suitable postprocessing then enables one to target either one of these peaks, leading to a selective separation of the anatomical images arising from either fibroglandular tissue (Fig. 3c) or from the silicone implant (Fig. 3d), both reconstructed from the same single-shot scan. Notice as well that if the SPEN image is processed without this procedure (Fig. 3b) the strongest peaks (fat and silicone) are restored to their original location, while the water peak is chemically shifted and overlaps/interferes with the silicone signal.

**Breast ADC Maps Based on Anatomical, Silicone-Free Images**

Seven healthy volunteers with a variety of silicone implant sizes were scanned using the sequence in Figure 1, and their breast ADC maps were generated. Figure 4 shows
representative results on a healthy volunteer, presenting a T2-weighted multiscan anatomical image (Fig. 4a), as well as anatomical b-zero images (Fig. 4b) and ADC maps (Fig. 4c) delivered by DWI SE-EPI and by multislice SPEN, respectively. To avoid folding artifacts the FOV used in the SE-EPI scans was chosen to cover the whole body along the anterior→posterior direction, whereas in the folding-free SPEN scans the FOV along the low bandwidth direction could be reduced to half this size. Even after masking low-intensity noises and ghosts, the b-zero images afforded by SE-EPI show a noticeable chemical shift displacement artifact of the silicone implant, which together with a sizable ghost of both the water and silicone signals (yellow arrows in Figure 4b) appear below the fibroglandular tissue. For the same volunteer and under very similar conditions SPEN succeeds to deliver images that are much less influenced by ghosts, showing the fibroglandular tissue in its right location and a considerably weaker silicone contribution. Still, a residue from the implant can be seen in the filtered anatomical SPEN image as a result of an imperfect elimination can occur in the final procedure here introduced for silicone removal is based on a manual ROI intervention, imperfect elimination can occur in the final step (Fig. 4b), and of standard availability in scanning softwares. As an alternative we compared SPEN’s performance with twice-refocused SE-EPI diffusion methods of common use in clinical scanners (41). When using such methods poor image quality and severe artifacts resulted, particularly when sizable fat and silicone resonance contributions were present in the targeted FOV. This study showed that sequences based on SPEN principles, could assist in handling such multiple chemical contributions in this spatially heterogeneous organ. Supporting this improved performance is SPEN’s ability to provide single-scan images that are less affected by field inhomogeneities than SE-EPI’s counterparts (25,37), as well as SPEN’s built-in capacity to separate multiple spectral contributions without demanding additional or modifications to the original single-scan 2D imaging sequence. Because the procedure here introduced for silicone removal is based on a manual ROI intervention, imperfect elimination can occur in the final magnitude images (Fig. 4b); still, a ≥90% elimination was readily achieved at the center of the silicone’s ROI. When combined with multislice operation and a diffusion-sensitizing protocol, a new approach to retrieve ADC maps of fibroglandular tissue in augmented breasts emerges. These SPEN measurements evidenced comparable ADC maps as those afforded by SE-EPI experiments, yet substantially freeer from a variety of artifacts. Both methods showed relatively low diffusion values (mean ADCs ≈ 1.6 × 10⁻³ mm²/s); these are probably related to low breast density and partial volume effects, yet in agreement with values of previously reported studies

A final example of the advantages resulting from this silicone-filtering DWI methodology is illustrated in Figure 5, which compares the performance of SE-EPI and SPEN sequences for a multislice scenario. Anatomical multislice T2-weighted images (Figure 5, upper row) outline the structural tissue organization of the breast, involving fatty and fibroglandular tissue together with the silicone implant. Upon evaluating the ADC maps arising from SE-EPI and from silicone-filtered SPEN experiments (Figure 5, middle and bottom rows respectively), much cleaner images can clearly be retrieved from the latter. The original compartmentalization of the fibroglandular tissues, the silicone implant and the fatty tissues are preserved for the different SPEN slices. The overall diffusion analyses for the fibroglandular tissues including all five subjects lead to mean ADCs of 1.63 ± 0.13 × 10⁻³ mm²/s for SE-EPI, and 1.59 ± 0.18 × 10⁻³ mm²/s for SPEN
When considering further DWI analyses incorporating SPEN-based chemical shift discrimination for evaluating breasts with silicone implants, the limited number of slices and limited repetition times TR that specific absorption rate (SAR) currently permits to collect at 3T, need to be taken into account. This limitation derives from the sequence’s use of a swept 180° inversion pulse, acting in the presence of a gradient that is
made purposefully strong for the sake of avoiding distortions. Indeed, the robustness at the heart of all SPEN-derived sequences stems from the fact that, because SPEN’s encoding/acquisition bandwidths (BW_{enc} = B - W_{enc}) are not restricted by Nyquist criteria, they can be – and usually are – set substantially larger than their EPI’s phase-encoded counterparts. Studies of challenging organs like the one in our manuscript rely on such advantage, and use ca. 5–10 times larger bandwidths than EPI to overcome offset- and susceptibility-derived effects. In such instances the effective average G_r gradient along the SPEN dimension will be 5–10 times larger than EPI’s SARs on the other hand, will be concomitantly higher than EPI’s by roughly the same amount. Approaches to alleviate this technical limitation will be presented in an upcoming study. Only five slices were shown in the proof-of-principle example summarized in Figure 5; still, it should be pointed out that SPEN-based experiments like this one could be repeated on up to 20 slices with relatively small SARs (≤40%) and in reasonably short TRs (≤20 s). These are suitable parameters if considering the use of this approach in diagnostic clinical settings.

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REFERENCES

Supporting Information

Additional Supporting Information may be found in the online version of this article.

SUP. FIG. S1. Resolution losses and spectral suppression effects expected from the masking procedure introduced in this study, illustrated on the spectral/spatial synthetic phantom illustrated on the left-hand panel. This phantom was simulated assuming a matrix size $512 	imes 128$, a 20-cm-square FOV, and a SPEN-axis offset displacement between the silicone and water peaks of 55 pixels, equivalent to a chemical shift separation of 4.3 ppm at the assumed 3T field. Indicated in this panel are also the SPEN/readout (RO) directions, as well as four single-pixel nicks that were introduced on purpose within the water ring to better appreciate the resolution degradation along the $y$-axis for the different $x$ positions. See text for further details.

FIG. S2. Illustrating the resolution losses associated by increasing spectral suppression masking regions, for the same synthetic phantom as introduced in Supporting Figure S1. As in that Figure the top panels illustrate the mixed shift/spatial 2D correlation arising upon Fourier processing the idealized phantom along the SPEN axis; indicated in these identical maps are progressively larger regions chosen to mask out a potential silicone contribution. The bottom panels indicate in turn the progressively stronger $y$-blurring, affecting the remaining water image associated to the masked-out $x$-coordinates, highlighted by the white brackets and evidenced by distortions in the $x = 0$ nicks.