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Positive Contrast Visualization of Nitinol Devices using Susceptibility Gradient Mapping

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Abstract

MRI visualization of devices is traditionally based on the signal loss due to T_2^* effects originating from the local susceptibility differences. To visualize nitinol devices with positive contrast a recently introduced post processing method is adapted to map the induced susceptibility gradients. This method operates on regular gradient echo MR images and maps the shift in k-space in a (small) neighborhood of every voxel by Fourier analysis followed by a center of mass calculation. The quantitative map of the local shifts generates the positive contrast image of the devices, while areas without susceptibility gradients render a background with noise only. The positive signal response of this method depends only on the choice of the voxel neighborhood size. The properties of the method are explained and the visualization of a nitinol wire and two stents are shown for illustration.

Introduction

The MRI visualization of local susceptibility differences, such as induced by metallic devices or by iron-oxide contrast labeled cells is traditionally based on the T_2^* effect they induce through the local magnetic field disturbances [1,2]. The decrease in T_2^* results in a signal loss and thus a negative contrast relative to the background tissue. For the visualization of devices, but also for ease of quantification in other applications, it is attractive to convert this signal loss into positive contrast on a background that is as

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List of symbols

δ	Greek delta
Δ	Greek (capital) Delta
γ	Greek gamma with strike-through
A_k	Capital A with subscript k
δ_n	Greek delta with subscript n
δ_{max}	Greek delta with subscript "max"
x	small x
π	Greek pi

homogeneously dark as possible for all unaffected tissue. In recent years, several approaches have been introduced that generate positive contrast by imaging susceptibility differences.

As a post-processing method, inverting the gray levels will yield positive contrast, but the background intensity variations are still present and quantification is therefore not straightforward.

Several methods for positive contrast generation are based on changes in the local resonance frequency: the induced magnetic field-offset changes the Larmor frequency locally and this effect can be exploited by using e.g. off-resonance frequency selective excitation [3], on-resonance background suppression (e.g. IRON [4]), off-resonance SSFP imaging with high or low flip-angles [5-7]. To these frequency based methods, methods with multiple acquisitions can be added which were originally designed for water/fat separation based on their frequency difference, such as SSFP offset averaging [8] and multiple echo combinations (e.g. IDEAL [9]).

A separate class of visualization approaches operates in k-space taking advantage of the shifted spectral information in k-space caused by the locally induced magnetic field gradients [10]. By adding additional gradients during imaging, or as a post processing step by moving the reconstruction window in k-space, the background can be suppressed, while the shifted susceptibility signal is conserved (the white marker method [11]). Another post processing method is the susceptibility induced gradient mapping (SGM) method, where the spectral shift is measured in a (small) neighborhood by local Fourier analysis [12]. The resulting shift can be measured on a voxel-by-voxel basis to yield an off-resonance image with a suppressed background. An inherent advantage of such a post-processing technique over a modified acquisition approach is that the original image depicting the surroundings of the device is unchanged and that co-registration is facilitated.

In this paper, the SGM method is adapted for the visualization of nitinol devices, which requires a highly localized visualization of the positive contrast, while the quantitative nature of the method is an additional benefit. The properties of the method are analyzed and its application is demonstrated through the visualization of two modern nitinol stents.

Theory

In the presence of a device in the static magnetic field of the MR scanner, there is a change in the local surrounding magnetic field. Depending on the static magnetic field strength and the susceptibility of the device, the induced field is localized to the immediate surroundings of the device leading to large local field gradients. In MRI these additional local field gradients cause image distortion in the readout direction visible as deformation of objects and white and black spots of signal accumulation and cancellation, respectively. These effects can be minimized by using strong (compared to the local gradients) readout gradients (see e.g. [5] and [13]). Another effect in both readout and phase encoding directions when using gradient echo imaging, is that the echo formation of the affected spins will be shifted away from the center in k-space. The extent of this relative shift depends on the strength of the local susceptibility induced gradient ΔG and the echo-time TE , which is the effective duration that ΔG exerts its influence on the imaging experiment. When neglecting the distortion in the read-out direction, the shift in k-space will then be the ratio of the time-susceptibility gradient area and the imaging-gradient surface area per k-space step. The shift in k-space for a Cartesian acquisition in units of steps in k-space in one direction is given by

$$\delta = \frac{\Delta G \cdot TE}{A_k} = \gamma TE \cdot \Delta G \cdot FOV$$

[1]

with $A_k = \Delta k$ the area under the imaging (readout or phase-encoding) gradient to add one step in k-space in that direction, FOV the field of view and γ the gyromagnetic ratio for protons divided by 2π ($=42.57$ MHz/T). By measuring this shift δ in all spatial directions, the local susceptibility gradient strength can be determined.

One approach to determine this shift is through local spectral analysis [12]: Assume a volume of interest (VOI) of n voxels around the voxel of interest. This VOI is Fourier transformed into the frequency domain thus related to the original k-space through a windowing function. When all voxels in the VOI are on-resonance, the peak of the obtained spectrum will be centered around zero. The signal intensity differences in the VOI yield a symmetrical spectrum around this zero frequency. However, when local magnetic field gradients are present, a part of the spectrum is shifted away from the zero frequency. Thus, the center of the spectrum will be shifted away from the origin. The shift depends on the relative weight of the off-resonance part, as not all magnetization in the VOI may shift or some may even shift in another direction. When the spatial extent of the susceptibility gradient field is smaller than a voxel, negative and positive shifts will cancel and no net shift is measured. Thus, it is important to match the spatial resolution of the image to the scale of the gradients to be visualized. In our case the stent struts are typically 1 mm apart and image spatial resolution has to be approximately n times (i.e. kernel size) higher.

For the measurement of the shift, we propose to determine the center of mass of the magnitude of the spectral signal inside the VOI. By using odd numbers for n , this is a weighted summation. When using even numbers of n , the calculation becomes more complicated as the spectrum is asymmetrical ranging from $-n/2$ to $n/2-1$. In the implementation by Dahnke et al. curve fitting was used to determine the spectral peak as a metric of the shift instead of the center of mass [12].

Consider a 1D example starting with an image with a field of view FOV and a matrix size m . The original voxel size Δx is FOV/m and the corresponding step size Δk in k-space is $1/FOV$. Now we locally sample a VOI of n voxels with a $FOV_{voi} = n\Delta x$. Note that this sampling of a subset in image-space equals to convolution of the data matrix with size m in k-space with a sinc function $n\Delta x \text{ SINC}(\pi k n\Delta x)$ [5]. Note that this convolution acts as a filter, decreasing the effective spatial resolution proportionally with n . By analyzing the data-points of the VOI in the frequency domain, the corresponding discrete steps Δk_{voi} in the convoluted k-space become $\Delta k_{voi} = m\Delta k/n$. For a certain choice of kernel size n , the measured shift δ_n in the center of mass is now expressed in k-space steps of m/n times larger than in the original k-space steps, thus

$$\delta_n = \frac{n\delta}{m},$$

[2]

Combining Eq.[1] and [2], the local gradient strength ΔG can be found as

$$\Delta G = \frac{\delta_n}{n \cdot \Delta x \cdot TE}$$

[3]

This 1D example described in the above paragraph is applicable to all spatial directions. The method can thus be applied to 2D and 3D images, where n does not necessarily have to be equal in all directions, but care has to be taken to account for possible differences in spatial resolution. In our implementation we chose to map the length of the vector created from the δ_n in the three orthogonal directions, and map this length as the positive contrast SGM image.

When using the center of mass metric to describe the spectral shift, the maximum shift δ_{max} in any direction is described by [Eq.1]. To describe the influence of noise, assume of a VOI with a noiseless k-space spectrum $p(k)$ and a shift δ determined with the weighted sum $\delta = \sum k \cdot p(k) / \sum p(k)$. Define signal $S = \sum p(k)$ and add (symmetrical) noise with $N = \sum q(k)$, then the maximum resulting shift δ_{max} will decrease to

$$\delta_{max} = \frac{\sum k \cdot (p(k) + q(k))}{\sum p(k) + q(k)} = \frac{\delta \cdot S}{S + N} = \frac{\delta \cdot SNR}{1 + SNR}, \quad [4]$$

with $SNR = S/N$ and $\sum k \cdot q(k) = 0$ because the noise is symmetric. In case of a low SNR experiment (e.g. at high resolution and/or very strong T2* induced signal loss), the measured shift will thus be affected by noise. The use of [Eq.3] will then underestimate the gradient ΔG .

The choice of the kernel size n changes the output of the method, see [Eq. 2]. Larger values for n decrease Δk_{voi} , thereby increasing the spectral resolution and making the procedure sensitive to smaller local gradients in a given image. However, this comes at a price of including more background signal or voxels with different gradients in the VOI, thereby lowering the relative signal contribution of the local field gradient present in a certain voxel. This in turn will decrease δ_n , limiting the use of [Eq.3]. The optimum choice of n will thus depend on the strength and spatial variation of the susceptibility gradients present, the image resolution and the available SNR.

Methods

In order to demonstrate the method and to characterize its properties, phantom measurements were performed: Two folded nitinol wires (about 30 cm in length and 0.5 mm in diameter) were suspended in 1 liter of 1% agarose inside a plastic container. One centrally placed wire was curved in a doughnut shape. The second wire was more densely packed in a ringlike shape and positioned at the nine o'clock position in Fig 1. To prevent mold and to supply a fatty structure in the background, a 7 mm layer of vegetable oil covered the agarose.

Imaging was performed on a 3T whole body MRI scanner (Achieva R1.7, Philips Medical Systems, Best, the Netherlands) with a 8-channel SENSE-head coil for signal reception. The scan parameters were: 3D-Gradient Echo, TR=23 ms, $\alpha=25^\circ$, coronal orientation, matrix $224 \times 224 \times 36$, voxel size $0.76 \times 0.76 \times 0.76 \text{ mm}^3$, readout bandwidth 434 Hz/pixel, one signal average, full echo, no ringing filter, linear shim, real and imaginary images reconstructed, scan time 3:06 min. Scans with incremental echo-times (TE) of 3, 4, ... 10, 12, 14, 16 and 20 ms were acquired. From the real and imaginary images, the complex images were reconstructed off-line on a separate computer (2.0GHz Pentium-M laptop) running the iX software development environment (<http://ix.isi.uu.nl>, Image Sciences Institute, Utrecht, the Netherlands). The spectral analysis method was implemented in C++ within the iX environment, where the FFTW library (FFTW 2.1.3, Massachusetts Institute of Technology, Cambridge, MA) is used for Fourier transformation for arbitrary n .

In a first example, the SGM method was applied to the entire 3D dataset. Source images and resulting maps were compared for different echo-times and (odd) kernel sizes n .

For the analysis of the positive signal we selected sets of voxels (SOVs) in the surroundings of the object of interest. This selection was performed by including voxels for which a specified shift in k-space was determined.

The influence of TE on the SGM images was examined by measuring the average spectral peak shift for SOVs in images with different TE using a fixed kernel size $n=3$. Four SOVs were constructed by selecting voxels inside a narrow interval with values in the range from 0.20 to 0.22 pixels of shift on SGM-maps that were generated from images acquired with TE=3ms, TE=6 ms, TE=10ms and TE=20 ms respectively. The SOVs were restricted to the area around the left wire in Fig 1. As a reference, a manually drawn SOV in the background was measured as well. For each SOV defined as mentioned above, the spectral shift was measured in the SGM images constructed from MR images with different TE (including the one that was used to define a specific SOV). The linear relationship between the measured shift and TE as predicted in [Eq.1] was verified by linear curve fitting and by inspecting the correlation coefficient (R^2). Values of $R^2 > 0.95$ were considered statistically significant. For the SOV representing measured k-space shifts of 0.20-0.22 pixels at TE=3 ms, the decline of the measured shift for large values of TE was tested for statistical significance.

The influence of the kernel size n of the local analysis VOI was investigated in a similar way as for the TE dependence. Starting with a kernel size of $n=9$ and TE=10 ms, three SOVs were created that represent k-space shifts ranging between 0.2 and 0.22, 0.4 and 0.44 and 0.6 and 0.66 pixels, respectively. Additionally, a SOV of background voxels was selected. Then, for every SOV with the selected shift for $n=9$, the spectral shift was determined for $n=3, 5$ and 7. The investigated value ranges were chosen in the midrange of values present in the SGM image. The starting kernel size $n=9$ was chosen to start with to ensure linearity (midrange values chosen for small n will exceed the dynamic range when the SOVs are applied to images with larger n). The expected linearity of the shift in pixels as a function of n as described by [Eq.2], was verified by linear curve fitting and quantification of the correlation coefficient (R^2). A value of $R^2 > 0.95$ was considered statistically significant.

For the purpose of detailed visualization, a high-resolution 3D image of the phantom described above was acquired. The scan parameters for gradient echo imaging were TR/TE/ $\alpha=9.4/4.0/22^\circ$, axial orientation, acquired voxel size $0.3 \times 0.3 \times 0.3 \text{ mm}^3$, acquired matrix $400 \times 400 \times 141$ interpolated to $800 \times 800 \times 282$. Readout bandwidth 816 Hz/pixel, one signal average, full echo, no ringing filter, linear shim, scan time 6:47 min and an 8 element SENSE-head coil was used for signal reception. Because of the size of the dataset, a sub volume was selected around the center wire of $238 \times 88 \times 214$ voxels (i.e. $36 \times 13 \times 32 \text{ mm}^3$). To

improve the signal to noise ratio, a non-linear edge enhancing filter was applied to the complex data, for details see [14].

The SGM method was then applied with $n=3$ and the result visualized with maximum intensity projection (MIP) in iX and a straightforward iso-surface rendering using a VTK routine (The Visualization Toolkit 4.2.1, www.vtk.org, © Ken Martin, Will Schroeder, Bill Lorensen).

To study the method in clinically approved nitinol devices, a 7×30 mm carotid stent (Acculink, Abbott Vascular Inc, Redwood City, CA) and an 8×40 mm peripheral stent (Xpert, Abbott Vascular Inc, Redwood City, CA) were suspended in gelatin. Imaging was performed with the stents aligned in parallel with the main magnetic field. The imaging sequence was a coronal 3D gradient echo spiral acquisition (with phase encoding in the slice select direction). Imaging parameters were TR/TE/α=17/2.9/40°, 7 ms readout, 400 spiral arms, reconstruction matrix 640×640×124 with a voxel size of 0.19×0.19×0.21 mm³, scan time 18:07 min. Real and imaginary images were processed as described above and after non-linear edge-enhancing filtering an SGM image was generated for kernel size $n=3$.

Results

In Fig 1 images of slices are displayed that were obtained at the level of the nitinol wires. The left wire (white arrow in Fig 1a) has a relatively strong susceptibility effect (dark rim). The same slice is shown for incremental echo times of 3, 10 and 20 ms. The matching slices of the SGM images for $n=3$ are displayed in the second row, and another row of SGM images of this slice for TE=10 ms with $n=5, 7$ and 9 is shown at the bottom. With an increase in TE, the susceptibility gradients gain more influence and the corresponding SGM images show an increased amount of positive signal (blooming around the wire). With increasing kernel size n , the extent of the positive contrast area increases as well, and smaller susceptibilities show up in the $n=9$ image, such as the enhancement at the top and bottom of the phantom. Simultaneously, the spatial resolution visibly decreases, as details between the windings become blurred (compare e.g. Fig 1e and 1i). The computation of the images takes less than a minute for $n=3$ (of a 224×224×36 image), up to 4 minutes for $n=5$, but up to 1 hour for $n=9$, as the amount of FFTW-calculations per voxel increases.

In Fig 2 the measured shift δ_n of the center of mass in 3D k-space is shown as a function of TE for a fixed value of $n=3$. The measurements are averaged over a SOV located near the wire on the left (in Fig 1) as described above. In Fig 2 an insert shows the spatial distribution of the SOVs, where the SOV defined by the value range of 0.2-0.22 on the image with TE=3 ms, is closest to the wire (shown bright), while the SOVs where that range is present on images with longer echo times, are located more peripherally (in a darker shade of gray). The plots of the measured shifts show a linear increase in shifts in k-space for smaller values of TE. For the first four data-points the correlation coefficient is $R^2 \geq 0.99$. However, the values level off for larger TEs. The maximum of the curves is around 0.3-0.35 pixels, which for TE=10 ms represents an average gradient $\Delta G = 0.3$ mT/m (see Eq. 3). The decrease of the shift of the curve for the SOV defined on the TE=3 ms image, but measured on images with TE>10 ms, is statistically not significant. For different SOVs the measured standard deviation (not plotted) is roughly proportional to the echo-time, i.e. for TE=3 ms the standard deviations range from 0.05 to 0.13, at TE=6 ms from 0.06 to 0.14, at TE=10 ms from 0.08 to 0.12 and at TE=20 ms from 0.06 to 0.12. In the background, the standard deviation is between 0.008 and 0.012.

In Fig 3 the results of the SGM method as a function of the kernel size n are shown for values obtained from SOVs close to the left wire (Fig. 1). In the selected SOVs, linearity

was observed between the measured shift and n , with correlation coefficients $R^2 \geq 0.97$. The inset in Fig 3 shows typical values present in the SGM image: higher values centrally (with 0.6 and 0.4 depicted in gray) and smaller values (0.2 shown in white) at the periphery of the susceptibility effect.

A more detailed visualization of the more complex shaped central wire from Fig 1 is shown in Fig 4. For reference, a slice of the original high-resolution dataset in the acquisition (transverse) plane is shown in Fig 4a together with a Minimum Intensity Projection of a 10.5 mm coronal slab in Fig 4b. Note the larger susceptibility effect (and thus more conspicuous SGM visualization) from structures perpendicular to the main magnetic field, as compared to wire-segments oriented in parallel.

In Fig 5 the carotid (left) and peripheral (right) stents are visualized. A photograph of the two stents is shown for reference of the actual strut geometry (Fig 5a). A 1.4 mm thick signal-averaged slab of the original dataset is shown (Fig 5b). Adjacent (Fig 5c), a 4 mm signal inverted minimum intensity projection of the stents is shown. In the second row, a 3D surface rendering is shown created from the SGM image (Fig 5d). Fig 5e shows a central 1.4 mm slab through the center of the SGM image comparable to Fig 5b. Adjacent to this, a MIP of 4 mm of the SGM image is shown (Fig 5f).

Discussion

In this paper we have described an MRI post-processing method to visualize devices by extracting a susceptibility-induced gradient-map through local analysis of the spectral shift (SGM). The required input for the method is a regular 2D or 3D (complex) gradient echo image. The method, originally described by Dahnke et al. [12], has been adapted for visualization of nitinol devices by expanding it and include odd values for the kernel size n . This enables a simple computation of the center of mass shift of the local spectrum in k-space. Furthermore, it obviates non-linear fitting to find the spectral peak, thus improving computational effectiveness, and ease of implementation. Moreover, for $n=3$ the positive contrast is better localized close to the device with a higher spatial resolution, improving the visualization (as compared to a minimum of $n=4$ in the original method).

When applying the SGM method to images with low SNR, the measured maximum shift in k-space will be limited. Thus, the relationship between n and the shift of the center of mass in k-space is not completely linear (see Fig 3). While the general trend appears to be linear, there is a sub-linear component visible for larger shifts with higher values of n . Analogous to Fig 2, a plateau will be reached where the values for δ_n will not increase, due to partial volume averaging and noise. An experiment with a SOV defined on the $n=3$ SGM image (least sensitive kernel size, thus representing large local gradients) and applied to $n=5, 7$ and 9 maps, showed indeed a constant measured average shift of 0.2 with large error-bars ranging from 0.08 to 0.18 (data not shown but analogous to top curve in Fig 2.).

The choice of the kernel size n determines the sensitivity for the k-space shift measurements and larger n will increase the sensitivity. Secondly, it determines the relative noise-sensitivity (see Fig 1 and 3, where the background value does not increase while sensitivity to the susceptibility increases as more pixels are included in the VOI). Finally, it determines the spatial resolution and localization of the measured shifts. When n increases, the size of the spatial kernel for the analysis filter increases and the effective resolution of the output map decreases. Even though the method is applied on a pixel-per-pixel basis, the effective spatial output-resolution will be (approximately n times) lower due to the convolution in k-space with the sinc function that determines the local VOI.

When optimizing an imaging sequence for device visualization with the SGM method, several considerations can be taken into account: First, the amount of T2*-dependent signal loss is indicative of the magnitude of the shift (see Fig 1). An intermediate amount of dephasing seems desirable, as small dephasing will show little shift (e.g. Fig 1d with TE=3 ms), while complete signal loss will decrease local SNR and thus limit the measurable shift (e.g. Fig 1f in regions close to the wire which are not brighter than the periphery, although the local field gradients are larger and thus the expected values in the SGM image). This constraint limits the choices for echo time and spatial resolution. For the native T2* visualization, the optimal parameters will likely go in the same direction. Further, the desired resolution of the SGM image has to be considered, as the spatial resolution of the input image has to be approximately n times higher than the output resolution, which may limit the choice of the receiver bandwidth to obtain sufficient SNR. For the visualization of stents, the distance between the struts is of the order of a millimeter or less. The already high spatial resolution used, was thus barely enough. *In vivo*, these resolutions would yield prohibitively long scan times, requiring further improvements in scan efficiency before clinical application would be possible. A third consideration in the optimization is the quality of the shimming. At the top and bottom edge of the circular phantom shown in Fig 1 there is residual enhancement, originating from a strong residual higher order B₀-field inhomogeneity consistent with linear shimming only. This effect is geometry dependent and may influence the choice of the spatial orientation of the imaged volume. However, higher order shimming may help to alleviate this problem [15].

The two nitinol stents were visualized with high resolution using a spiral acquisition. While the calculations are less intuitive than for Cartesian imaging, the location in k-space is still given by integrals of the gradients over time. Thus, the concept of k-space shifts induced by the local field gradients acting between excitation and echo-time can still be used to explain the SGM method. In the native images however, the field distortions have a different behavior. Around the (dark) struts a small white halo is seen (Fig 5b), resulting from image distortion (with signal displacement and accumulation). Both the native image and SGM image created with $n=3$, show the overall geometry of the stents rather well, but the finer details of the struts are no longer visible. For example, the surface pattern of the peripheral (right) stent as seen in Fig 5c and 5f represent the closed loop between the parallel ring-shaped components and the joints between them. The fine triangular structure of the struts creating these loops is however not directly apparent (Fig 5a). The crown-shaped intensity at the top and bottom of the stents stems from the field disturbances at the endpoints of wires in parallel to the main magnetic field. These field effects as well as the attenuated signal due to shielding inside the stents (Fig 5b) are expected to be geometry dependent (orientation of the object with respect to the main magnetic field, dimensions of the object etc.), and have already been discussed in the literature [16]. A small value of n was chosen to maximize the spatial resolution of the SGM image.

As quantification of the field disturbance is a less interesting feature for visualization purposes, the main advantage of the SGM method is the very homogeneous background suppression (apart from very small air inclusions, best visible next to the carotid (left) stent). The inverted native image shows residual background inhomogeneity due to B₀-inhomogeneity after volumetric shimming (white arrow in Fig 5c), even without structures in the background, such as found *in-vivo*. However, the applicability of *in-vivo* device visualization using SGM still needs to be determined in future research.

Conclusion

The presented Susceptibility Gradient Mapping method is specifically adapted to visualize conventional nitinol devices with positive contrast. By using a modified measure for the

spectral shift in k-space, the computational complexity is decreased. Additionally, the analysis neighborhood size, the only post-processing parameter to be chosen, can be made smaller, therefore allowing processing in a region closer to the device. As with regular SGM, the method exploits the susceptibility information present in regular 2D or 3D gradient echo MRI acquisitions. The applicability of the method has been demonstrated *in vitro* through visualization of two nitinol stents.

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Fig 1.

Original gradient echo image of the agarose phantom imaged with (a) TE=3ms, (b) TE=10 ms and (c) TE=20 ms. The phantom contained two folded nitinol wires. The one in the center was curved and the second on the left (arrow) was more densely packed in a ringlike shape. Small air inclusions were present (see e.g. arrowhead). Second row (d,e,f) are the matching SGM images with $n=3$ for TE=3, 10 and 20 ms respectively. Third row (g,h,i) shows SGM images for TE=10 ms with $n=5,7$ and 9. The increased sensitivity reveals the shimming limitations in the phantom with a background gradient remaining. The window-level is equal for these 6 images.

Fig 2.

Average shift δ_n of the center of mass in k-space measured in sets of voxels (SOVs) on $n=3$ SGM maps as a function of the echo time (TE). The SOVs were defined by selecting voxels that had a response within a range of 0.2 to 0.22 in the SGM map for a particular TE (the corresponding data points are marked by the gray line). Chosen were (●) TE=3 ms, (■) TE=6 ms, (▲) TE=10 ms, and (○) TE=20 ms respectively. A linear fit is shown through the first four data-points. For reference a background SOV (□) is shown. The inset on the right shows the location of the SOVs using a T2*-image and a corresponding projection of 11 slices of the different SOVs with a range from white representing the SOV defined on the TE=3 ms image (●) which is close to the wire up, to dark gray for the SOV from the TE=20 ms image (○) at the outer edges.

Fig 3.

Average shift δ_n of the center of mass in k-space measured in sets of voxels (SOVs) on the SGM maps for different kernel sizes n with TE=10 ms. The SOVs were determined by selecting voxels that for $n=9$ have a shift between (■) 0.60 and 0.66, (▲) 0.40 and 0.44, (●) 0.20 and 0.22. These SOVs were subsequently applied to investigate what shift these voxels have for $n=3, 5$ and 7 . The small gray boxes next to the $n=9$ markers, indicate the ranges used to define the SOV. A SOV in the background (○) was drawn for reference. The linear curve with the correlation coefficient R^2 fit is shown for guidance. The shown values include one standard deviation and the data-points for ▲ and ○ are shifted $-0.05/+0.05$ in the x-direction for visual clarity. The inset shows a MIP of five slices of the SOVs with (■) in dark gray, (▲) in light gray and (●) in white.

Fig 4.

(a) Slice from a high-resolution 3D dataset at the level of the central wire (arrow). (b) Minimum Intensity Projection of 10.5 mm slab in the coronal plane after non-linear filtering of a subvolume (input to the SGM method). The main magnetic field was oriented top-down in this image. (c) Maximum Intensity Projection of the SGM method with size $n=3$. Note the larger thickness in the horizontally (perpendicular to the main magnetic field B_0) oriented struts (white arrow) as compared to the parallel ones (arrowhead). (d) Iso-surface rendering of the central wire based on the data shown in (c).

Fig 5.

(a) Photograph of the carotid (left, 7x30 mm) and peripheral (right, 8x40 mm) stent mounted on pencils to show the struts. Black arrows in the magnification show where the struts are interconnected. (b) 1.4 mm thick average slab of the original spiral gradient echo dataset, showing signal loss at the struts (black arrowhead), and signal enhancement around them (white arrowhead). Inside the stents there is signal loss attributable to shielding (black arrow). At the top and bottom little (round) field disturbances (white arrow) are seen. (c) Inverted 4.2 mm slab minimum intensity projection of the native dataset. Some background inhomogeneity is visible (white arrow). (d) Isosurface rendering of the SGM image. Note the small air inclusions in the gelatin (black arrowhead). (e) 1.4 mm thick average slab of SGM image through center of the stents. (f) 4.2 mm slab maximum intensity projection of the SGM image through the stent wall.