Three-dimensional assessment of susceptibility-induced signal voids caused by active cardiac implants in the setting of 3.0T CMR

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Abstract

Purpose

Recent technical advancements allow cardiac MRI (CMR) examinations in the presence of so-called MRI conditional active cardiac implants at 3.0T. However, the artifact burden caused by susceptibility effects remain an obstacle.

Methods

All measurements were obtained at a clinical 3.0T scanner using an in-house designed cubic phantom and optimized sequences for artifact evaluation (3D gradient echo sequence, multi-slice 2D turbo spin echo sequence), as well as reference sequences according to the ASTM were applied. Four representative active cardiac devices and a generic setup were analysed regarding volume and shape of the signal void. For analysis, a threshold operation applied to the grey value profile of each data data set was used.

Results

The presented approach allows the evaluation of signal void and shape even for larger implants such as ICDs. The void shape is influenced by the orientation of the B0-field and the chosen sequence type, as well as the distribution of magnetic material within the implants. The void volume depends both on
the device itself, and the sequence type. Disturbances in the B0 and B1 fields exceed the visual signal void.

Conclusions

This work presents a reproducible and highly defined approach towards characterising both signal void artifacts at 3.0T and their influencing factors.

**Keywords** (3-6): Artifacts, CMR, 3.0T, active implants, signal void

**Introduction**

Active cardiac implants such as pacemakers (PM) and implantable cardioverter-defibrillators (ICD) offer protection against the effects of arrhythmias and the risk of sudden cardiac death in patients with structural pathologies of the heart. Once implanted, these implants have long been considered an absolute contraindication for the assessment of myocardial structure and function with cardiac MRI (CMR). This prevents the (re-)assessment of cardiac morphology, acute and chronic damage as well as an exact evaluation of left and right ventricular function by CMR during the further clinical course of these patients. The electric and ferromagnetic components of active cardiac devices interact with the strong magnetic and electromagnetic fields of the MR system and potentially cause their malfunctioning and permanent damage. Newer developments of so-called MR conditional devices have largely overcome these technical challenges under well-defined conditions. The combination of a specific device system, device programming and a defined MR setting would allow a safely performed CMR examination. However, the artifact burden induced by the MR-conditional implant itself continues to limit the applicability of CMR scans in patients with such devices.

The presence of active cardiac implants induces complex susceptibility artifacts that appear in two ways: first, the magnetic material produces a signal void in proximity to the implant; and second the visible regions are affected by geometric distortions throughout the visible volume. The artifact burden increases with the strength of the magnetic field, and thus poses a greater challenge for successful imaging at 3.0T. At the same time, the higher field strength offers a higher signal to noise ratio than 1.5T scanners and thus provides the potential to allow faster image acquisition and higher image resolution.
At 1.5T, approaches for minimizing the artifacts during CMR imaging have been developed. Modifications of established protocols such as the use of spoiled gradient echo sequences after administration of contrast agents for CINE imaging, rather than balanced steady state free precession (bSSFP) sequences, and broad band pulses for late enhancement imaging have been used to obtain diagnostic image quality in the presence of active cardiac devices. However, these modifications have not yet widely been transferred to 3.0T. Clinical feasibility studies have shown the potential of modified sequences for CMR at 3.0T but do not systematically evaluate the quantitative extent of susceptibility artifacts of different devices.

The different image properties of the susceptibility artifacts have to be analysed separately because they require different measurement setups. On the one hand, a homogeneous background for the evaluation of the signal void is required, on the other hand means to quantify the geometrical changes due to distortion have to be provided. The present work focuses on the signal void and aims to establish an approach towards the precise three-dimensional determination of dimensions and shape of signal void artifacts in phantom measurements as a first step towards image optimization at 3T CMR. Aspects of geometric distortions are discussed in a different publication.

**Materials and Methods**

**Phantom**

The analysis of the signal void requires a sufficiently large phantom that is capable to contain the signal void in all three spatial directions and that offers a homogenous background. A cubic phantom was designed that is filled with a homogeneous aqueous medium. The outer dimensions, as dictated in two dimensions by the 60 cm bore diameter of the scanner, measure 28x28x28 cm. In order to minimize B1-shading artifacts while maintaining an acceptable image contrast within the aqueous filling, a conductivity of approximately 0.5 S/m was obtained by adding 2.5 g/l NaCl. Additionally, 1g/l Cu2SO4 was added to achieve reasonable T1 times. The phantom contains only minimum material for the mechanical fixation of the implant (implant holder) and has a coordinate system marked on the outside to ensure reproducible phantom placement within the scanner (Fig. 1). The implant holder is adjustable within the phantom and allows three different positions of the implants within the phantom (Fig. 1).
Figure 1:

The phantom consists of an acrylic glass canister A, closed by the lid C. The seal B ensures a water-tight closure of the phantom and prevents spills of the filling during positioning. D shows a scan through the phantom, showing the implant holder F. The implant holder F allows three different orientations of the device relative to the B0 field (G). Changes in the orientation were achieved by turning the implant holder within the phantom. The coordinate system on the phantom guided the repositioning.

Orientation 1 was defined by X: left-right, Y: foot-head, Z: bottom-up. Orientation 2 was defined by X: left-right, Y: top-down, Z: foot-head. Orientation 3 was defined by X: foot-head, Y: right-left, Z: bottom-up.

Implants

Representative active cardiac implants covering the application spectrum were selected for the measurements. The chosen implants were a cardiac loop recorder; a pacemaker; and two ICDs, one of which optimized for reduced susceptibility artifacts (Fig.2A). Due to their functionality, the implants not only differ by size but also by the amount of magnetic material.

The pacemaker and the ICDs were scanned without leads attached since the susceptibility artifacts caused by leads are negligible and were not an object of investigation.
Additionally, a generic setup mimicking an implant with distributed magnetic material was used to investigate the impact of the distribution on size and shape of the signal void. Magnetic material from ICD transformers (ferrite core material) was placed in different configurations in a piece of modelling clay having the dimensions of an ICD. Two placements were investigated in more detail, while using the same amount of magnetic material (Fig. 2B):

The magnetic material was all in the center of the setup - configuration “C” (central);

Half of the magnetic material was placed into a corner and the other half into the opposite one - configuration “D” (diagonal).

The piece of modeling clay was positioned in the phantom like a normal implant.
A) Active cardiac devices used for the measurements. All implants are manufactured by Biotronik, Germany. Dev. 1: Cardiac loop recorder (“Biomonitor 2”), Dev. 2: Pacemaker (Enticos 4 DR), Dev. 3: ICD (Ilesto 7 HF-T), Dev. 4: ICD (Activor 7 HF-T QP)

B) Generic setups: The magnetic material is either all placed in the center (configuration C) or with one half in one corner and the other half in the opposite corner to be at the maximum distance possible in a state-of-the-art ICD (configuration D).

Measurement setup

All measurements were performed on a clinical 3T MR scanner (AchievaDS, Philips Healthcare, Best, The Netherlands). For signal reception, commercial anterior and posterior body surface coils were used (dStream Whole-Body; Philips). For the measurements, the implant was positioned at the isocenter of the scanner. The pacemaker and the ICDs were positioned in the phantom with the device header pointing in direction of the Y-axis. The cardiac loop recorder was aligned on the implant holder diagonally along the line y=x. Since the distribution of magnetic material inside the implant relative to the B0 field orientation may affect artifact extent and shape, all measurements were performed in three orientations, each time aligning a different axis of the implant with the B0 field (Fig. 1).

In order to quantify the measurement uncertainties regarding the signal void measurements, all implants were scanned three times with both sequences using an identical setup. Repetition of any prescans was omitted in order to avoid a potential influence of pre-scans used for optimization of imaging contrast during the clinical routine 19,20.

MR protocol and sequences

MR imaging was performed using a number of sequences covering different needs. All sequences used a Cartesian acquisition scheme in transversal orientation.

A set of reference sequences was defined in close following of the standard for the evaluation of MRI image artifacts as described in the ASTM publication F2119-07 21. The reference sequences comprised both a multi-slice gradient echo and a multi-slice spin echo sequence with fixed values for TR, TE, receiver bandwidth, spatial resolution and flip angle. The reference gradient echo sequence deviated in its TR setting from the ASTM standard (reference sequence 50 ms, ASTM sequence 100-500 ms) allowing for a faster image acquisition. The other deviation was a higher matrix (256 x 256) than the
ASTM standard (256 x 128) for both sequences. Both factors have no influence on the artifact size of the active cardiac device. The exploration of the artifact size with the reference sequences had shown to be suboptimal because the long echo times and small bandwidth resulted in extremely large signal voids extending beyond the edge of the phantom particularly for the larger devices 3 and 4, both of them being ICDs. Furthermore, imaging times became unsuitably long when covering the entire phantom at an isotropic resolution.

In a second set of sequences, consisting of a 3D gradient echo sequence and a multi-slice 2D turbo spin echo sequence, we performed higher resolution MR imaging with optimized imaging parameters to allow for improved assessment of artifact size and type. Whereas a 3D approach was chosen for the gradient echo sequence, an interleaved multi-slice 2D sequence with adjacent slices (inter-slice gap = 0) was used for the TSE sequence in order to avoid an unreasonably long scan duration. Both sequence types allowed for the 3D visualization of artifact size and shape within the boundaries of the phantom.

The effects of the implantable devices on the main magnetic field (B0) and the flip angle (B1) were assessed by mapping both characteristics with dedicated sequences. For B0 mapping, a 3D gradient echo sequence was performed twice with different echo times (delta TE = 0.1 ms). For B1 mapping, a dual TR-sequence was applied (delta TR = 120 ms). The nominal flip angle was 60°. For both sequences, automatic map generation was performed on the MR system.

Quadratic field of view and coverage in orthogonal (z-) direction were set to 352 mm to cover the entire cubic phantom with some additional space about 20% larger than the inner volume of the phantom to avoid that structures of interest extended beyond the FOV (Fig. 1).

The main imaging parameters are listed in Table 1.

Tabl. 1: Main imaging parameters

|                        | Reference sequences | Optimized sequences |
|------------------------|---------------------|---------------------|
|                        | Grad echo           | Grad echo           |
|                        | Spin echo           | Spin echo           |
| Dimensions             | 2D MS               | 3D                  |
|                        | 2D MS               | 2D MS               |
|                        | 3D                  | 3D                  |
|                        | 3D                  | 3D                  |
| Image type             | FFE                 | FFE                 |
|                        | SE                  | TSE (5 echoes)      |
|                        |                     | FFE                 |
|                        |                     | FFE                 |
| TR (ms)                | 50                  | 20                  |
|                        | 500                 | 17560               |
|                        | 20                  | 30                  |
|                        | 17560               | 30                  |
|                        | 30 / 150            | 30 / 150            |
| TE (ms)                | 15                  | 3.2                 |
|                        | 20                  | 27                  |
|                        | 3.2                 | 3.1 / 3.2           |
|                        | 2.2                 | 2.2                 |
| **Echo spacing (ms)** | n/a | n/a | n/a | 9.1 | n/a | n/a |
|----------------------|-----|-----|-----|-----|-----|-----|
| **Flip angle (deg)** | 30  | 90  | 20  | 90  | 60  | 60  |
| **Flow comp**        | No  | No  | Yes | yes | No  | No  |
| **Field of view (mm)** | 352 | 352 | 352 | 352 | 352 | 352 |
| **Acq. Matrix**      | 256 | 256 | 176 | 176 | 88  | 88  |
| **Nr of slices**     | 60  | 60  | 176 | 176 | 88  | 88  |
| **Spat. Resol. (acq) (mm3)** | 1.38 x 1.38 x 5 | 1.38 x 1.38 x 5 | 2 x 2 x 2 | 2 x 2 x 2 | 4 x 4 x 8 | 4 x 4 x 8 |
| **BW (Hz/pix)**      | 125.2 | 125.2 | 382.4 | 473.5 | 498.4 | 498.4 |
| **Acq. Dur. (mm:ss)** | 6:28 | 17:12 | 10:24 | 21:04 | 3:52 | 11:35 |
| **Spat. Resol. (recon) (mm3)** | 1.38 x 1.38 x 5 | 1.38 x 1.38 x 5 | 1 x 1 x 1 | 1 x 1 x 2 | 2 x 2 x 4 | 2 x 2 x 4 |
| **FoV dir**          | RL / AP | RL / AP | RL | RL | RL | RL |
| **Fat shift direction** | P / L | P / L | P | P | P | P |
| **Orientation**      | Tra | Tra | tra | tra | Tra | Tra |
| **Max B1_rms (uT)**  | 0.77 | 1.60 | 0.89 | 1.62 | 1.27 | 0.73 |
| **SAR level (W/kg)** | <0.3 | <1.3 | < 0.4W/kg | < 1.3 | <0.8 | <0.3 |
| **Db/dt (T/s)**      | 47.6 | 33  | 50.4 | 38.9 | 23.4 | 50.3 |

**Image Processing**

The image data were stored in DICOM format. All subsequent processing was performed with custom software scripted in MATLAB (R2019b, Mathworks, USA). First the data were converted to 3D matrices, resampled and if needed an interpolated to obtain indentically sized voxel of 2x2x2mm voxel resolution for all data sets. A brightness correction was performed by histogram stretching to compensate for differences in image brightness and to be able to apply identical gray thresholds for all data.
Evaluation of the signal void

Data segmentation

A low pass filter was applied first to eliminate the structures of the implant holder and was realized by an averaging filter with a convolution kernel of 9x9x9 voxels. Segmentation of the signal void was done by a threshold operation applied to the gray value profile of each digital image data. Along the gray value profile, darker regions were encoded with values near “0” and bright regions with values near “1”. The transition between the two extreme values has a finite and in general sigmoidly shaped slope. The steepest slope is at 50% of the sigmoid curve, thus offering the most exact cut-off between signal void and “normal” signal intensity. Accordingly, the signal void was extracted as the dark part below a threshold of 50%. To measure the actual volume of the signal void the voxel volumes meeting the aforementioned conditions were summed up.

Metrics

To quantify the signal void two metrics were regarded: the actual volume of the signal void and the extent of the signal void’s bounding box. The bounding box was determined by minimizing its volume, solving an optimization problem using MATLAB’s fminsearch algorithm with the six roto-translation degrees of freedom in order to calculate the volume.

B0 and B1 maps

B0 and B1 maps were analyzed by drawing circular regions of interest (ROIs; area 3cm²) near the corners (17 cm from the device) and edges (12 cm from the device) of the phantom (26 ROIs per data set), as well as around the signal void (4-12 cm from the device). Measurements with a device present were compared with those obtained from a phantom without device.

Results

All four devices were measured with the reference sequences, the spin echo (2D-TSE) and the gradient echo (3D-TSE) sequence in all three orientations and with two repetitions.

The main influencing factors on shape and volume of the signal void are the chosen sequence type and the implant-type itself (Fig. 3).
Figure 3:
Segmented signal voids induced by the implants Dev1-4 scanned with the 2-DTSE and the 3D-TFE sequences in orientation 1. The red box in each coordinate system represents the boundary box. The absolute values of the boundary box are given below the coordinate system. Dev 1: cardiac loop recorder. Dev 2: Pacemaker, Dev 3 and 4: ICDs.

In general, gradient echo-based sequences provoke larger signal void volumes than spin echo-based sequences. Compared to the ASTM reference sequences, the optimized 3D-TFE sequence caused signal voids smaller by at least a factor of four for all four tested implants. These smaller signal voids do not exceed the boundaries given by the phantom size, thus allowing the analysis of the three-dimensional signal void shape (Fig. 4). Similarly, the optimized 2D-TSE resulted in smaller signal void volumes in comparison with the reference sequences. However, the reduction in the signal void caused by the 2D-TSE depends on the size of the signal voids in the reference sequences. Whereas small artifacts are not significantly reduced, larger artifacts in the reference sequence are reduced by a factor of two when the 2D-TSE sequence is used. Additionally, the shape of the signal void is predominantly influenced by the direction of the B0 field, whereas the orientation of the implant within the isocenter has little impact. Fehler! Verweisquelle konnte nicht gefunden werden.
Figure 4: Morphology of the signal voids produced by the 2D-TSE and 3D-TFE sequence as compared to reference sequences, measured in orientation 1 and with Dev 4.

When the ICDs (Dev 3 and Dev 4) were measured in orientation 2, the 3D-TFE sequence caused signal voids up to 10% larger than when measured in the two other orientations (Fig. 5).
Figure 5: Comparison of signal void volumes depending on the scan sequence and the device (phantom) orientation. Dev1: cardiac loop recorder, Dev2: pacemaker, Dev3 and Dev4: ICDs.

The measurements with the generic setups showed that the positioning of the magnetic material within the device itself influences the artifact extent. The two generic setups showed similar signal void shapes regardless of the positioning of the magnetic material. Each magnetic element caused a separate but identically shaped signal void. Depending on the relative positioning of each magnetic element these twin-like structures overlap but still were be visually distinguished (Fig. 6). Interestingly, the total volume of signal void caused by the twins (configuration D) is less than the volume of the signal void in configuration C. This applies for both the spin-echo and the gradient echo sequences. However, the bounding box dimensions for the configuration D are larger as compared to the configuration C (Fig.2).
Figure 6: Generic setups and their artifacts. C describes a centric position of all magnetic material, D the splitting of the magnetic material into two diagonally positioned halves. All scans were performed in orientation 1. The red boxes mark the bounding box dimensions. The absolute values of the boundary box are given below the coordinate system.

Changes in B0 were more pronounced with larger devices. Near the corners or edges of the phantom, there was no major noticeable shift in B0 for devices 1 and 2. Closer to the devices, shifts of -580 Hz to +260 Hz and -760 to +435 Hz were measured for device 1 and 2, respectively (Fig.7).
Figure 7: B0 (upper row) and B1 (lower row) maps of a central slice within the cubic phantom. The left image of each row shows the empty phantom. The scales range from -5000 Hz to +5000 Hz for B0 and from 50% to 150% relative flip angle for B1 (with 100% indicating the nominal flip angle). The disturbances in the B1 field in the empty phantom are a standing wave effect because the edge length of the phantom is close to the magnetic wave length at 3.0 T. Accordingly, disturbances in B1 in the presence of an active device are measured relatively to the empty scan.

For devices 3 and 4, there were considerable shifts in B0 even close to the edges of the phantom and even when no signal void or other artifact could be seen in the gradient and spin echo images. Maximum B0 differences were as much as +1160 Hz for device 3 and as much as +1440 Hz for device 4. These maximum offsets occurred for orientation 2 in z-direction, i.e., in the direction along the bore. Closer to the devices, offsets were even larger: -1280 Hz to +1420 Hz for device 3, and -1330 Hz to +1220 Hz for device 4.

B1 measurements showed very little variance for devices 1, 2, and 4. Whereas there was a small trend towards smaller B1 values particularly in the front slices (farthest from the front opening, “cranial”), differences were much smaller than the standard deviation over the entire phantom and not significant. For device 3, however, there was a strong trend towards smaller B1 field in cranial direction, with the areas in the corner showing a B1 field reduced by up to 35% compared to the measurements in absence of a device. On the other hand, none of devices led to the generation of any hot spot: the maximum B1 value found was 107% of the nominal flip angle in the absence of a device, and 108% in the presence of a device (Fig. 7).

Uncertainty assessment

In the chosen setup without repeating the prescans the repeat measurements for both spin echo and gradient echo sequences showed a very precise reproducibility of the presented results with less than 2% deviations from the mean value for the spin echo sequence and less than 4% for the gradient echo sequence. Exemplarily, for Dev 4, the signal void caused by the gradient echo ranged between 1551 mm³ and 1596 mm³ and between 674 mm³ and 686 mm³ for the spin echo sequence.

Discussion

The first step towards optimizing CMR image quality in the presence of active cardiac implants at 3T is the systematically performed three-dimensional analysis of the artifact extend and influencing factors. The different qualities of susceptibility induced artifacts, signal void and signal distortion, demand
different settings for a systematic evaluation. Therefore, this work focuses on the former, using a cubic phantom with homogeneous filling and of sufficiently large volume to cover sufficiently the signal void.

When evaluating passive and active implants in the surrounding of MRI, the ASTM standard had been established for clinically established field strengths \(^{21}\). However, originally designed for measurements at 1.5T, the ASTM standard proved to be unsuitable for the purpose of this work, where the effects at 3T are investigated and considerably larger artifacts occur than at 1.5T. According to the ASTM standard, the artifact itself is defined as a deviation in the signal intensity of at least 30% from a reference scan without present device. However, this approach requires a perfectly matched pair of reference and artifact scans without the influence of repeated prescans after repositioning. In a clinical operation mode, it is not possible to omit these prescans. Additionally, it is not possible to apply the ASTM standard definition to an in vivo application in patients with already implanted devices due to the lack of a reference scan. In contrast the approach chosen for the presented work uses the grey profile of each image data set, thus omitting the necessity of reference scans. The sequence protocols of the ASTM standard are set to maximize the artifact size, causing signal voids that exceed the boundaries of the phantom thus limiting the analysis of the signal void shape. The sequences optimized for our phantom, though inducing overall smaller signal voids, allowed for an image quality and spatial resolution appropriate for evaluation with high confidence at 3T. The benefit of this optimized approach is evident when focusing on devices with larger amounts of electrical components, e.g. the ICDs (Dev. 3 and 4).

The differences between Dev3 and Dev4 highlight the role of the magnetic components within the active cardiac devices. The artifact sizes differ significantly all the while both devices offer similar technical features. The measurements on the generic setups demonstrate the influence of the device design on the artifact volume based on the distribution of the electric components within the device. These findings showed the potential that lies within the modification of the device setup with regards to the modification of artifact burden. Although the design of the implants is not within the influence of the CMR user, information on the behavior of a specific device might help to plan the CMR examination.

\(B_0\) measurements showed the expected increase in \(B_0\) shift with larger devices. It is notable that, even outside of the signal void, considerable \(B_0\) shifts may be present. This has a direct effect on any MR measurement employing a frequency-depending pulse, such as fat suppression or an inversion pulse. This pulse type is used for instance in black blood imaging or LGE. If the preparatory pulse bandwidth is chosen too narrow, it may result in insufficient magnetization preparation and lead to image artifacts. \(B_1\) was found to be less affected by the presence of a device, and if so, mainly in the corners of the phantom, i.e., far away from the isocenter and thus most likely not in the main area of interest.
However, it must be kept in mind that in these regions, B1 may be reduced and thus the actual flip angle may be smaller than the nominal flip angle. Again, this is expected to affect features such as fat suppression or inversion pulses. The absence of B1 hot spots supports the assumption that, also in the presence of a device, no unwanted heating occurs provided that the restrictions prescribed by the MR conditional device are observed. It must be kept in mind though that our measurements were performed on devices that were not switched on and were not connected to leads. It can be concluded that the orientation of the implant relative to the B0 field has a minor effect on the size and the shape of the signal void that in practice most likely can be neglected. The outcome of the current study is still limited as we obtained data from only four implants. Although these are spanning a broad field of applications, they originate all from the same manufacturer. However, the main conclusions are considered of general validity. The quantification of other artifact qualities, such as distortion require a different measurement set-up, and thus will be analysed separately.

All used implants have a Titanium housing that produces rather negligible susceptibility artifacts itself. The eddy currents induced in these housings by the gradient system, however, produce additional artifacts due to the magnetic field related to these currents. These are superimposed on the susceptibility artifacts and are not separated on this study.

Conclusion

The presented work establishes an analysis routine for a systematic analysis of device related signal void quantification that can be transferred to application in vivo. The results not only highlight the effect of the sequence type on the artifact extend, but also the relevance of the device type and design on the artifact burden.

Conflict of interest:

Dr. Weiss is an employee of BIOTRONIK SE & Co. KG, Berlin, Germany, and Dr. Weber is an employee of Philips GmbH, Hamburg, Germany. Prof. Bauer is a scientific advisor for BIOTRONIK SE & Co. KG, Berlin, Germany. Also, this work has been partially funded by BIOTRONIK SE & Co. KG, Berlin, Germany.

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