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A novel and rapid approach to estimate patient-specific distortions based on mDIXON MRI

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Abstract

While MRI-only radiation treatment planning (RTP) is becoming more widespread, a robust clinical solution for patient-specific distortion corrections is not available. This work explores B_0 mapping based on mDIXON imaging, often performed for MR-only RTP, as an alternative to separate dual-acquisition gradient-recalled echo (GRE) imaging, with the overarching goal of developing an efficient and robust approach for patient-specific distortion correction. Initial benchmarking was conducted by scanning a phantom and generating B_0 field maps with two approaches: (1) conventional B_0 mapping and (2) experimental mDIXON imaging. Distortion maps were derived from the field maps and compared. The head and neck regions, including brain, of 10 healthy volunteers were then evaluated at 1.5 T and 3 T. Distortion maps were again compared between approaches, using difference maps and histogram analysis. Overall, conventional B_0 mapping was well approximated by mDIXON imaging: The distortions of 95% of the voxels in the phantom estimated by mDIXON and conventional B_0 mapping differed by <0.02 mm (1.5 T) and <0.04 mm (3 T), while the 95-percentiles of the distortions estimated by conventional B_0 mapping were <0.06 mm (1.5T) and <0.12 mm (3T). In head and neck the distortions of 99% of the voxels were within ± 0.2 mm at 1.5 T for both approaches and within ± 0.4 mm and ± 0.5 mm at 3 T for mDIXON imaging and conventional B_0 mapping, respectively. The majority of differences in-vivo were confined to regions with high spatial variation of the B_0 field, mostly around internal air cavities. For 1.5 T, the mDIXON imaging-based correction alone reduced the 95-percentile of distortions from 0.15 mm to 0.03 mm and within the brain from 0.06 mm to 0.02 mm. Slightly lower reductions were observed at 3 T. In conclusion, mDIXON imaging closely approximated conventional B_0 mapping for patient-specific distortion assessment. Estimates in the brain were in good agreement, and slight differences were observed near air/tissue interfaces in the head and neck. Overall, mDIXON imaging-based B_0 field maps may be advantageous for rapid patient-specific distortion correction without additional imaging.

Keywords

Distortion; MR-only radiation treatment planning; MRI

1. Introduction

In MR-only-based radiation therapy planning (RTP), geometrically correct MR imaging is essential, because target tumor and organs-at-risk are delineated on MR images, and dose planning and patient positioning for therapy are performed based on these delineations. There is a strong interest to move from the CT-MR-based workflow to the MR-only variant to eliminate redundant CT scans, and more importantly to avoid geometric uncertainties during CT-MR registration, which can be in the order of 2 mm for brain [1,2], 3 mm for prostate, and 5 mm for liver [3]. Geometric accuracy of MR images is limited by various types of distortions. System-level distortions arise from gradient nonlinearity [4,5] and B_0 field inhomogeneities inherent to the scanner. Both effects are largely corrected during image reconstruction based on per-system reference data [6,7]. However, chemical shift of fat and magnetic susceptibility effects cause distortions in readout direction for sequences with Cartesian k-space acquisition [5] that depend on the patient and thus typically require additional data acquisition to perform a correction. Spatial variations in susceptibility, most pronounced at interfaces with large susceptibility differences such as air/tissue or bone/metal, cause B_0 field inhomogeneities up to about 4ppm in the human head, which can cause distortions of up to 4 mm at the sinus/tissue interface in the brain when scanning with only moderate bandwidth [8]. Chemical shift and patient-specific distortions due to variations in susceptibility are commonly addressed by scanning at high read-out bandwidth. In addition, B_0 mapping with a dual-acquisition gradient-recalled echo (GRE) sequence is increasingly used to quantify and correct for such distortions that are related to B_0 field inhomogeneities (in this work called “distortions”), but this sequence requires additional scan time and image processing steps such as phase unwrapping to apply the corrections.

This work proposes a novel and rapid approach to the assessment of these types of distortions by using a B_0 field map derived from mDIXON imaging. Using mDIXON imaging is favorable since it is commonly performed for RTP (i.e., for synthetic CT generation [9,10,11,12]) while enabling inherent simultaneous acquisition of B_0 field maps and RTP data, which minimizes the risk of misalignments due to patient motion.

2. Methods and Materials

2A. Phantom scan methods

For initial benchmarking, a 27 cm fluid-filled sphere was first scanned with a conventional dual-acquisition GRE sequence that subsequently acquired each k-space profile twice at two different echo times. This conventional B_0 mapping is considered the “standard” approach for patient-specific distortion assessment [4, 8, 13]. Detailed acquisition parameters are summarized in Table 1 for all sequences. The GRE sequence was optimized to estimate the B_0 distribution as closely as possible in the given imaging time based on the following considerations: the first two in-phase echo times of water and fat were selected to avoid any

phase errors due to the chemical shift of fat for subsequent in-vivo scans, radiofrequency spoiling was applied to suppress any previous coherences for optimum image quality, and a flip angle of 12 deg was used to balance the signal levels of water and fat.

For the experimental approach images were acquired with a T1-weighted bipolar gradient-recalled dual-echo mDIXON sequence in the sagittal orientation to facilitate shorter scan times [14]. This sequence is applied clinically to derive synthetic CTs (MR for Calculating Attenuation (MRCAT), Philips Healthcare [11]) in conjunction with a model-based segmentation [15]. While classical DIXON sequences use in-phase and out-of-phase echo times, the modified DIXON (mDIXON) sequence applied here uses shorter echo times that are automatically chosen by the scanner to optimize for minimal scan time while still ensuring a robust water-fat separation [14]. All parameters were kept as close as possible to the clinical protocol defined for MRCAT [11].

The T2-weighted turbo spin echo (T2-w TSE) sequence (turbo factor = 70, acquisition matrix = 212 x 343 x 160) was used to acquire high resolution axial images using the same field of view (FOV).

All data were acquired on both a 1.5T and a 3T MR system (Ingenia, Philips Healthcare, Best, The Netherlands). High bandwidths were used for all scans in this study to minimize off-resonance-induced distortion, fat shift and scan time. The bandwidth of conventional B₀ mapping and mDIXON imaging for 3T was doubled vs 1.5T to obtain a fat shift of 0.5 pixels at both field strengths as required by the MRCAT solution of synthetic CT generation [11]. T2-w scanning was performed with maximal bandwidth. The vendor-provided 3D gradient-nonlinearity-correction was used for all scans.

A combination of a 16-element anterior coil array, a 12-element posterior coil built into the patient support, and two circular coils positioned left and right of the phantom was used for all scans. Geometrical parameters for the mDIXON sequence including the field of view, pixel sizes, encoding directions, and shim settings were chosen as for the conventional GRE sequence, except for the slice thickness, which was increased for conventional B₀ mapping to reduce scan time. Compressed sensing (CS factor = 3) was applied to all imaging to keep in-vivo scan times of the mDIXON and B₀ mapping sequences below 2 min to minimize motion between scans.

2B. Volunteer scan methods

Ten healthy male volunteers (age 47.8 ± 7.9 (range: 30 to 62 years) were scanned after consenting to a local ethics approved study for head and neck imaging at both 1.5T and 3T MR systems (Ingenia, Philips Healthcare, Best, The Netherlands). A combination of a 16-element anterior coil array, a 12-element posterior coil built into the patient support, and two circular coils positioned on the left and right of the head and neck was used similarly to the recommended set-up for clinical RTP (Figure 1) [12]. Volunteers were immobilized with straps across the upper head and below the chin and instructed to keep still to avoid motion of the mouth and tongue region.

2B. Phantom and Patient Data Processing

The in-phase echo times for conventional B_0 mapping resulted in field maps with frequency ranges of ± 108.5 Hz for 1.5T and ± 217 Hz for 3T. Since off-resonances up to about twice this range were observed, phase unwrapping was performed [16]. Distortion maps for the respective T2-w scans were calculated from the unwrapped field maps $f(x)$ obtained from B_0 mapping or mDIXON imaging according to

$$\Delta x = \frac{\Delta x_{acq}}{BW_{acq}} (\Delta f(x) + f_0^{map} - f_0^{T2}) \quad (\text{Equation 1})$$

where x_{acq} and BW_{acq} are acquisition pixel size in mm and acquisition pixel bandwidth in Hz of the T2-w scan in readout direction, respectively, f_0^{T2} is the resonance frequency in Hz of the T2-w scan, and f_0^{map} is the resonance frequency of either B_0 mapping or mDIXON imaging. The two f_0 terms were included because the system software enforces updates of the resonance frequency for each scan within the exam. The sum $f(x) + f_0^{map}$ is the actual frequency distribution observed during mapping, and its difference to f_0^{T2} accounts for any distortions that will be observed in the T2-w scan given this distribution.

Distortion maps obtained from mDIXON imaging, conventional B_0 mapping, and voxel-wise difference maps were calculated for each phantom and volunteer scan. For all cases, conventional B_0 mapping is regarded as the reference as it is the established methodology to recover the field map required for distortion assessment [4, 8, 13]; hence the distortion maps acquired with conventional B_0 mapping are considered as ground truth for this evaluation. To assess the overall agreement between the mDIXON and B_0 mapping techniques, cumulative histograms of the absolute values of the distortion maps were calculated which combine data from all ten volunteers per field strength in the in-vivo cases. To assess the residual distortion remaining after distortion correction using the mDIXON-based field maps, cumulative histograms of the absolute values of the distortion difference maps were calculated. This is based on the assumption that B_0 mapping is regarded as the reference for recovering the field map. To quantify differences in distortion magnitude between both approaches in the brain region, a fully automatic model-based segmentation algorithm was used to segment the brain [17], and cumulative population-based distortion histograms were calculated. Background noise outside of the anatomy and the shoulder regions were excluded from all histogram calculations.

3. Results

3A. Phantom Study

Distortions estimated for the T2-w scan based on the conventional (B_0 mapping) and experimental (mDIXON) approaches, including difference maps, are shown in Figure 2. Overall, similar distortion patterns are observed across the phantom volumes for both techniques. The main component of the distortion difference between mDIXON imaging and B_0 mapping is a spatial gradient along the readout direction (vertical direction in Figure 2).

Cumulative histograms of the absolute distortion as measured with the different approaches are depicted in Figure 3a. Distortions calculated at 1.5 T were lower than 0.06 mm in 95% of all voxels for both approaches. At 3 T, the 95-percentiles were 0.11 mm and 0.12 mm for the experimental and conventional approaches, respectively. Figure 3b displays the cumulative residual distortions after either no correction at all or after correction using only the mDIXON approach. The 95- and 99-percentiles for residual distortions after such corrections are displayed in Figure 3c for quantitative comparison. Applying the mDIXON correction reduced the 95-percentile of the distortion distribution from 0.06 to 0.02 mm at 1.5 T and from 0.12 to 0.04 mm at 3 T.

3B. Volunteer Study

As shown in Figures 4 and 5 depicting two different volunteers each at both field strengths, the typical spatial distributions of distortions calculated for the T2-w imaging scan based on either mDIXON imaging or B_0 mapping were qualitatively very similar for volunteer data. As expected, distortions were largest at air-tissue interfaces near the shoulders, the neck, the chin, around the major air cavities in the head, and at the tip of the lungs for both approaches. The distortion difference between mDIXON imaging and B_0 mapping is globally much lower than the total distortion estimated by B_0 mapping, indicated by ratios of 95-percentiles of distortion difference to total distortion by B_0 mapping of 22% for 1.5 T and 29% for 3 T. Almost all differences larger than 50% observed between the two approaches occur locally in regions with high spatial variation of the B_0 distribution, mostly around nasal, oral, and ear cavities, the sphenoid sinus, and dental fillings as observed in the volunteer displayed in Figure 5 (left columns for each field strength).

Cumulative histograms for distortions based on both mapping approaches for the full FOV except the brain and for the brain only are depicted in Figure 6 (a, d). Distortions in full FOV except brain were lower than 0.17 mm in 95% of all voxels for both approaches at 1.5 T. In the brain, these 95-percentiles were 0.054 mm and 0.061 mm for the conventional and the mDIXON approach, respectively. At 3 T, the observed respective 95-percentiles were approximately doubled. Figure 6 (b, e) displays the cumulative residual distortion after correction with the mDIXON approach only in comparison to distortions without any correction. For 1.5 T the mDIXON correction alone reduced the 95-percentile of distortion in the full FOV except brain from 0.17 mm to 0.04 mm (76% reduction). The respective reductions for brain only were from 0.06 mm to 0.02 mm (66% reduction). Reductions for 3 T were from 0.37 mm to 0.12 mm (68% reduction), and from 0.1 mm to 0.05 mm (49% reduction) in brain. All 99-percentiles for these cases are displayed in Figure 6 (c, f).

4. Discussion

This study sought to assess the feasibility and accuracy of using an mDIXON-based field map for distortion assessment as an alternative approach to conventional B_0 mapping with the overarching goal of reducing total exam time for MR-only RTP. Evaluating both approaches in a phantom offered the opportunity to benchmark the mDIXON approach while eliminating motion and co-registration errors. The distortion difference between both mapping approaches as depicted in Figure 2 is dominated by a spatial gradient in the readout

direction. This can be explained by the different eddy currents for the mDIXON readout and the B_0 mapping readout [14]. The bipolar mDIXON readout is optimized for short scan times, whereas the unipolar B_0 mapping readout minimizes the errors in the measurement of the B_0 distribution. A large fraction of the difference is compensated by a 1-dimensional eddy current compensation method included in the mDIXON reconstruction [14]. However, residual phase errors again have a spatial distribution that is dominated by the read-out gradient field itself.

In human subjects, overall, the mDIXON approach provided an estimate of distortions comparable to the conventional B_0 mapping reference method even in regions with larger off-resonances e.g. at the chin, the neck, and large parts of the shoulders. Major distortion differences between both approaches occur mainly where the spatial gradient of the distortion is large such as near the sinus cavities or ear canals, whereas no constant spatial gradients along the read-out directions were noticeable as in the phantom. The potential cause for the increased differences between both approaches at locations with locally increased gradients of the B_0 distribution is that the mDIXON reconstruction essentially smoothes the field map. This is required to increase the robustness against water-fat swaps in the water-fat-separation as needed by MRCAT generation. A potential solution to this would be to increase scan resolution at the expense of longer scan time or to perform a triple-echo mDIXON sequence allowing for robust water-fat separation and high-fidelity B_0 mapping. While this solution would increase scan time, it may also be considered to derive an additional confidence map based on the magnitude of the spatial gradient of the dual-echo mDIXON-based B_0 map, which may be subject of future work.

The cumulative distortion histograms including the phantom data in Figure 3 (a) and all volunteer data in Figure 6 (a, d) directly compare the distortions as measured by mDIXON-based mapping and B_0 mapping. However, to determine from a clinical perspective whether the mDIXON-based mapping alone would be sufficient for distortion estimation, it is required to compare the residual distortions *after* applying the mDIXON-based correction to the distortions without any correction. This comparison is given in Figure 3 (b, c) and Figure 6 (b, c, e, f). For these histograms, B_0 mapping was considered to be the reference, assuming that it is the best estimate of the off-resonance distribution. The graphs “no corr” are based on the B_0 mapping data, because B_0 mapping is assumed to represent the distortions if no correction is applied. The graphs “corr by mDIXON” display the histograms of the differences between mDIXON and B_0 mapping, because after correction with the mDIXON-based data, this difference represents the residual distortion. The percentiles in the bar charts are displayed to visualize the data points for 95% and 99% of the histograms for a more insightful comparison. The histograms and bar charts reveal that the mDIXON approach corrects for a large amount of the overall distortions (e.g. 95-percentile of distortion reduced by 78% in the full FOV at 1.5T) without a dedicated sequence or additional scan time. While this result was obtained for the head and neck region, here chosen for a valid comparison as discussed below, it is expected that similar reductions can be obtained in other body parts.

In this study, readout bandwidth selection for the T2-w scans was set unusually high (820 Hz/mm at 1.5T, 713 Hz/mm at 3T) in order to reduce overall scan time. Clinical T2-w scans used for delineation of organs-at-risk typically require higher SNR and therefore use rather

lower bandwidths. Two recent papers on aspects of distortions in clinical RTP have used lower bandwidths in the order of ~135–395 Hz/mm [12, 13]. If these parameters were considered, distortions, scales of histograms and percentiles presented in this work would increase by a factor of two to six according to Equation 1. Nevertheless, the overall magnitude of distortions has been reported to be quite low in the brain and head and neck regions [8]. In this study only the distortions for the T2-w scans were calculated although distortion assessment for other sequences including the mDIXON scan itself can be conducted using appropriate parameters. Self-correction of the mDIXON scan in lieu of B_0 mapping avoids any inter-scan motion between the acquisition of the field map and mDIXON imaging, reduces the time between the acquisition of the field map and any subsequent scan, and hence reduces the probability of inter-scan motion.

A limitation of this study is that it was performed on head and neck and brain regions of volunteers and not patients with pathology. The head and neck is of particular interest because it offers a valid comparison between the two B_0 mapping approaches due to relatively limited inter-scan motion, presence of internal air cavities generating interface effects, and the ability to evaluate distortions for both brain and head and neck disease sites.

Calculations of distortion maps were performed offline for this proof of concept study. To implement mDIXON-based distortion assessment and correction in clinical routine, calculations according to equation 1 and the respective distortion corrections [4, 5] could be integrated into the MR reconstruction software.

5. Conclusions

Our work shows that a large fraction of patient-specific distortions can be corrected with an mDIXON-based field map only. Because this field map is estimated within the mDIXON reconstruction for the purpose of water-fat separation as required in CT-simulation, the respective correction saves the scan time for a dedicated field map, which is typically more than for the mDIXON scan itself. While this study was performed in volunteers for the head and neck and brain regions, the overall technique is expected to translate to patients and other anatomical sites.

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Figure 1:

(a) Volunteers were immobilized with straps below the chin and across the upper head. (b) A combination of a 16-element anterior coil array, a 12-element posterior coil inside the patient support, and two circular coils positioned left and right of the neck was used.

Figure 2: Distortions for the T2-w scan for a subset of slices covering the entire phantom as estimated from the experimental mDIXON approach and conventional B₀ mapping at 1.5 T and 3 T, along with the difference between these maps. Note that the ranges of the distortion color scale vary with field strength and are halved for the difference maps.

Figure 3:

(a) Cumulative histograms of the distortions in the phantom as estimated by the experimental mDIXON and conventional B₀ mapping approaches. (b) Cumulative histograms of residual distortions after using either no correction or the mDIXON-based correction (linked by arrows), assuming that all distortions can be fully corrected using conventional B₀ mapping. (c) Statistical results of residual distortions for phantom data.

Figure 4:

Top two rows: Distortions calculated for the T2-w scan parameters from experimental mDIXON and conventional B_0 mapping. Third row: Differences between distortions estimated from mDIXON and conventional B_0 mapping. Bottom row: mDIXON in-phase images shown for anatomical reference. All images show corresponding sagittal slices for the same volunteer at both 1.5 T and 3 T.

Figure 5:

Top two rows: Distortions calculated for the T2-w scan parameters from experimental mDIXON and conventional B₀ mapping. Third row: Differences between distortions estimated from mDIXON and conventional B₀ mapping. Bottom row: mDIXON in-phase images shown for anatomical reference. All images show corresponding axial slices for the same volunteer at both 1.5 T and 3 T. The volunteer is different from the one in Figure 4.

Figure 6:

(a) Cumulative histograms for distortions in volunteers calculated by conventional field maps and mDIXON imaging considering the full field of view (FOV) except the brain voxels. (b) Cumulative histograms of residual distortions after using either no correction or the mDIXON-based correction (linked by arrows), assuming that all distortions can be fully corrected using conventional B₀ mapping. (c) Selected percentiles of residual distortions for (b). Lower row: Respective histograms considering only brain voxels. All histograms include data from all scanned volunteers. Note the scale difference between the full FOV and brain results.

Table 1:

Acquisition parameters for MRI sequences used for phantom and healthy volunteer scans. The scan resolution in readout direction is marked with *.

Field Strength (T)	Sequence	Acquisition $V_{FH}/V_{AP}/V_{RL}$ (mm)	Bandwidth (Hz/pixel)	α (deg)	TR (ms)	TE ₁ (ms)	TE ₂ (ms)	T _{acq} (min:s)
1.5	mDIXON	1.2*/1.2 /1.2	426	10	7.7	2.2	4.9	01:46
	B0-map	1.2*/1.2 /3.4	432	12	13.0	4.6	9.2	01:59
	T2	2.0/1.2/1.2*	983	90	1250	135	-	05:52
3.0	mDIXON	1.2*/1.2/1.2	848	10	5.0	1.5	2.9	01:08
	B0-map	1.2*/1.2/3.4	842	12	10	4.6	6.9	01:37
	T2	2.0/1.2/1.2*	855	90	1250	139	-	05:51