Improved T1 Mapping and DCE-MRI Quantification for Prostate at 3T by incorporating B1 Inhomogeneity Correction

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Introduction

Dynamic Contrast Enhanced (DCE) MRI is used for the assessment of tumor vascular properties with application to prostate cancer detection, characterization, and treatment monitoring [1]. MR signal intensity changes versus time during the uptake of Gad-DTPA are measured and used in conjunction with pharmacokinetic (PK) models to provide a number of tumor vascular properties. An initial step in the PK analysis requires conversion of signal intensity vs. time into contrast-agent concentration (C(t)) vs. time. Since signal intensity changes are non-linearly related to contrast agent concentration, this requires knowledge of pre-contrast tissue T1 values. Variable Flip Angle (VFA) imaging is a preferred T1 mapping method since it provides T1 maps using the same 3D SPGR sequences that are used for DCE acquisition, so that identical spatial resolution and coverage can be obtained in reasonable acquisition times. VFA analysis fits the imaging equation as a function of flip angle \( \alpha \) to obtain pixel-wise T1 values. However, VFA suffers from large errors at higher field strengths due to B1 field inhomogeneity and applied flip angles (\( \alpha_{\text{applied}} \)) differing from the actual flip angles (\( \alpha_{\text{actual}} \) seen by the tissue with \( K = \alpha_{\text{actual}} / \alpha_{\text{applied}} \)) varying spatially within the imaged volume. In this work, we demonstrate the application of the Bloch-Siegert-based B1 estimation method [2], a validated method for measuring \( \alpha_{\text{actual}} \) to correct the VFA curves and the DCE curves, thus obtaining improved T1 maps and PK values. Such a corrected method has promise in improving DCE-MRI analysis and providing consistent results allowing improved cancer detection and characterization.

Methods

26 subjects were scanned on a GE (Waukesha, WI) 3.0T Twinspeed HDx system after obtaining informed consent under IRB approved protocols.

**Acquisition:** (A) VFA protocol: 3D FSPGR, SPECIAL fat-suppression, TR 15ms or TR 9.1ms, TE 3.1ms. FA 21/18/15/12/9/6 degrees. 14-16 slices. Slice thickness 6mm. Matrix 128x256. FOV 26x26 cm\(^2\), BW ±15.6Khz. (B) Bloch-Siegert (BS) protocol: 2D SE, TR/TE/Flip 950ms/22ms/90 degrees, 2 KHz off resonance Bloch-Siegert B1 pulse [2], Matrix 128x128, FOV 30x30cm\(^2\), BW ±31.3Khz. Slice thickness 6mm. Acquisition time 4min/16 slices. (C) DCE protocol: 3D FSPGR, SPECIAL fat-suppression, TR/TE/FA 4.0/1.4ms/15 degrees. 14-16 slices, 160x256. FOV 26x26cm\(^2\), BW ±64Khz, Slice Thickness 6mm, imaged every 5 sec for 6 minutes after administration of 0.1 mmol/kg GD-DPTA i.v. at 0.3 cc/sec.

**Analysis:** (A) B1 mapping: B1 spatial re-sampling was performed to match the DCE and T1 images. B1 maps in units of \( \mu \text{T} \) were divided by the nominal applied B1 of 5000 \( \mu \text{T} \) to yield maps of K, which were truncated to retain only values of 0.5 < K < 1.5 to exclude holes created by brachytherapy needles or prior biopsies. (B) T1 mapping: The SPGR equation SI(\( \alpha \)) = M0\( \sin \alpha \) * (1 - exp(-TR/T1)) / (1 - cos \( \alpha \) * exp(-TR/T1)) was fitted to the signal versus \( \alpha \) curve. B1 correction was incorporated by correcting the flip angles using \( \alpha_{\text{actual}} = \alpha_{\text{applied}} * K \) on a pixel-wise basis. (C) DCE Signal Correction: The ratio of signal intensity at time t, SI(t), to signal intensity at baseline SI\(_{\text{pre}}\) is SI\(_{\text{pre}}\) / SI(t) = [1-exp(-TR/T1\(_{\text{pre}}\))] / [1-cos \( \alpha \) exp(-TR/T1\(_{\text{pre}}\))] * [1-cos \( \alpha \) exp(-TR/T1\(_{\text{t}}\))] / [1-exp(-TR/T1\(_{\text{t}}\))], where we utilized the corrected \( \alpha_{\text{actual}} \). The resulting T1(t) was then used to compute C(t), which was then fed into PK modeling, using the Extended Tofts model.

Results

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<tr>
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<th>TR=15ms (13 cases)</th>
<th>TR=9.1ms (15 cases)</th>
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<tbody>
<tr>
<td></td>
<td>W/o correction</td>
<td>B1 Correction</td>
</tr>
<tr>
<td>T1 (ms) mean</td>
<td>977.87</td>
<td>1261.89</td>
</tr>
<tr>
<td>std/mean</td>
<td>0.12</td>
<td>0.15</td>
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Table 1: Prostate T1 mean/std of all cases for long/short TR at 3T.

![Image](image1.png)

**Fig.1:** T1 map (ms) before (left) and after (middle) B1 correction, and corresponding B1 map (\( \mu \text{T} \)) (right). Average ROI T1\(_{\text{before}}\) = 895.80 ms, T1\(_{\text{after}}\) = 1540.59 ms.

**Fig.2:** T2w image (A); ADC map (B); K\(_{\text{trans}}\) maps for 4 methods (i)-(iv). Yellow highlighted region denotes confirmed tumor.

Conclusion

We presented improved T1 mapping and PK quantification in prostate DCEMRI at 3T by incorporating B1 inhomogeneity correction using the Bloch-Siegert B1 mapping method. We validated the method on 28 subjects and demonstrated good T1 quantification using long TR VFA sequences combined with B1 correction. We also showed better PK maps by incorporating B1 correction into DCEMRI quantification.

References


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