In vivo measurement of $T_1$ and $T_2$ relaxation times in healthy human prostate at 7T

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Introduction

MR imaging and spectroscopy at ultra-high field (UHF) strengths (>7T) have potential advantages for the clinical assessment of prostate cancer as well as for the development of new imaging biomarkers for this disease. However, abdominal MR at these field strengths is challenging because of strongly inhomogeneous B1 fields, increased B1 absorption and increased specific absorption rate (SAR). Apart from requiring dedicated RF shimming and SAR supervision equipment, this means that any sequence to be used must operate under constraints of low maximum B1 amplitude as well as low time-averaged RF power. Knowledge of the $T_1$ and $T_2$ relaxation times of water protons in the tissues of interest is useful for development and optimization of such sequences. Here we present in-vivo measurements of these quantities in the peripheral zone (PZ) and transition zone (TZ) of the prostate and adjacent smooth muscle (SM) tissue in healthy volunteers at 7T.

Methods

Ten healthy volunteers with a median age of 32 years (range 20–38) and a median weight of 79.5 kg (range 62–100 kg) were examined on a 7T whole-body MR system (Magnetom 7T, Siemens Healthcare, Erlangen) using an 8-channel Tx/Rx external body array coil with meander elements [1] and custom-built B1+ shimming and SAR supervision software [2]. After B1 shimming, RF shimming optimized the phases of the 8 coil elements for maximum phase coherence within a region of interest (ROI) drawn around the prostate. B1+ shimming was repeated if necessary. Flip angle (FA) calibration was performed with a single-voxel spectroscopy sequence with adiabatic refocusing pulses to minimize sensitivity to B1 inhomogeneity (semi-LASER [3], duration of each refocusing pulse segment 20 ms, TE 96 ms). The RF amplitude of the excitation pulse was varied until a maximum magnitude of the water peak from a volume of interest (VOI) approximately covering the whole prostate was reached. This was assumed to correspond to 90° excitation, and RF amplitudes in following sequences were scaled relative to this value. Spatial FA variations were estimated using B1 maps created with a single-slice pre-saturated spoiled gradient echo sequence [4] (TR/TE 1000/5 ms, resolution 4.7x4.7x8 mm, pre-saturation/excitation FA approx. 60°/10°, scan time 20 s). $T_1$ relaxation times were measured in 3 volunteers in a progressive saturation experiment with fixed TR and varying flip angle, using a spoiled gradient echo (SPGR) sequence with FA of 10°, 20°, 30° and 40°, TR/TE 173/4.6 ms, resolution 1.4x1.4x3.5 mm, scan time per FA 35 s. The signal curve in each voxel was linearized, and $T_1$ maps were created by linear least-squares fits to the transformed data [5]. Monte Carlo simulations accounting for Rician noise were performed to ensure that this approximation yielded a bias <1%, so that the main source of error in the $T_1$ estimates was FA uncertainty. ROIs were drawn around the prostate and smooth muscle on a single slice on the $T_1$ maps. $T_2$ relaxation constants were measured using a single-slice multiple spin echo sequence with prolonged echo times (7.68 ms) to reduce the RF peak amplitude, an echo spacing of 15.0 ms, a total number of 8 echoes and a TR of 2.5s (resolution 0.73 x 0.87 x 3.0 mm$^3$, scan time 7 minutes). A mono-exponential decay curve was fit to the measured signal attenuation in each pixel, excluding the first echo to reduce sensitivity to B1 calibration uncertainties and inhomogeneity, to yield quantitative $T_2$ maps of the prostate and surrounding tissues. ROIs were drawn over the complete peripheral zone (PZ), transition zone (TZ) and smooth muscles left and right (SML, SMR) of the prostate to estimate $T_1$ values in these tissues. $T_2$-weighted fast spin-echo (FSE) images were acquired using refocusing pulse durations of 7.68 ms, TR/TE 3000/71 ms, echo train length 9, echo spacing 17.7 ms, resolution 0.75 x 0.75 x 3.0 mm$^3$, bandwidth 130 Hz/pixel.

Results and Discussion

Figure 1a shows an example of a quantitative $T_1$ map. No contrast between different anatomical zones within the prostate is observed. $T_1$ values were calculated in prostate and muscle in 3 volunteers (Table 1). Relative B1+ maps (figure 1b) indicated substantial differences in B1+ amplitude between muscle (arrows) and prostate (red contour). Monte Carlo simulations were performed to quantify the $T_1$ bias due to this flip angle uncertainty, and the values for muscle presented in Table 1 are corrected for this effect. Uncertainties in all $T_1$ values were estimated by considering both the SD of $T_1$ values observed within an ROI, which intrinsically contains the influence of B1 inhomogeneity, and uncertainty in the flip angle calibration. The latter is difficult to quantify, as proton motion due to e.g. peristaltic movement or rectal gas can strongly influence the B1+ field. For the uncertainties specified in Table 1 we crudely assumed a relative FA calibration uncertainty of 10%, which would result in a contribution to the relative $T_1$ uncertainty of ~21%, as was calculated using Monte Carlo simulations. $T_1$ measurements were successfully performed in 10 volunteers (Figures 1c and 2). Substantial variation between volunteers was observed particularly in the PZ, but differences between anatomic zones within the same volunteer were significant (Wilcoxon signed ranks test, $p<0.005$ in all comparisons). $T_1$ measurements were repeated several months later in 3 volunteers, yielding a maximum difference between mean $T_1$ values in the same volunteer of 12.2 ms in PZ, 9.7 ms in TZ and 7.8 ms in muscle. The present experiments suggest an optimum TE of 60–70 ms for T2w FSE imaging. FSE imaging acquired with a TE of 71 ms (Figure 1d) indeed shows clear contrast between PZ and TZ, but the SNR of these images needs further improvement. This could e.g. be achieved by further optimizing the acquisition parameters and by adding an endorectal receive coil [6]. The latter may also help reduce organ motion and the detrimental effects of air in the rectum, which could improve $B_0$ field homogeneity.

Conclusions

$T_1$ and $T_2$ relaxation times of healthy human prostate were measured in-vivo at 7T. These measurements are of aid in the development and optimization of prostate imaging and spectroscopy sequences at this field strength.

References

[1] Orzada, ISMRM 2009, 2999

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Table 1. Calculated $T_1$ relaxation times [s]. No B1+ map was available in one volunteer, so that correction of the value in muscle for FA inhomogeneity was not possible.

<table>
<thead>
<tr>
<th></th>
<th>Prostate</th>
<th>Muscle</th>
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<tbody>
<tr>
<td>Volunteer 1</td>
<td>2.0 ± 0.5</td>
<td>–</td>
</tr>
<tr>
<td>Volunteer 2</td>
<td>2.1 ± 0.6</td>
<td>1.9 ± 0.5</td>
</tr>
<tr>
<td>Volunteer 3</td>
<td>1.9 ± 0.5</td>
<td>1.4 ± 0.4</td>
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Figure 1. Transverse quantitative $T_1$ map (a), relative B1+ map (b), quantitative $T_2$ map (c) and $T_2$-weighted FSE image (d) of the prostate of a healthy volunteer.

Figure 2. ROI-averaged $T_2$ relaxation constants [ms] in 10 healthy volunteers. PZ: peripheral zone; TZ: transition zone; SM: smooth muscle.