T2* Correction Using B0 Mapping for Renal BOLD Quantification

Joshua D Kaggie[1], Vivian S Lee[1], and Glen R Morrell[1]
[1]*Utah Center for Advanced Imaging Research, Radiology, University of Utah, Salt Lake City, UT, United States*

Target Audience: Investigators who seek to use BOLD imaging to assess renal physiology

INTRODUCTION:

Chronic kidney disease (CKD) is a major cause of morbidity and mortality. Renal parenchymal hypoxia plays a central role in the progression of CKD. Despite the importance of renal hypoxia in the progression of CKD and the potential for therapeutic intervention for CKD treatment, **no satisfactory *in-vivo* method exists for measuring kidney oxygen levels.** Baseline values of T2* (R2*) from BOLD-MRI have been reported to show differences between normal kidneys and diabetic kidneys correlating to CKD in humans[(1)] and a rat model[(2)], while another study shows no correlation at all[(3)]. Changes in T2* in the kidneys after administration of a diuretic such as furosemide have been correlated with severity of CKD in human and animal subjects[(4-6)], and yet have not correlated with CKD in diabetic rat models[(7)].

Whether BOLD MRI, with or without diuretic challenge, reflects the severity of CKD remains an open question. Among the potential confounding factors are B0 inhomogeneities that can result in varying T2* values between measurements. B0 inhomogeneities become more significant at larger voxel sizes and higher field strengths, which can result in irreproducible T2* measurements. In our own renal BOLD experiments, we have found that B0 inhomogeneity significantly affects T2* estimation (**Fig. 1**). In this work, B0 maps are used to correct the T2* images for more accurate quantification on a Siemens 3T Tim Trio magnet.

METHODS:

T2* Correction: T2* shortening results from B0 inhomogeneities across a voxel and is related to T2 by: $1/T2^* = 1/T2 + \gamma\Delta B0_{inhom}$.[(8)] B0 inhomogeneity maps can be easily created from renal BOLD data. We correct the measurement of T2* due to B0 inhomogeneities, with the following relation:

$$1/T2^*_{corrected} = 1/T2^*_{measured} - \gamma\Delta B0_{inhom} \qquad (1.1)$$

Despite B0 correction, the corrected T2* will not equal T2 due to factors other than B0 field inhomogeneities, such as caused by magnetic susceptibility differences among various tissues, chemical shift, and the applied gradients.[(8)]

A B0 map that measures the difference from the static magnetic field can be obtained by acquiring images at two separate echo times and calculated with the following relation: $B0 = (\theta_{TE1} - \theta_{TE2}) / [2\pi\gamma(TE1-TE2)]$.[(9)] A $\Delta B0_{inhom}$ map is calculated by taking the derivative of B0 in each physical direction, and then calculating the magnitude of the contributions after correcting for voxel thicknesses:

$$\Delta B0_{inhom} = \sqrt{\left(\frac{dB0}{dx}\right)^2 + \left(\frac{dB0}{dy}\right)^2 + \left(\frac{dB0}{dz}\right)^2} = \frac{1}{2\pi\gamma\Delta TE}\sqrt{\left(\frac{d\Delta\theta}{dx}\right)^2 + \left(\frac{d\Delta\theta}{dy}\right)^2 + \left(\frac{d\Delta\theta}{dz}\right)^2} \qquad (1.2)$$

In Vivo Studies: Conventional 2D breathhold and 3D BOLD images using free-breathing and phase navigation[(10)] were acquired on a Siemens Trio 3T scanner with IRB approval. 2D BOLD images were acquired on two separate days with the following scan parameters: (Day 1, Fig 1A): FOV=400x400x8mm, voxel size = 1.6x1.6x8.0mm, TR = 54ms, TE = 3,6,9,12,15,19,22,25,28,31,34,38ms, flip angle = 25°, averages = 1, scan time = 14s; (Day 2, Fig 1B) used the following changed parameters: voxel size = 2.1x2.1x8.0mm, TR = 55ms, TE = 3,6,9,12,15,19,22,25,27,30,32,34ms, scan time = 11s. 3D BOLD images were acquired with the following parameters: FOV=280x280x16, voxel size = 2.2x1.1x5.0mm, TR = 95ms, TE = 5,11,17,23,29,35ms, flip angle = 25°, averages = 1, total acquisition time = 12m44s, with fat suppression and with a nominal phase navigation efficiency of 25%. A single slice was chosen for statistical analysis. Routine pre-scan shimming was performed automatically by the scanner for all sequences.

RESULTS:

Representative 2D images in a healthy volunteer (Fig 1) show variability likely due to B0 inhomogeneity. A single slice in a healthy volunteer obtained from 3D BOLD (Fig 2) shows regions with large B0 inhomogeneities that contribute to signal loss (as indicated by the arrows). T2* correction caused mostly minor changes (~3ms) in the T2* maps, although some major changes (>10ms) can be seen in the subtracted image (Fig 2G).

The median T2* changed from 39.2ms to 43.3ms across all of the cortical regions and from 36.0 to 38.8ms in the medullary regions. For the whole kidney, the median T2* changed from 36.2ms to 40.3ms, the mean R2* changed from 35.5Hz to 31.4Hz, and the standard deviation of R2* was 22.7Hz uncorrected and 20.4 Hz corrected.

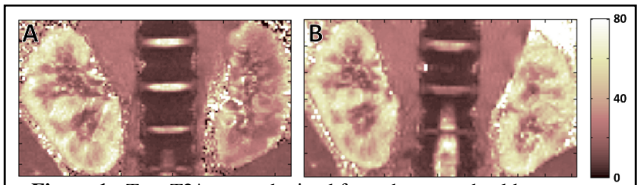

Figure 1: Two T2* maps obtained from the same, healthy volunteer obtained a day apart using 2D BOLD. The mean T2* of the left kidney in (A) is 36ms, while the right kidney mean in (A) is 56ms. The mean T2* in the left and right kidney in (B) are 53 and 54ms, respectively. The left kidney results in (A) do not match the results in (B), demonstrating inaccurate results most likely caused by day-to-day variation in B0 homogeneity.

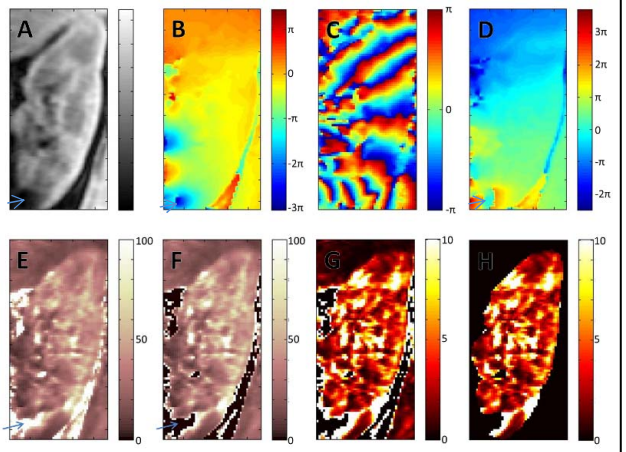

Figure 2: Images from a central kidney slice acquired with free-breathing 3D BOLD. (A) Magnitude and (B) phase image of the first echo. The arrows indicate areas with significant signal loss. B0 inhomogeneities are responsible for this signal loss. (C) Phase image of the final echo. (D) Phase difference between first and second echo, which is proportional to B0. (E) T2* map. (F) T2* map corrected by $\Delta B0_{inhom}$. (F) Difference between the original T2* map and the corrected T2* map. (G) The region shown here was used for statistical calculations for the whole kidney.

DISCUSSION:

The changes in T2* due to B0 inhomogeneities in our experiment are around 4ms, obtained with a 3D BOLD sequence with relatively small voxel sizes. The T2* differences due to B0 inhomogeneity are on the order of the changes seen with diuretic (~ 7 ms)[(6)], which suggests that B0 correction may be necessary at 3T. Larger B0 inhomogeneities will occur with larger voxel sizes, typical of 2D BOLD imaging, further corrupting the measured T2* values. 3D BOLD imaging using small voxel sizes and B0 correction should result in more accurate T2* maps for studying renal disease.

CONCLUSION:

T2* measurements used for renal BOLD-MRI can be corrupted by B0 inhomogeneities, that should be corrected for more accurate T2* quantification.

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REFERENCES: 1. Yin WJ et al. Eur J Radiol 2012;81(7):1426.
2. dos Santos EA et al. Invest Radiol 2007;42(3):157.
3. Michaely HJ et al. Kidney Int 2012;81(7):684.
4. Warner L et al. Invest Radiol 2011;46(1):41.
5. Priatna A et al. JMRI 1999;9(6):842.
6. Prasad PV et al. Circulation 1996;94(12):3271.
7. Ries M et al. JMRI 2003;17(1):104.
8. Chavhan GB et al. Radiographics 2009;29(5):1433.
9. Park HW et al. MRM 1988;7(3):340.
10. Morrell G et al. 21st ISMRM, 2013; SLC, UT, USA. p 1569.