Improved R2* Measurement Accuracy with Absolute SNR Truncation and Optimal Coil Combination

X. Yin[1,2], S. Shah[3], P. Kellman[4], and A. C. Larson[1,2]

[1]Department of Radiology, Northwestern University, Chicago, IL, United States, [2]Department of Electrical Engineering and Computer Science, Northwestern University, Evanston, IL, United States, [3]Siemens Medical Solutions, Chicago, IL, United States, [4]Laboratory of Cardiac Energetics, NIH, Bethesda, MD, United States

INTRODUCTION

Accurate R2* measurements are critical for many quantitative imaging applications. Typically, R2* is estimated via mono-exponential fitting of signal decay within a series of GRE images sampled at increasing TE. Surface coil array images at each TE are conventionally combined using a root sum-of-squares (RSS) approach prior to data fitting procedures. However, rapid signal decay can result in poor SNR at later TEs. RSS approaches rectify and bias noise resulting in systematic fitting errors [1]. To improve the accuracy of R2* measurements, various models have been established to correct the data prior to fitting [2] or to improve image SNR for superior fidelity at later TEs [3]. There remains uncertainty as to which approach is optimal for R2* mapping. The objective of our work was to investigate the accuracy of each method during low SNR R2* measurements (common for visceral organs located relatively far from the surface coil array). By combining a truncation model [2] with array coil combination methods [4], R2* maps were derived from both truncated RSS and optimal B1-weighted combined (OBC) datasets. We compared the accuracy of these methods in phantoms and healthy volunteers.

METHODS

Comparison of the truncation models: We compare three different models for R2* mapping. (1) *Model A - SNR_{global} RSS:* Here, truncation model of [2] was modified to select the longest TE utilized for data fitting. First, SNR_{global} was calculated as signal divided by the standard deviation of the image background noise [2] for RSS combination. Thereafter, all echoes with SNR_{global} below a given threshold were excluded from R2* calculations, thereby truncating the range of included TEs. This is a commonly utilized approach but it does not account for spatially variant noise characteristics. (2) *Model B - SNR_{voxel} RSS:* Here images were obtained in SNR units by first measuring coil-wise noise characteristics using noise-only pre-scans, pre-whitening the image data and then correcting noise-bias after RSS combination [4]. Voxel-wise R2* measurements were performed with each fitted TE range truncated based upon these absolute voxel-wise SNR estimates. (3) *Model C - SNR_{voxel} OBC*: The only difference between Model B and Model C was that signals from multiple coils were combined using OBC approach rather than RSS combination.

Data acquisition: R2* phantoms were constructed by filling cylindrical 2L polystyrene bottles with water doped with either 1.0 mmol/L (*Phantom I*) or 0.4mmol/L (*Phantom II*) $MnCl_2.H_2O$. Studies were performed using a 3T clinical scanner (Magnetom Trio, Siemens AG HCS, Erlangen, Germany) with 8-channel head coil for phantom measurements and a 32-channel A/P surface array coil for IRB-approved volunteer measurements. For data acquisition a multi-gradient-echo (MGRE) sequence was used with parameters: TR(phantom/leg/abdomen)=5000/1000/100ms, echo train length=24, TE=2-65ms, bandwidth=640Hz/pixel and slice thickness=5mm. We achieved a range of different SNR levels using flip-angles (FA) = 2°, 5° and 10°. Gold standard R2* measurements were established through averaged datasets acquired using the body transmit/receive coil (32 averages for phantoms and 7 averages for volunteer leg studies).

Data Analysis: R2* maps were calculated via voxel-wise non-linear least square fitting of the mono-exponential decay signal components. For both phantom and volunteer leg studies, we computed voxel-wise absolute error (VAE) between gold standard R2* maps and comparison model R2* maps. Mean VAE was evaluated for each model at SNR truncation thresholds incremented from SNR=0 to SNR=30 (steps of 0.1 SNR units).

RESULTS

Phantom: R2* values were found to be 133 s^{-1} for *Phantom I* and 55 s^{-1} for *Phantom II*. At FA = 2°, 5° and 10°, mean VAE at truncation thresholds from 0 to 30 are displayed in **Fig.1** for both phantoms. VAE was reduced with a) increasing SNR (increased FA) and b) lower R2* value (Phantom I vs. II). As depicted by the VAE curves, error decreased as low SNR TEs were removed and increased as high SNR TEs were removed. In general, SNR_{voxel} OBC method achieved minimum VAE level with little need for truncation of low SNR data (FA=2°) and no need for truncation with high SNR data (FA=10°). SNR_{global} and SNR_{voxel} RSS methods required data truncation at higher SNR thresholds to achieve their optimal R2* estimates.

Volunteer: For lower leg scans, R2* for bone marrow and muscle tissue was 86 s^{-1} and 40 s^{-1}, respectively. **Fig. 2** depicts the mean VAE curves at FA= 2°, 5° and 10°. Similar to the phantom studies, SNR_{voxel} OBC produced relatively low error levels and reduced the need for truncation. SNR_{global} RSS and SNR_{voxel} RSS required truncation to improve the accuracy of R2* estimates. Fast signal decay within bone marrow produced larger errors along with a greater need for truncation. Muscle signal decayed at a slower rate and therefore produced lower error and less of a need for truncation. For abdominal scans with no truncation, the SNR_{voxel} OBC method offers the potential to significantly improve R2* measurement accuracy within deeper visceral tissue structures, **Fig. 3**.

CONCLUSIONS: As expected, R2* measurement accuracy improved with increased SNR and slower signal decay rates. Within a setting of low SNR with fast signal decay, truncation can be an effective method to reduce measurement error. However, in practice, the SNR threshold providing minimum error may be unknown and therefore the SNR_{voxel} OBC method is the optimal choice given the minimal truncation requirements for this approach.

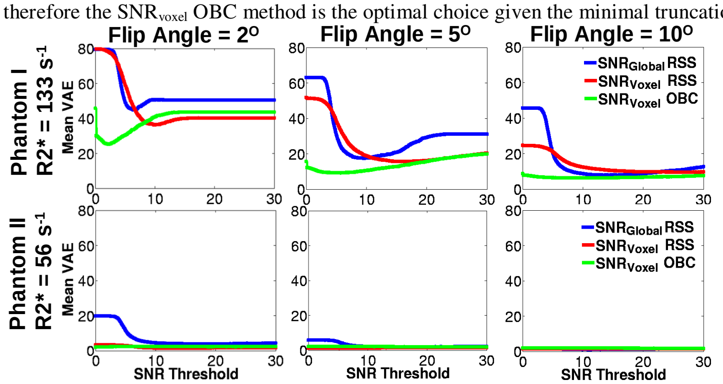

Fig.1 Phantom Studies: Mean VAE curves at flip angle=2°, 5°, 10° for Phantom I and II. Error decreases with increasing SNR (higher flip-angle) and decreasing R2* value. Limited truncation is needed for SNR_{voxel} OBC as opposed to SNR_{global} RSS and SNR_{voxel} RSS, particularly at higher SNR.

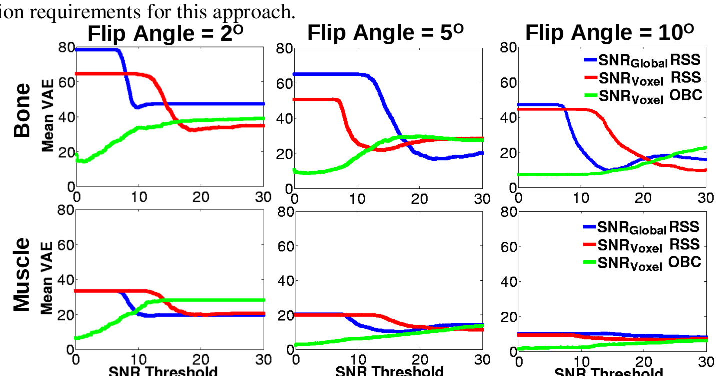

Fig.2 Volunteer Leg Studies: Mean VAE curves at flip angle=2°, 5°, 10° for bone marrow and muscle R2* measurements. Higher SNR and lower R2* values corresponded to lower error levels. SNR_{voxel} OBC reduced the need for truncating data during fitting procedures.

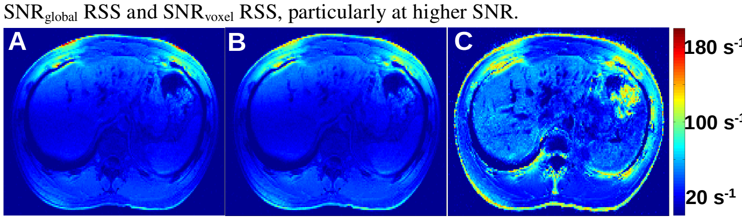

Fig.3 Abdominal Studies: R2* maps reconstructed using (**A**) SNR_{global} RSS, (**B**) SNR_{voxel} RSS, and (**C**) SNR_{voxel} OBC methods. Notice that methods A and B bias liver R2* values (greater R2* at high SNR periphery nearer to surface coils) whereas method C appropriately depicts relatively uniform R2* within normal liver parenchyma, independent of tissue position relative to surface coil array.

REFERENCE: [1] Gilbert G, IEEE Medical Imaging 2008; 26: 1428-1435. [2] He T, MRM 2008; 60: 250-256. [3]Graves M. et al, JMRI 2008; 28: 278-281. [4] Kellman P. et al, MRM 2005; 54: 1439-1447