Proton-Constrained $CMRO_2$ Quantification with Direct ^{17}O-MRI at 3 Tesla

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Introduction

Abnormalities in brain metabolism are found in various diseases: tumors, cerebrovascular diseases and neurodegenerative diseases, such as Parkinson's, Alzheimer's and Huntington's. A useful biomarker of the metabolic brain activity is the cerebral metabolic rate of oxygen consumption ($CMRO_2$) [1] which can be quantified with ^{15}O positron emission tomography (PET) [2], and direct or indirect ^{17}O MRI methods. In direct ^{17}O MRI the signal change of ^{17}O nucleus is detected during and after the administration of a ^{17}O-enriched gas. To overcome the low SNR, ^{17}O-MRI has so far only been applied at ultra-high fields ($B_0 \geq 7T$) [3,4,5]. To make $CMRO_2$ quantification available at clinical field strengths ($B_0 \leq 3T$), where the SNR is smaller [6], prior information from co-registered ^1H MRI can be used in combination with iterative constrained image reconstruction. So far, only the object shape using a binary mask (BM) has been used for constrained reconstruction; however, this mask cannot distinguish different brain tissues. In this work we use an anisotropic non-homogeneous diffusion operator [7] on ^1H MRPAGE data as a constraint to improve the image quality of ^{17}O images and the localized $CMRO_2$ quantification in a frontal lobe.

Materials and Methods

For the constrained reconstruction, a dynamic *in vivo* ^{17}O measurement consisting of 45 3D data sets of the brain was used. The data was acquired with a temporal resolution of 1 min and was obtained in a four phase $CMRO_2$ inhalation experiment in a healthy volunteer (male, age 49y) at a clinical 3 Tesla MR system (Tim Trio, Siemens). Using a custom-built TxRx ^{17}O head coil [8], a density-adapted projection sequence (DAPR) [9] and a re-breathing system, the ^{17}O 3D data sets were acquired during a baseline phase under free breathing (10 min), an ^{17}O inhalation phase (5 min), a re-breathing phase with a closed breathing circuit (8 min), and a final wash-out phase (22 min), during which the volunteer was breathing room air. In total, 2.5 L of 70% enriched ^{17}O gas (NUKEM Isotopes) was delivered. The following imaging parameters were used: nominal resolution: $(10\ mm)^3$, TE = 0.52 ms, TR = 8 ms, BW = 150 Hz/px, T_{RO} = 6.7 ms, 7500 projections x 128 radial points interpolated to a 128^3 matrix.

^1H 3D MPRAGE data (resolution: 0.6x0.6x1 mm^3, TI = 1100 ms) were used for co-registration, tissue segmentation, and iterative reconstruction. In addition, after co-registration a numerical ^{17}O brain phantom was constructed from the segmented WM, GM and CSF compartments of the MPRAGE data with the acquisition parameters of the ^{17}O-MRI measurement. A T_2^*-decay (2 ms) and Gaussian noise were applied using a measured SNR of 20. To simulate the dynamic ^{17}O MRI experiment, the intensities of the WM and GM were modulated with concentration-time curves of a 4-phase kinetic model [4] and literature $CMRO_2$ values [2].

Both phantom and experimental ^{17}O images were reconstructed with Kaiser-Bessel (KB) gridding method with and without Hanning filter. Alternatively, an iterative reconstruction was applied by minimizing the objective function

$$J(\boldsymbol{\rho}) = \|\mathbf{A}\cdot\boldsymbol{\rho}-\boldsymbol{y}\|^2 - \lambda_D \int \boldsymbol{\rho}\nabla(\mathbf{D}\nabla\boldsymbol{\rho}) \quad \text{with} \quad \boldsymbol{D} = \left(\mathbf{1} - \frac{\boldsymbol{g}\cdot\boldsymbol{g}^T}{|\boldsymbol{g}|^2} \Big/ \sqrt{1+\frac{\boldsymbol{g}^2}{a^2}}\right) \qquad (1),$$

where $\mathbf{A}$ denotes the system matrix that maps the image $\boldsymbol{\rho}$ to the corresponding raw data $\boldsymbol{y}$. In the regularization term the gradient operator $\boldsymbol{g}$ is applied to the ^1H MRPAGE image ($\boldsymbol{g} = \nabla \mathrm{Im}_{\mathbf{1H}}$). The factors λ_D and a were chosen providing the best correspondence with PET $CMRO_2$ values [2] without image distortions: $a = 2\cdot 10^{-03}\overline{Im_{1H}}$, λ_D= 30 / 4000 for phantom / *in vivo*. $CMRO_2$ values in the frontal lobe were determined by a fit with the 4-phase kinetic model.

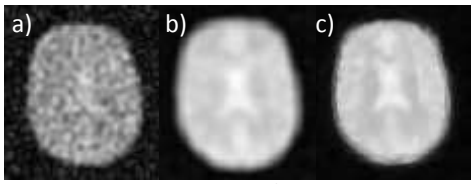

Fig. 1*: Transverse slice of the simulated brain phantom reconstructed with Kaiser-Bessel (KB) gridding method without (a) and with Hanning window (b) and with ^1H constraint (c). TA = 10 min.*

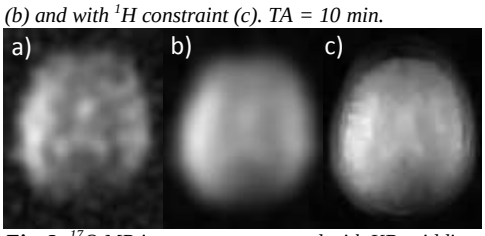

Fig. 2*: ^{17}O MR images reconstructed with KB gridding without (a) and with Hanning window and with proton constraint (c). TA = 10 min*

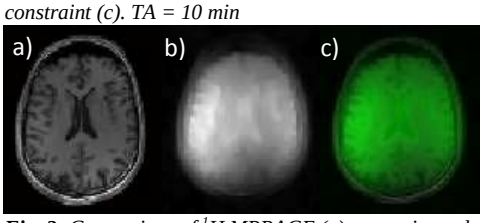

Fig. 3*: Comparison of ^1H MPRAGE (a), co-registered ^{17}O MRI image with proton constrained with TA = 10 min (b and green in c) and fusion of both images (c).*

Results and Discussion

Proton-constrained ^{17}O images show better quality compared to KB regridding (Fig.1-2), since they contain more visible anatomical information (cf. the ventricles). The brain phantom simulations show that decent precision of $CMRO_2$ values in 1 ml of GM cannot be obtain without filtering. ^1H-constrained reconstruction gives most precise values in phantom simulations and localized *in vivo* $CMRO_2$ quantification in GM and WM in a frontal lobe (Fig. 4). For GM the value is in a good agreement with literature PET values [2] and has a factor of 1.8-2.8 smaller uncertainty compared to KB gridding. For WM $CMRO_2$ value is underestimated, since it is affected by partial volume effects (PVE) with other brain tissues. To increase the resolution of the low-SNR ^{17}O MRI data at 3T, in brain tumor studies ^1H-constrained iterative reconstruction is an interesting alternative to conventional post-processing methods such as filtering, which might also help to overcome limitations due to PVEs.

Tissue	$CMRO_2$ [µmol/g_{tissue}·min]						
	Simulated brain phantom			In vivo ^{17}O MRI			^{15}O-PET
	no Hann	with Hann	^1H constrained	no Hann	with Hann	^1H constrained	
GM	1.65±0.63	1,01±0.15	1.12±0.15	1.55±0.34	1.68±0.21	1.56±0.12	1.59±0.23
WM	0.75±0.23	0.80±0.18	0.70±0.14	- *	1.16±0.39	1.12±0.18	0.62±0.10

Fig. 4*: $CMRO_2$ values obtained in the simulated phantom (in 1 ml GM and WM) and with direct ^{17}O-MRI at 3 Tesla (in 2 ml GM and 4 ml WM) compared with PET literature values from [2].*was not defined due to low SNR*

References

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