Joint Estimation of Attenuation and Activity Distributions for Clinical non-TOF FDG Head Patient PET/MR Data Employing MR Prior Information

T. Heußer[1], C. M. Rank[1], M. T. Freitag[2], H.-P. Schlemmer[2], A. Dimitrakopoulou-Strauss[3], T. Beyer[4], and M. Kachelrieß[1]

[1]Medical Physics in Radiology, German Cancer Research Center (DKFZ), Heidelberg, Germany
[2]Department of Radiology, German Cancer Research Center (DKFZ), Heidelberg, Germany
[3]Clinical Cooperation Unit Nuclear Medicine, German Cancer Research Center (DKFZ), Heidelberg, Germany
[4]Center for Medical Physics and Biomedical Engineering, Medical University of Vienna, Austria

Correspondence to: thorsten.heusser@dkfz.de

GERMAN CANCER RESEARCH CENTER IN THE HELMHOLTZ ASSOCIATION

50 Years – Research for A Life Without Cancer

Introduction

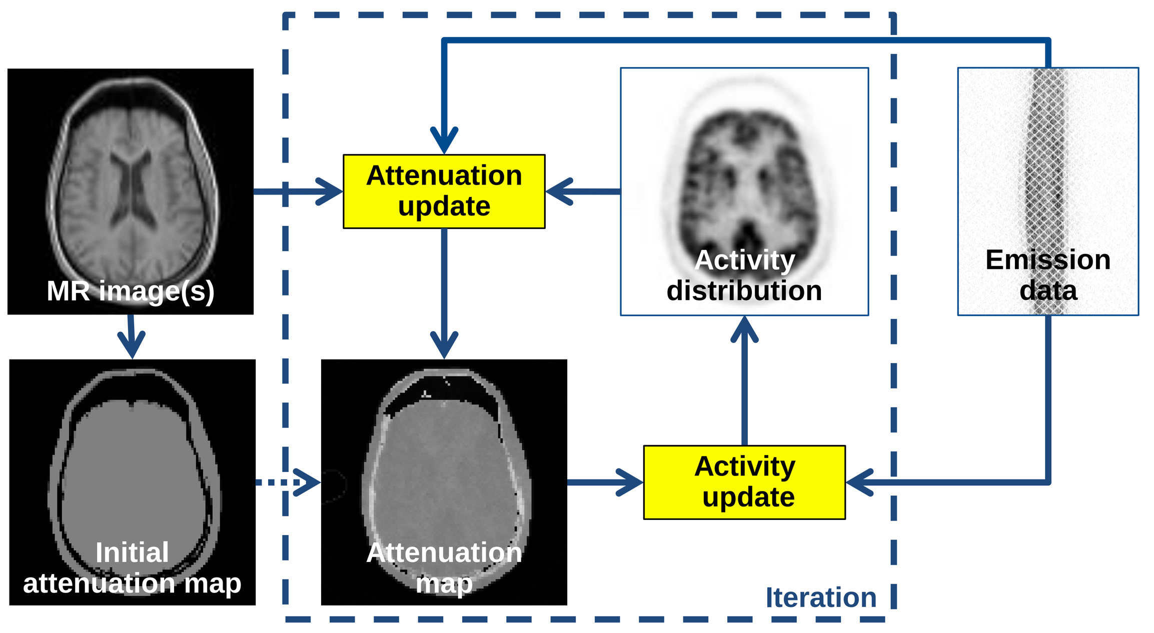
Accurate quantification of the radiotracer activity distribution in positron emission tomography (PET) mandates attenuation correction (AC). In combined PET/MR imaging, AC is a major challenge since direct conversion of the MR information into corresponding PET attenuation coefficients is not possible. The standard MR-based approach (MRAC) performs a segmentation into several tissue classes but neglects bone attenuation and, therefore, leads to an underestimation of the reconstructed PET activity distribution by up to 30%. Ongoing efforts aim at improving MRAC by considering bone attenuation derived from, e.g., ultrashort-echo-time (UTE) sequences, patient databases, or additional low-dose CT scans. We have recently proposed an algorithm to jointly estimate attenuation and activity distributions for non-TOF PET/MR[1]. The algorithm uses simultaneous reconstruction of attenuation and activity from the PET emission data, updating attenuation and activity in an alternating manner. Voxel-dependent expectations on the attenuation coefficients based on MR prior information are used during the attenuation update. Since the proposed algorithm is based on the maximum-likelihood reconstruction of attenuation and activity (MLAA)[2], we call it MR-based MLAA (MR-MLAA).

Materials and Methods

MR-MLAA uses the PET emission data and available MR images as input data. The algorithm aims at optimizing a cost function consisting of the standard log-likelihood L and two prior terms, L_S and L_I. Solving the cost function is done by iterative optimization, alternately updating the activity while keeping the attenuation constant and vice versa. The activity update is performed by standard maximum likelihood expectation maximization (MLEM) while the attenuation update is done using a gradient-descent method for transmission tomography[3]. Only the attenuation update is affected by the prior terms. The aim of the smoothing prior L_S is to favor attenuation maps which are locally smooth. It is realized by a Gibbs probability distribution penalizing differences in the attenuation values between neighboring voxels. The aim of the intensity prior is to favor the occurrence of pre-defined attenuation values. In contrast to the original MLAA, the intensity prior is realized as a local, voxel-dependent probability distribution of expected attenuation coefficients. It is defined as linear combination of the air/bone intensity prior L_{AB} and the soft tissue intensity prior L_{ST}, both of which are directly defined using piecewise linear functions as proposed in reference [2]. The voxel dependency is based on the weighting parameter $0 \leq \omega(\boldsymbol{r}) \leq 1$ which is given by a mask derived from the available MR images. The mask consists of two segments. One segment contains all voxels which are assumed to represent either air or bone ($\omega = 0$), the other one contains all voxels assumed to represent soft tissue ($\omega = 1$). The strength of the intensity prior relative to the likelihood function (and to the smoothing prior) can be chosen individually for the air/bone and the soft tissue segment using the parameters β_{AB} and β_{ST}, respectively. To obtain an initial attenuation map, the voxels corresponding to the air/bone segment are set to $\mu = 0$ while voxels corresponding to the soft tissue segment are set to $\mu = 0.01$ mm^{-1}. Both mask and initial attenuation map are smoothed to allow for intermediate values for ω and μ, respectively.

We evaluated our proposed algorithm for five clinical ^{18}F-FDG-PET/MR patient datasets of the head region, acquired with a Siemens Biograph mMR (Siemens Healthcare, Erlangen, Germany). Attenuation correction was performed using the standard MR-based attenuation map as provided by the scanner, the attenuation map obtained by MR-MLAA, and a CT-derived attenuation map, which was co-registered to the MR and scaled to 511 keV. Reconstructions employing ordered subset expectation maximization (OSEM) with 3 iterations and 21 subsets were performed using the different attenuation maps.

MR-MLAA Workflow



MR-MLAA Algorithm

Cost function *C*:

$$C(\boldsymbol{\lambda}, \boldsymbol{\mu}) = \underbrace{L(\boldsymbol{\lambda}, \boldsymbol{\mu})}_{\text{Standard log-likelihood}} + \underbrace{L_{\mathrm{S}}(\boldsymbol{\mu}) + L_{\mathrm{I}}(\boldsymbol{\mu})}_{\text{Prior terms}}$$

Log-likelihood *L*:

$$L(\boldsymbol{\lambda}, \boldsymbol{\mu}) = \sum_j \left(p_j \ln \hat{p}_j - \hat{p}_j\right)$$

with estimated projections $\hat{p}_j = \frac{a_j}{n_j} \sum_i M_{ij} \lambda_i + \frac{s_j}{n_j} + r_j$

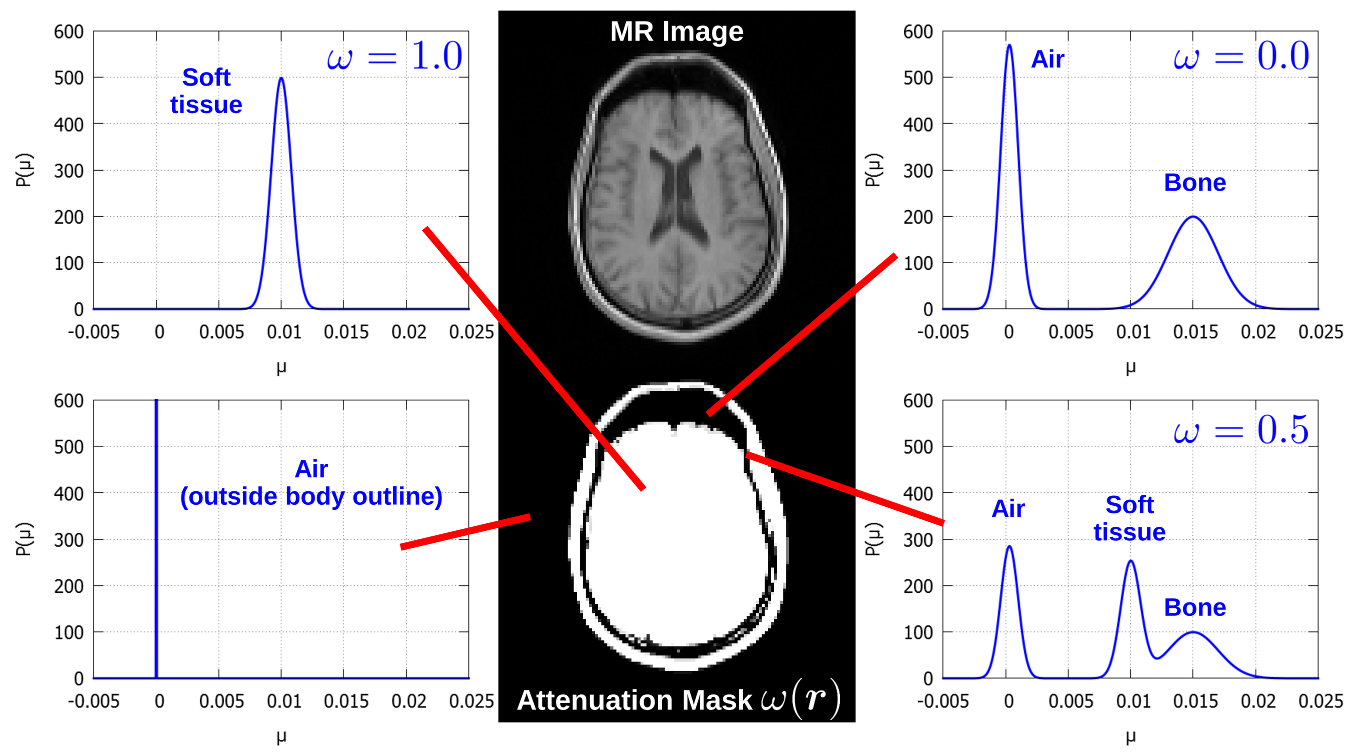
and attenuation factor $a_j = e^{-\sum_i \mu_i l_{ij}}$

Intensity prior L_I:

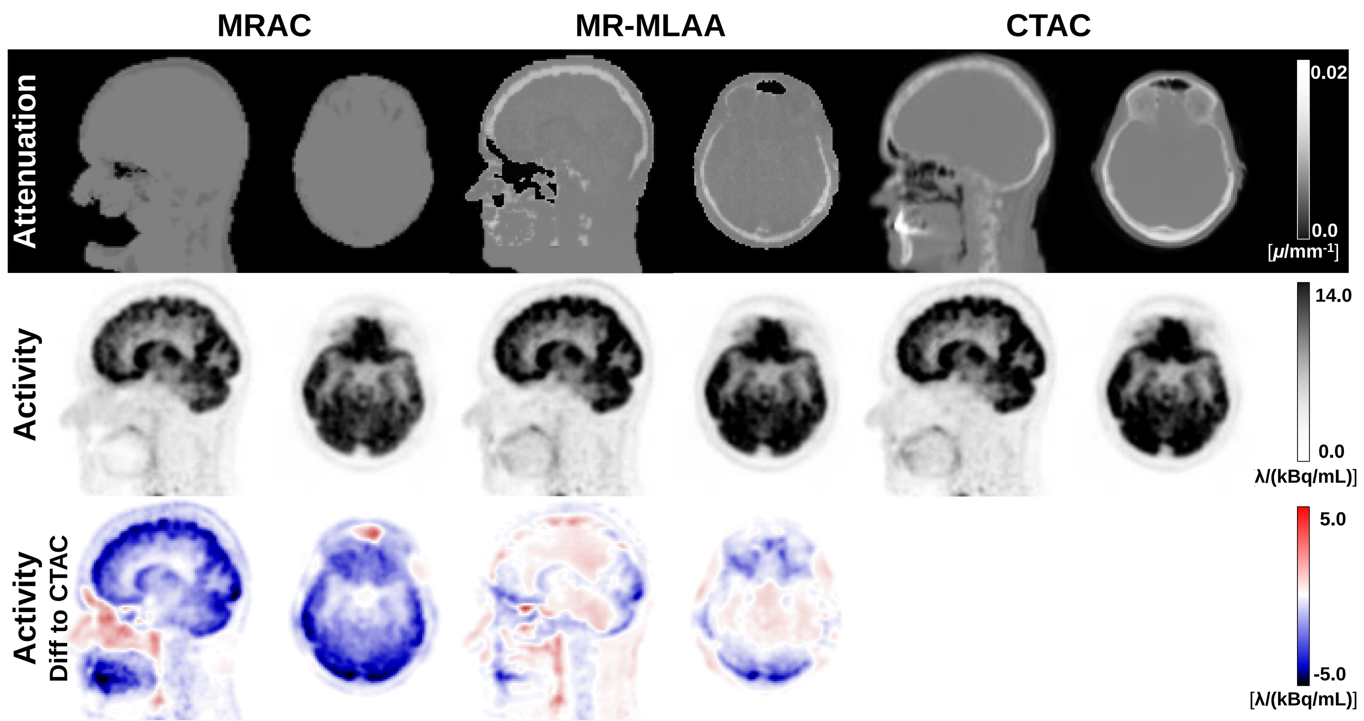
$$L_{\mathrm{I}}(\boldsymbol{\mu}) = \omega(\boldsymbol{r}) \beta_{\mathrm{ST}} L_{\mathrm{ST}}(\boldsymbol{\mu}) + (1 - \omega(\boldsymbol{r})) \beta_{\mathrm{AB}} L_{\mathrm{AB}}(\boldsymbol{\mu})$$

$\boldsymbol{\lambda}$ Activity
$\boldsymbol{\mu}$ Attenuation
i Voxel index
j LOR index
p_j Emission data
n_j Normalization
s_j Scatter
r_j Randoms
M_{ij} System matrix element
l_{ij} Intersection length of voxel *i* and LOR *j*

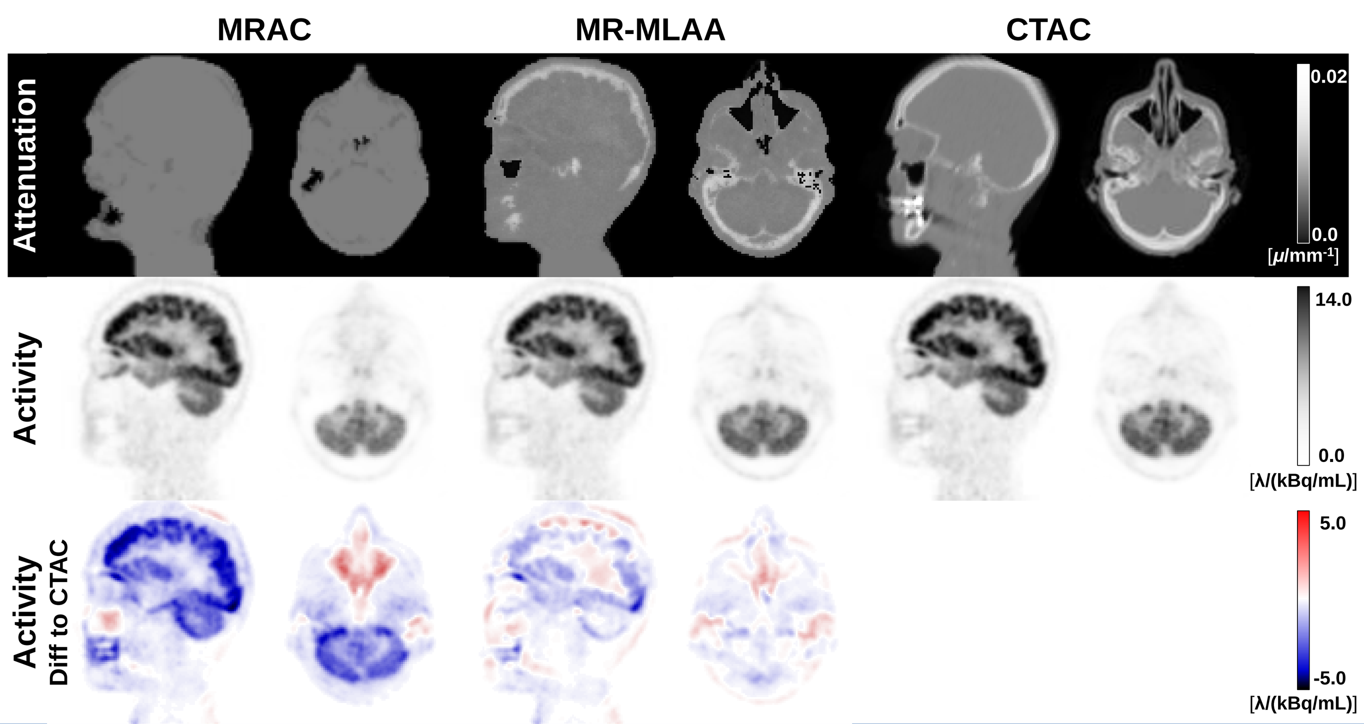
Intensity Prior L_I



Results Patient 1



Results Patient 2



Results

The attenuation map obtained by standard MRAC does not contain any information on bone tissue. This leads to an underestimation of the reconstructed activity, especially in the vicinity of bone. Across the five investigated datasets, the average activity underestimation evaluated in the full brain is 10.7% for MRAC compared to CTAC. Using MR-MLAA, bone information in the attenuation map can mostly be recovered with only few misclassifications of air/bone as soft tissue and bone as air. At the same time, air cavities like the frontal and nasal sinuses are preserved. Due to the presence of bone tissue in the MR-MLAA attenuation map, the activity estimation is greatly improved. The average activity underestimation across the five datasets is reduced to 3.4% evaluated in the full brain and compared to CTAC. The presented method will, potentially, benefit from additional time-of-flight (TOF) information.

Acknowledgements

This study was supported by the Helmholtz International Graduate School for Cancer Research, Heidelberg, Germany.
Parts of the reconstruction software were provided by RayConStruct® GmbH, Nürnberg, Germany.

[1] T. Heußer, C.M. Rank, M.T. Freitag, A. Dimitrakopoulou-Strauss, H.-P. Schlemmer, T. Beyer, and M. Kachelrieß, "MR-Consistent Simultaneous Reconstruction of Attenuation and Activity for non-TOF PET/MR," IEEE Trans. Nucl. Sci., in press.
[2] J. Nuyts, P. Dupont, S. Stroobants, R. Benninck, L. Mortelmans, and P. Suetens, "Simultaneous maximum a posteriori reconstruction of attenuation and activity distributions from emission sinograms," *IEEE Trans. Med. Imaging*, vol. 18, no. 5, pp. 393-403, 1999.
[3] J. Nuyts, B. De Man, P. Dupont, M. Defrise, P. Suetens, and L. Mortelmans, "Iterative reconstruction for helical CT: a simulation study," *Phys. Med. Biol.*, vol. 43, no. 4, pp. 729-737, 1998.