ML and MAP PET reconstruction with MR-voxel sizes for simultaneous PET-MR

Martin A. Belzunce, Abolfazl Mehranian, James Bland and Andrew J. Reader

Abstract—**The introduction of clinical simultaneous PET-MR scanners has brought new opportunities to use anatomical MR images to assist PET image reconstruction. In this context, MR images are usually downsampled to the PET resolution before being used as anatomical priors in MR-guided PET reconstruction. However, the reconstruction of PET images at the MR-voxel size could achieve a better utilization of the high resolution anatomical information and improve the partial volume correction obtained with these methods. When the PET reconstruction needs to be done in a higher resolution matrix a number of artifacts arise in the image reconstruction, depending on the projector and system matrix used. In this work, we propose a method that modifies the system matrix to overcome these difficulties and we show reconstructed images of a NEMA phantom and patient data for standard and high resolution image sizes. The higher resolution reconstructed images show a better delineation of the edges and a modest improvement of the contrast in the smallest spheres of the NEMA phantom. In addition, we evaluated the method for MR-guided MAP reconstruction, where patient data was reconstructed using a Bowsher prior computed from the T1-weighted image in its original resolution. The reconstructed images with MR-voxel sizes showed a better definition of the structures of the brain and quantitatively better contrast in the striatum, showing that MR-guided MAP reconstruction with MR-voxel size can enhance the partial volume correction.**

I. Introduction

THE introduction of clinical simultaneous PET-MR scanners has brought new opportunities to use MR information to assist PET image reconstruction. For example, MR images can be used as anatomical priors in Bayesian maximum *a posteriori* (MAP) PET image reconstruction with the purpose of suppressing noise and doing partial volume correction (PVC) [1], [2]. In this context, MR images are usually downsampled to the PET resolution before being used in MR guided PET reconstruction. However, the reconstruction of PET images at the MR voxel size could achieve a better utilization of the high resolution anatomical information and improve the partial volume correction obtained with these methods.

The voxel size of the PET reconstructed images is related to the radial sampling of the lines of response and therefore to the crystal size of a given scanner. When the PET reconstruction needs to be done in a higher resolution matrix, for example to accomplish full use of the anatomical information in MR guided image reconstruction, a number of artifacts arise in the image reconstruction, depending on the projector and system matrix used.

In this work, we consider the pitfalls of performing PET image reconstruction with a smaller voxel size than the standard native pixel size of the scanner. We propose a method to overcome these difficulties and we show reconstructed images of phantom and patient real data for standard and MR-voxel sizes. Finally, we evaluate the method for MR guided MAP reconstruction using the T1-weighted image as anatomical prior in its original resolution.

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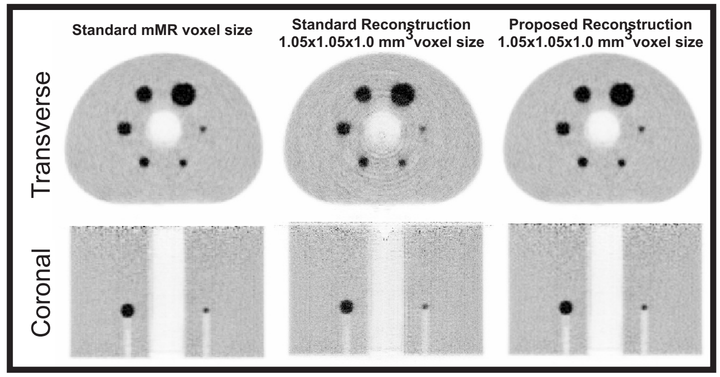

Fig. 1. NEMA phantom scan reconstructed in the standard mMR voxel size and in a higher resolution matrix size ($1.0 \times 1.0 \times 1.0$ mm^3 voxel size) with the standard MLEM reconstruction using a Siddon projector (middle) and the proposed method (right).

II. Materials and Methods

The Siemens mMR PET-MR scanner was used to evaluate the problems that arise when reconstructing images in a higher resolution than the standard voxel size. The mMR scanner sinograms have a radial bin size of 2.04455 mm and the standard reconstructed images have a $2.0865 \times 2.08625 \times 2.03125$ mm^3 voxel size.

Fig. 1 shows a reconstructed image with the MLEM algorithm using the Siddon projector [3] for the standard and a $1 \times 1 \times 1$ mm^3 voxel size, where the reconstruction in a higher resolution matrix (middle column) introduces artifacts and gaps in the images. The main issue in this reconstruction is produced by the under-sampling of the projection data when using a Siddon projector with radial sampling smaller than the voxel sizes.

This issue can be solved by modifying the ray-tracing projector to take every pixel into consideration. For example, two different approaches can be used:

- Employing a multi-ray projector, where each sinogram bin is projected using multiple rays [4]. The number of rays needed depends on the pixel size of the reconstructed images and for that reason it involves a higher computational cost that scales with the upsampling factor, making it not very efficient.
- Using an interpolation matrix, where an interpolation matrix is introduced in the system matrix:

$$P_{hr} = X_{lr} D_{hr \to lr} \tag{1}$$

where P_{hr} is the projection system matrix that projects a high resolution image into the standard sinogram size of the scanner, X_{lr} is the Siddon projector for the standard voxel size and

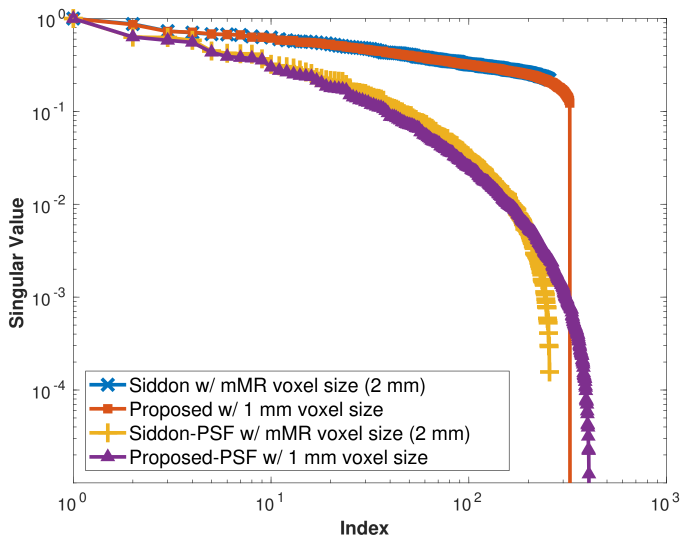

Fig. 2. Singular values of the standard (2 mm voxel size) and the proposed (1 mm voxel size) system matrices, without and with PSF modelling, for a patch in the centre of the FOV.

$D_{hr \to lr}$ is a matrix that interpolates a high resolution image into the standard image size.

In this work, we will focus in the second approach since it is more efficient and flexible. An important aspect of the implementation of this method is the upsample interpolation method used in the computation of the transpose of the projection matrix needed in iterative reconstruction:

$$P_{hr}^T = D_{hr \to lr}^T X_{lr}^T \tag{2}$$

where the upsample matrix $D_{hr \to lr}^T$ needs to be the transpose of the downsample matrix $D_{hr \to lr}$ to avoid having an unmatched projector/backprojector.

The MLEM reconstruction using this method is defined by:

$$f^{k+1} = \frac{f^k}{D_{hr \to lr}^T X_{lr}^T A N 1} D_{hr \to lr}^T X^T A N \frac{b_a}{N A X_{lr} D_{hr \to lr} f^k + s + r} \tag{3}$$

where f^k is the reconstructed image at iteration number k, A and N are diagonal matrices with the attenuation and normalization factors and s and r are the scatter and randoms estimates respectively.

To have a better insight into the potential benefits of using smaller voxel sizes, in Fig. 2 the singular values of the standard (2 mm voxel size) and the proposed (1 mm voxel size) system matrices, without and with PSF modelling, are shown. The matrices were computed for a $16 \times 16 \times 8$ mm^3 patch (i.e. $16 \times 16 \times 8$ and $8 \times 8 \times 4$ voxels respectively) in the centre of the field of view (FOV). It can be seen that, despite the downsampling factor, the proposed system matrix is able to recover higher frequencies than the standard method. This is more notable for the case of PSF modelling, where the inherent recovery of higher frequencies is enhanced.

A. Image Reconstruction and Evaluation

A 2 hour scan of the NEMA IQ phantom was used to evaluate quantitatively the reconstructed images for the standard voxel size and for a $1 \times 1 \times 1$ mm^3 voxel size. All the phantom spheres were filled with an activity concentration of 3.8 times the background. The contrast and coefficient of variation (COV) for every sphere were computed as stated in the NEMA standard. For the standard voxel size, the images were resampled into the higher resolution matrix size before computing the metrics.

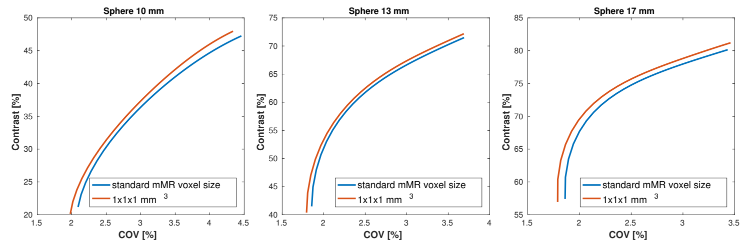

Fig. 3. COV vs contrast as function of the iteration number in the 3 smallest spheres of the reconstructed images of the NEMA phantom, for the standard voxel size and the higher resolution voxel size.

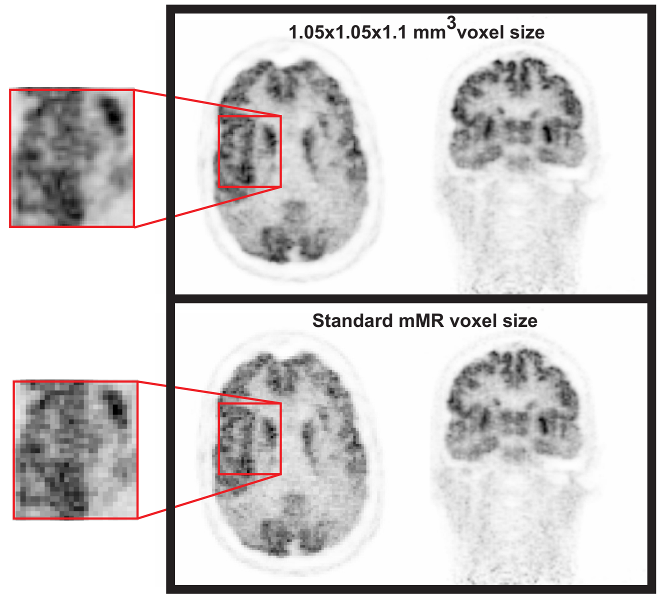

Fig. 4. MLEM reconstruction at iteration number 100 of a [^{18}F]FDG brain study using the MR voxel size and the modified system matrix (top), and using the standard mMR voxel size and Siddon system matrix (bottom).

In addition, patient data acquired with the mMR scanner of a [^{18}F]FDG brain study with a total of 5×10^8 prompt counts was used to evaluate the method. The images were reconstructed with the MLEM algorithm for the standard voxel size and for a $1.05 \times 1.05 \times 1.1$ mm^3 voxel size (MRvox). The latter corresponds to the resolution of the T1-weighted image acquired simultaneously during the PET scan. For the reconstruction into a higher resolution matrix the method described in the previous section was used, while for the native mMR PET voxel size the standard system matrix was used. In addition, the patient data was reconstructed with a MR guided MAP reconstruction using a Bowsher prior [5] computed from the T1-weighted image using 40 neighbours in a 5x5x5 neighbourhood for the two different voxel sizes. For the standard voxel size, the T1-weighted image was downsampled to match the PET matrix. In order to assess quantitatively the reconstructed images, the contrast and coefficient of variation in the caudate and putamen were computed for every iteration. For the standard voxel size, the images were interpolated into the higher resolution matrix before computing these metrics. All the methods were run for 300 iterations.

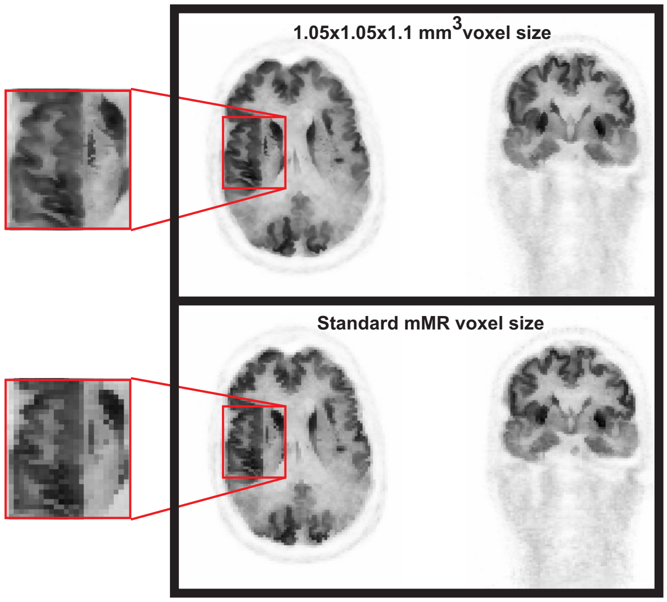

Fig. 5. MAP reconstruction using Bowsher prior computed from the T1-weighted image, at iteration number 100, of a brain study using the MR voxel size using the modified system matrix (top), and with the standard system matrix and mMR voxel size (bottom).

III. Results

On the right image of Fig. 1, the NEMA phantom scan was reconstructed with $1.05 \times 1.05 \times 1.1$ mm^3 voxel sizes. It can be seen that the presented method can overcome the problems that appear when reconstructing PET images in a smaller voxel size with a Siddon projector. Fig. 3 shows the contrast and noise for the three smallest spheres where the image reconstructed in the higher resolution matrix achieved a higher contrast of only 1% respect to the reconstructed image with the standard voxel size, therefore no particularly visible difference is observed between them.

In Fig. 4 the reconstructed images of the brain study are shown for $1.05 \times 1.05 \times 1.0$ mm^3 and standard voxel sizes at iteration number 100. The MR-guided MAP reconstructed images are shown in Fig. 5, where the reconstructed image in the MR voxel size shows a better delineation of the structures of the brain and therefore it could be employed to improve the PVC already achieved with the Bowsher prior.

In Fig. 6, a quantitative comparison is drawn between MLEM, MAP MR-guided and MAP MR-guided with PSF modelling reconstructions, for standard and MR voxel sizes. The use of MR voxel size (MRvox) achieved better contrast when comparing same methods, although at the cost of higher noise. How ever, the combination of MR-guided MAP reconstruction with PSF modelling and MR voxel sizes enhances considerably the contrast while reducing and controlling the noise levels and artifacts. The reconstructed images for the different methods at iteration number 300 are shown in Fig. 7.

IV. Conclusions

The method proposed to reconstruct PET images in a higher resolution matrix allows the use of smaller voxel sizes where higher frequencies can be recovered. Accordingly, the reconstructed images with MR-voxel sizes achieved better contrast than with the standard PET voxel size. In addition, when doing MR-guided reconstructions, the anatomical prior can be used in its native size, preserving high frequency details. For this case, the benefits of reconstructing in MR voxel sizes were more significant, showing a better delineation of the edges and a considerable improvement of the contrast. The combination of using MR-guided MAP with PSF modelling and MR voxel sizes proved to be the best way to enhance PVC, while controlling noise and reducing artifacts.

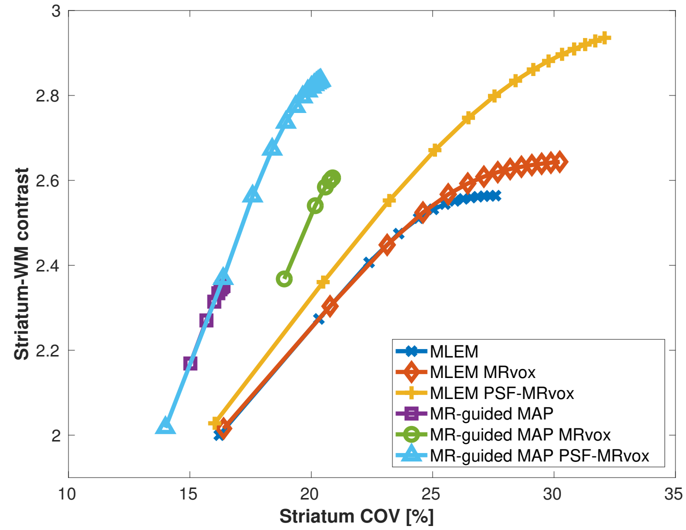

Fig. 6. COV vs contrast in the caudate and putamen as function of the iteration number for MLEM, MAP MR-guided and MAP MR-guided with PSF modelling reconstructions using standard and MR voxel sizes.

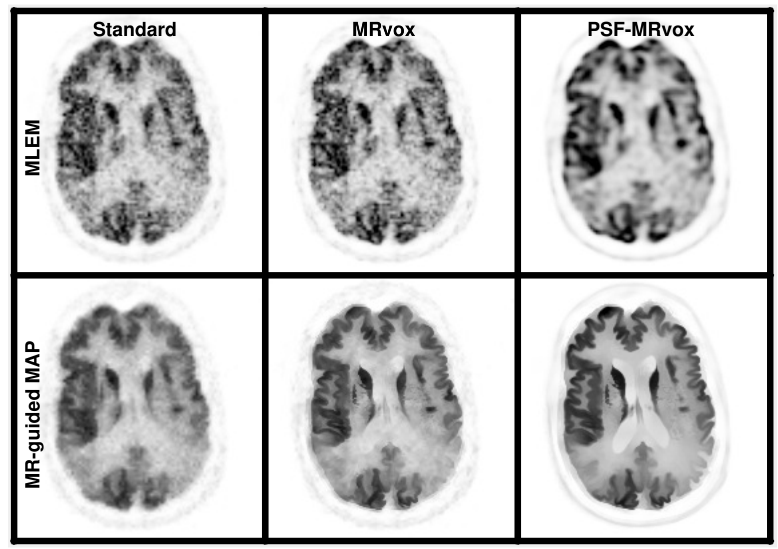

Fig. 7. MLEM, MAP MR-guided and MAP MR-guided with PSF modelling reconstructed images for standard and MR voxel sizes at iteration number 300.

V. Acknowledgments

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