PET ATTENUATION CORRECTION WITHOUT TRANSMISSION SCAN

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ABSTRACT
Attenuation correction (AC) for Positron Emission Tomography (PET) data improves the image quality and enhances quantitative accuracy of radioactivity distribution determination. In dedicated, high resolution breast cancer scanners, this correction would enhance the proper diagnosis in early disease stages. In whole-body PET scanners, AC is usually taken into account by means of transmission scans, either by external radioactive rod sources or by Computed Tomography (CT), considerably increasing the radiation dose to the patient and time needed for the exploration. In this work we propose a method for breast shape identification, by means of PET image segmentation. The breast shape identification will be used for the determination of the AC. For the case of a specific breast PET scanner the procedure we propose should provide AC quite similar to that obtained by transmission scans as we take advantage of the breast anatomical simplicity. Experimental validation of the proposed approach with a dedicated breast PET prototype is also presented. The main advantage of this approach is an important dose reduction since the transmission scan is not required.

KEY WORDS
Medical Imaging, Image Processing, and Signal Processing, Attenuation Correction, Breast PET

1 Introduction

In Positron Emission Tomography (PET) scanners, data attenuation is originated by the loss of true events due to scatter and absorption. The larger the path of the photon in the tissue the higher the likelihood of being absorbed or scattered. One, or even both, of the photons emitted after positron annihilation can be scattered, without complete absorption, and may or may not be detected in another detector. If this occur, additional noise is added to the recorded data. However, PET attenuation effects differ from single photon imaging attenuation effects. In PET, both annihilation photons can be attenuated, and consequently the probability that this occur is much higher than in single-photon imaging. This is true even though the PET photon energy is higher than the typical single photon energy. The most important effect of attenuation is overall count losses, thus reducing the data signal-to-noise ratio (SNR) and worsening the quantification of radioactivity distributions. The increased noise effects can not be solved, while the quantitative accuracy of radioactivity distribution can be recovered with attenuation correction (AC). Another effect of attenuation is to introduce non-uniformities into reconstructed images. This will produce images showing artificially depleted radioactivity deeper in the body, while a “bright” outer contour of the body is observed. In some cases, the radiation emitted in selected directions is much more attenuated than radiation emitted in other directions, which is often observed for the bladder [1]. In these cases, the object will appear elongated along the direction of least attenuation.

Another characteristic of PET attenuation is that usually, at least one of the emitted photons has to travel a substantial amount of tissue thickness, even if the radiation is emitted near the body edge. This implies that, since the total path length is the same for all sources lying on the line that joins two detectors, the probability of attenuation is the same for all such sources, independent of source position. This is true even if the source is positioned outside the body. Therefore, the problem of correcting for photon attenuation in the body is that of determining the probability of attenuation for each line of response (LOR) of the scanner PET. In PET scanners, AC is usually made by means of transmission scans. In these procedures, the count rate from an external source obtained for each line of response is compared with and without (blank scan) the studied body. These measurements can be performed using several different source and detector configurations. The most common approach is to use a long-lived rotating rod positron emitter source ($^{68}$Ge), although single photon emitter sources has been also proposed [2]. However, the impact of transmission sources on the emission data is not negligible and results in bias which may not be acceptable for quantitative analysis, particularly when estimating tracer uptake in small tumors [3], as could be the case in breast studies. With the advent of dual modality scanners capable of acquiring PET and CT data during the same imaging session, there has been considerable effort invested in developing methods to make use of CT data for PET AC, resulting in a significant reduction of the scan time (i.e. two minutes for a torso CT, compared to about 20 minutes for “conventional” transmission scanning) [4]. However, as the energy of CT photons are quite
2 Materials and Methods

2.1 MAMMI Breast PET scanner

Positron emission tomography (PET) scanning is routinely in assessing therapy response, staging and restaging of breast cancer [8, 9]. However, whole-body PET scanners are limited by low geometric sensitivity, low spatial resolution and signal loss due to attenuation in the body, and this reduces their utility in detecting small lesions (< 1 cm) and/or those with low tracer uptake [10]. Since breast cancer patients frequently present lesions smaller than 1 cm in size, this is a significant limitation in the management of primary breast cancer. To overcome this, several groups are working on dedicated breast imaging systems based on high resolution detectors that can be placed close to the breast [11].

Our group has recently developed a dedicated breast PET under the sixth European framework network program within the project Mammography with Molecular Imaging, MAMMI. MAMMI Breast PET scanner [12] consists of 12 detector modules [13] shaping a dodecagon with a scanner aperture of 190 mm. Each detector head contains a scintillating LYSO crystal, a photo-detector and the readout and high voltage power supply electronics (Fig. 1). The most noticeable feature when compared with other PET scanners is the use of $40 \times 40 \times 10 \ mm^3$ monolithic crystals, instead of small pixellated arrays of crystals. Monolithic crystals have been previously used by our group in a small gamma camera for intra surgical applications and in a PET scanner specifically designed for small animal imaging [14]. Position sensitive photomultiplier (PSPMT) H8500 manufactured by HAMAMATSU PHOTONICS was selected as photo-detector. Each PSPMT has 64 anode segments (PADs). All the information from the anode PADs is summarized using two resistive networks so that only five signals (four for the X and Y positions and an additional signal for the Z information) are generated for each event.

MAMMI PET images are reconstructed with a 3D Maximum Likelihood Expectation Maximization (MLEM 3D) [15] with a voxel size of 1 mm$^3$. Dead time, scatter and random events are compensated in the reconstruction. Depth Of Interaction (DOI) is also considered thus making it possible the correct identification of LORS with large incidence angles. DOI correction permits to consider not only coincidences between opposite detectors but also coincidences among one detector and its 7 opposite detectors. Therefore a 170 mm diameter transaxial FOV is available in the MAMMI Breast PET scanner. Detector size restricts the FOV length along axial direction to 40 mm per frame. Here, a precise vertical actuator allows one to sequentially move the detector ring increasing the axial FOV up to 140 mm. The new and dedicated MAMMI breast PET has demonstrated to acquire images with a spatial resolution near 1.5 mm. This scanner has been successfully tested at the pre-clinical level, and a clinical validation is under progress at the Netherlands Cancer Institute in Amsterdam.

2.2 Experimental Measurements

Clinical Acquisitions PET images were acquired 90 min after administration of 180-240 MBq $^{18F}$-FDG with the patients placed in prone position, on a special device that allows the affected breast to hang down offering its maximum volume. Acquisition time was 5 min at each scanner position, resulting in 10-20 min studies (depending on breast length).

Phantom Acquisitions The breast phantom used to validate the correction consisted of a truncated cone of 140 mm
and 30 mm diameters, and 150 mm axial height, see (Fig. 2(a)). Four spherical inserts of 8 mm, 6 mm, 4 mm, and 3 mm in diameter, see (Fig. 2(b)), represented different tumor sizes. The spherical inserts were filled with a 700 μCi/ml FDG solution in all the acquisitions. The breast phantom was filled with a FDG solution which represented the background activity emitted by the breast. Three different background scenarios, with background concentrations of 42 μCi/ml, 89 μCi/ml and 160 μCi/ml were acquired.

2.3 Breast Segmentation

In order to properly calculate AC factors (ACf) we must perform breast identification from PET image. The method we propose is to perform breast image segmentation. In all cases under study, FDG radiotracer was administered to the patients. The procedure we have developed can be divided into a four step process. These consisted of reducing image noise, emphasize the properties of the image we were interested in, separate air and breast regions, and check and correct breast segmentation by considering a priori knowledge of breast anatomy. The images obtained after each step are shown in (Fig. 3).

**Smoothing** Once breast PET image was reconstructed, the uncorrected PET image was median filtered. Median filtering was preferred to average or gaussian filtering in order to prevent from image resolution loss because of the smoothing procedure.

**Equalization** Image values were equalized to emphasize the air to breast transition in the image. Since absolute activity values “seen” in the image depend on the dose injected to the patient, relative thresholds were chosen to separate air from normal breast tissue and normal breast tissue from tumor regions. The lower and upper limits for the equalization procedure were empirically fixed to $P_{30}$ and $P_{75}$ percentiles of the smoothed image.

**Thresholding** Although still being important, threshold selection for the segmentation of air and breast became less critical after the equalization step, see (Fig. 4). Only small
variations in breast size were appreciated if the segmentation threshold was fixed from 30% to 60% of the maximum image value. Segmentation thresholds below 20% could lead to label some noisy air regions as breast.

**Restoration**  
Breast regions with small FDG uptake were air labeled in some cases, see (Fig. 3). This effect was appreciated in some big breasts in the region closest to the chest. To compensate for this undesirable effect a binary processing was performed. In all analyzed images we observed that FDG uptake in the nipple region was slightly higher than in breast region. Therefore, pixels in the nipple region were always labeled as breast. Since breast area increases from nipple to chest we choose the nipple as seed for a selective growing. We walk through the image plane by plane from nipple to chest, for each plane we verified that all voxels assigned to air were also assigned to air in the previous plane. In case a voxel labeled as air in the current plane was labeled as breast in the previous plane, we relabeled the voxel as breast. This step permitted us to minimize the underestimation of ACf’s caused by the wrong labeling of breast regions with low FDG uptake.

2.4 Attenuation Correction

Once we have properly labeled air and breast voxels in the reconstructed image, AC is relatively easy to be derived. For each LOR, the total path length in air ($L_{air}$) and within breast ($L_{breast}$) are obtained. Linear attenuation coefficients at 511 keV for breast ($\mu_{breast} = 0.098 \text{ cm}^{-1}$) [16] and air ($\mu_{air} = 0.0 \text{ cm}^{-1}$) are assigned to breast and air regions respectively. ACf for each LOR are then calculated as:

$$ACf_{LOR} = \exp(\mu_{breast}L_{breast}) \ast \exp(\mu_{air}L_{air})$$

and the total counts obtained for each LOR is then increased by a factor $ACf_{LOR}$. MLEM 3D reconstruction is then repeated, now considering the corrected counts for each LOR.

Since segmented image with the linear attenuation coefficients constitutes the attenuation map, and the contribution of each voxel to each LOR integral is considered in the MLEM system matrix. Practical implementation of $ACf_{LOR}$ in (1) was done by projecting the segmented image. Projection step in MLEM provides us with the sum of attenuation coefficients along each LOR. Therefore the corresponding ACf for the LOR is straightforward calculated by exponentiation.

3 Results

Accurate quantification of AC performance requires large, *i.e. ~ 2 hours* long, PET acquisitions [6]. However our main concern was to validate the feasibility of the proposed AC in daily routine 5 min long PET acquisitions. Large PET measurements improve signal to noise ratio (SNR) leading to straightforward segmentation procedures the performance of which can be strongly affected when the SNR diminishes. Since the purpose of this work was to develop and validate the effect of the proposed AC in daily routine PET images we selected those measurements which were the most similar to clinical images. The drawback of the decision we made is that only a qualitative discussion on the performance of the proposed AC is possible with the measurements presented in this work.

The calculated AC procedure under study was experimentally validated with the breast phantom described above (Fig. 2(a)). MAMMI data images were acquired and with and without AC. In all three cases, *i.e. 42 $\mu$Ci/ml, 89 $\mu$C/ ml and 160 $\mu$Ci/ml as background levels*, the reconstructions without AC show an underestimation of the activity distribution. As it was expected, attenuation effect introduces non-uniformities into reconstructed images, showing artificially depleted radioactivity deeper in the breast phantom, while a bright outer contour is observed in the uncorrected image (Fig. 5).
In order to quantify the AC effect on the MAMMI final images, two different transverse emission profiles of the breast phantom were considered (Fig. 6) and (Fig. 7). These profiles were obtained for a uniform background level of $160 \mu Ci/ml$. However no significant differences have been found for the other two background levels analyzed. In both cases (cold and hot profile) the AC we propose significantly reduces the non-uniformities into reconstructed image (depletion in the inner region while overestimate at the phantom edges). We have divided the line profiles into different regions and we have calculated for these regions the global mean AC factor, i.e. the ratio between the total integrated activities in the selected regions, with and without AC Table 1 and Table 2. If we consider the region that covers the whole line profile, a global attenuation factor of about 3 is obtained, regardless the line profile selected (hot or cold profile) (Table 1 and Table 2, region III). Moreover, at the phantom edges region a mean attenuation factor around 2.1 is obtained, being valid for both profiles, within statistical fluctuations. As it was expected, higher global attenuation factors are obtained in the inner region (see Table 2, regions IV and V). Despite the simplicity of the proposed model, we have observed that it is able to reduce significantly the non-uniformities caused by attenuation, increasing at the same time the global activity derived from the MAMMI breast PET images.

### 3.1 Activity Profiles

Table 2

<table>
<thead>
<tr>
<th>ROI</th>
<th>Uncorr.</th>
<th>Corr.</th>
<th>Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>I</td>
<td>$9.16 \times 10^{-4}$</td>
<td>0.0019</td>
<td>2.09</td>
</tr>
<tr>
<td>II</td>
<td>0.0011</td>
<td>0.0026</td>
<td>2.25</td>
</tr>
<tr>
<td>III</td>
<td>0.0074</td>
<td>0.0229</td>
<td>3.08</td>
</tr>
<tr>
<td>IV</td>
<td>0.0016</td>
<td>0.0058</td>
<td>3.64</td>
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<tr>
<td>V</td>
<td>0.0014</td>
<td>0.0048</td>
<td>3.52</td>
</tr>
</tbody>
</table>

### 3.2 Interslice Uniformity

Qualitative analysis of the proposed AC method for the MAMMI breast PET has been shown in previous section. As it is visually appreciated in (Fig. 5), the expected flat activity distribution is recovered after the application of the method. Since the quantification with the MAMMI prototype is still under progress, accurate quantitative analysis of the AC is beyond the scope of this work. However, a basic statistical description of the performance of the AC is possible with the measurements here presented.

As seen in previous section, the effect of the AC on the activity distribution depends on the depth of the voxel in the object under study. Therefore, to appreciate the AC correction ten 5 mm height annular Regions Of Interest (ROI), depicted in (Fig. 8), were defined. Inner and outer
The ROIs being evaluated corresponded to regions in the background of the PET image. In order to minimize the contribution of noise in the evaluation of the AC, differences among average activity measured in the ROIs of each slice were analyzed. Background activity was uniform in the whole phantom. Therefore, differences between activities measured in different ROIs were expected to diminish when applying the AC. To compare the similarity among measured activity when applying the AC, differences between extreme mean activity values measured in the ROIs of each slice were calculated according to (2).

\[ U = \left( \frac{\max(R_i) - \min(R_i)}{\max(R_i) + \min(R_i)} \right) \times 100 \]  

(2)

where \( R_i \) was the average activity measured in the \( i \)-th ROI of each slice. \( \max R_i \) and \( \min R_i \) were respectively the maximum and minimum of the ROI mean activities in each slice.

A remarkable reduction in differences between mean activity measured in different ROIs, when the AC was applied, is reflected in Table 4. In all background scenarios considered in this work, differences among average activity measured in the ROIs of the Up slice dropped from about 75 % to 10 %. For the Down slice the U values were significantly lower (~ 40 %). However, when the AC correction was taken into account these differences dropped to 10 %, similar to that observed for the Up slice’s ROIs.

Differences among average activity in Up slice were twice as much as that observed in Down slice, when the AC correction was applied similar differences were appreciated in Up and Down slices. Larger differences were measured in the Up slice because of the larger size of the phantom in the Up slice. Due to the larger size of the phantom in the Up slice the effect of the AC was stronger.

### 4 Conclusion

In this work we present the experimental validation of PET attenuation correction based on breast segmentation of uncorrected PET images acquired with the MAMMI breast PET prototype. Actual breast acquisitions were used to verify the correct breast identification from uncorrected PET images. The performance of the attenuation correction is validated with a breast phantom with known uniform activity distribution. In (Fig. 5(a)) is shown the fake activity distribution retrieved when the AC was not considered. And how the proposed AC restores the true uniform activity distribution present in the phantom (Fig. 5(b)).

The goal of this work was the validation of the proposed AC in actual clinical applications. Therefore, phantom measurements used in this study were aimed at reproduce actual clinical measurements. They did not allow us an accurate quantitative analysis because measurements with high SNR are required for such analysis. However, the measurements presented in Table 4 reflect a strong reduction, from 70 % to 10 %, in the variability of activity measurements in different ROIs of the uniform phantom when the proposed AC was considered.

### Acknowledgement

This study has been partially funded by European Commission under the project “MAMmography with Molecular Imaging MAMMI. FP6-2005-LIFESCIHEALTH-7”.

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**Figure 8. Regions Of Interest defined to analyze the proposed AC**

**Table 3**

<table>
<thead>
<tr>
<th>ROI</th>
<th>Up max</th>
<th>Up min</th>
<th>Down max</th>
<th>Down min</th>
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</thead>
<tbody>
<tr>
<td>ROI-1</td>
<td>124</td>
<td>95</td>
<td>104</td>
<td>80</td>
</tr>
<tr>
<td>ROI-2</td>
<td>95</td>
<td>70</td>
<td>80</td>
<td>60</td>
</tr>
<tr>
<td>ROI-3</td>
<td>70</td>
<td>45</td>
<td>60</td>
<td>40</td>
</tr>
<tr>
<td>ROI-4</td>
<td>45</td>
<td>20</td>
<td>40</td>
<td>20</td>
</tr>
<tr>
<td>ROI-5</td>
<td>20</td>
<td>0</td>
<td>20</td>
<td>0</td>
</tr>
</tbody>
</table>

**Table 4**

<table>
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<tr>
<th>Activity</th>
<th>Background</th>
<th>UnCorr</th>
<th>Corr</th>
<th>UnCorr</th>
<th>Corr</th>
</tr>
</thead>
<tbody>
<tr>
<td>HIGH</td>
<td>78.5</td>
<td>10.5</td>
<td>41.2</td>
<td>11.0</td>
<td></td>
</tr>
<tr>
<td>MEDIUM</td>
<td>72.0</td>
<td>8.4</td>
<td>43.0</td>
<td>10.4</td>
<td></td>
</tr>
<tr>
<td>LOW</td>
<td>81.1</td>
<td>11.6</td>
<td>53.1</td>
<td>4.2</td>
<td></td>
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</tbody>
</table>

**Table 3**

<table>
<thead>
<tr>
<th>ROI</th>
<th>ROI diameters (in mm) used to analyze the performance of the AC.</th>
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<tbody>
<tr>
<td>ROI</td>
<td>Up max</td>
</tr>
<tr>
<td>-----</td>
<td>--------</td>
</tr>
<tr>
<td>ROI-I</td>
<td>124</td>
</tr>
<tr>
<td>ROI-II</td>
<td>95</td>
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<tr>
<td>ROI-III</td>
<td>70</td>
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<tr>
<td>ROI-IV</td>
<td>45</td>
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<tr>
<td>ROI-V</td>
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References


