Component based normalization method for rotating dual head PET scanner

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Abstract. Component based normalization is a well-established method to calculate correction factors for unbiased and reliable PET reconstruction. Several methods have been studied and validated for cylindrical PET scanners. In this work we adapted a method already presented for cylindrical scanners to rotating dual head PET scanners. The model included corrections for detector efficiency, axial and transaxial geometric effects, crystal interference and attenuation corrections. Results from a simulated realistic dual head PET showed that the adaptation is valid. The images are significantly improved in terms of homogeneity, resolution and background contribution.

1. Introduction
Reliable image reconstruction in Positron Emission Tomography (PET) is only achieved after proper correction of the sensitivity of different lines of response (LOR). Sources of these differences rely in the variations in detector efficiency, geometric effects, mashing of neighbouring LORs etc [1,2].

In direct calculation of the normalization factors, a linear, planar or cylindrical source is used to measure the efficiency of each LOR [1-5]. The main drawback of direct normalization is the need for long acquisitions. On the other hand, component (or model) based normalization, by decomposing the correction factors to several parameters, is possible to accurately estimate the correction factors with relatively shorter acquisitions.

Usually, the parameters are grouped in the detector intrinsic properties and scanner’s geometry. In addition, a series of correction factors related to the timing properties of the scanner have been proposed (e.g. dead-time correction, random correction etc) [1-5].

In this study we used a simulated model of a rotating dual head PET scanner, installed in our laboratory. The GATE simulation toolkit was used [6]. The aim was to investigate the application of well validated component based normalization algorithms from cylindrical PET scanners [7,8]. We used a planar source and no rotation in order to calculate the correction factors and extrapolated analytically the resulted factors the over 180° angular coverage. In order to validate our calculations we used two phantoms a uniform cylindrical and an assembly of 21 capillary cylinders evenly spaced every 1 cm.

2. Material and Methods

2.1. GATE simulations
The geometry of a dual head PET scanner, installed in our laboratory, was used as a prototype for the development of the simulation’s geometry. The detailed scanner’s presentation and evaluation as well as the validation of the simulation models have been presented previously [9,10]. Briefly the scanner is based on two Hamamatsu H8500 PSPMTs which are coupled with two 20×20 LSO:Ce crystal matrices, with crystal size 2×2mm².

The two heads were rotated in unison for 180°, with 10° step and the acquisition duration on each step was 60 sec. The head separation distance was 5 cm and the 176Lu intrinsic radioactivity was taken into consideration. The low level threshold was set to 200 keV and the energy window was set to 350-650 keV.

The simulated data were binned into 3D sinograms, with bin size ~0.3 mm. The angle of rotation and the angle of the detector with respect to the center of the head’s rotation were summed for the calculation of azimuthal angle of each LOR, as illustrated in Figure 1(a). The 88 angular positions reflect the sum of events over all detectors located every ~2° during in each step of rotation, as illustrated in Figure 1(b).

![Figure 1](image.png)

**Figure 1.** (a) For calculation of the azimuthal angle of each event the angle of the detector with respect to the centre of the head was calculated. (b) The azimuthal angle of the detector was added to the rotation angle of the heads for the calculation of the LOR’s azimuthal angle.

### 2.2. Phantoms

In this study two phantoms were used. First a homogeneous water cylindrical phantom (Ø 4cm) coupled with a ⁶⁷Ga ion source. Second a phantom assembled by 21 cylindrical tubes made from water, evenly spaced every 1cm. The diameter of each capillary was 1.1 mm in order to simulate the size of regular capillary tubes.

The data were reconstructed using a full 3D MLEM algorithm implemented in STIR [11] using 5 subsets after 5 subiterations. The mean depth of interaction inside the crystal was taken into account. But no arc-correction, random, scattered and dead-time corrections were performed.

### 2.3. Normalization components

For the calculation of the normalization components a 7×7 cm² planar phantom located in the centre of the FOV, was simulated only for one rotation step. The simulated data then were extrapolated using geometrical calculations for all rotation steps. The low hardware threshold was set high to 200 keV with the energy window from 350-750 keV. This configuration was chosen in order to minimize the contribution of low energy photons to system’s dead-time while. In addition, it has been shown that in small animal imaging the contribution of scattered photons down to 350 keV is not significant, while the system benefit from the higher count rate [12].

The normalization components that were included in the model were the axial and transaxial geometric factor, block sensitivity correction, crystal interference, detector efficiency and attenuation correction. Even though in GATE detectors have uniform efficiency it has been shown that a correction should be applied [8].
3. Results
The comparison between the non-normalised and normalized images from the cylindrical and the 21 capillaries phantom, are displayed in Figure 2(a-d). In Figure 2(e-h), one may see their corresponding profiles, which represent an average of the three central rows. As one can see in Figures 2(a) and (b) the homogeneity inside the cylinder is strongly improved. Near the edge of the cylinder is observed a depth of iteration artefact. Furthermore, a strong contribution of scattered photon is observed, the use of STIR’s scatter correction function should help reduce this effect [13].

Regarding the 21 capillaries phantom, before the normalization the peaks for each capillary are higher toward the centre of the FOV and a strong background area is located around the centre. After normalization we get similar intensity values for each capillary, the background in the centre is strongly reduced and a resolution recovery is observed. The shape of the capillary’s profile towards the edges of the FOV is gradually deformed, which is a common effect due to the parallax effect.

![Figure 2](image-url)

**Figure 2.** Non-normalised image from the cylindrical (a-b) and 21 capillaries phantom (c-d) and their corresponding profiles (e-f) and (g-h). The profiles are the average of the three central rows.

4. Conclusion and Discussion
Normalization for PET data from a dual head planar PET may be performed using directly adapted algorithms from cylindrical PET scanners. The acquisition of the normalization data can be performed only for one acquisition step using a planar source and then use geometric calculations to calculate the correction factors for the rest of the rotation steps.

Acquisition of data for one rotation angle significantly reduces the normalization scan time but is susceptible to errors like error of rotation and errors in the placement of the phantom.

By correcting for detector efficiency, axial and transaxial geometric effects, block sensitivity and attenuation the reconstructed images are strongly improved near the centre, in terms of homogeneity, intensity and background noise. Introduction of more correction components in our model will further improve the quality of the images on the expense of additional complexity of the algorithm.

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