DIGITALIZATION OF A MAMMOGRAPHIC PHANTOM VIEW THROUGH A MONTE CARLO SIMULATION

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Abstract: The aim of this study is the development of a methodology to reconstruct the image of detail defects from a mammographic phantom employed in quality control testing. The MCNP-4c2 code has been used to model the mammography unit and the CIRS 11A (MAMMO PHANTOM SP01) mammographic phantom. This phantom is made of poly-methyl methacrilate and contains a reference point, contrast and resolution detail targets, groups of microcalcifications and fibres. The image in a mammogram is produced by the unscattered fraction of the emitted X-rays, and the scattered fraction, not useful to imaging purpose. The unscattered fraction has been scored with a photon flux tally (F5), employing a point detector matrix below the mammographic phantom. The MCNP code has been modified and recompiled to let a large number of tallies per input file. Point detectors have an inherent singularity with the square distance from the point of interest to the detector. Therefore, the use of a point detector tally to score both fluxes has a well behaviour only with the uncollided fraction (variance zero). Instead of scoring with a point detector the scattered fraction, it has been used a track length estimate tally (F4), modeling a lattice of rectangular cells. Photons have been forced to reach each one of the lattice cells with a matrix of DXTRAN spheres, centred upon them. The characteristic curve of the mammographic film has been obtained throughout several experimental measures of grey levels with different aluminium layers and Monte Carlo simulations, obtaining a response function between output air kerma and grey level. The simulated images agree with the range of values, indicating that this method will be suitable for training purposes or phantom designing.
1. Introduction

Mammography is a non-invasive technique used in the diagnosis of breast diseases in women. A breast screening program employs mammography in the early detection of abnormalities in the female breast, mainly microcalcifications, which could develop a breast carcinoma. Quality control testing in mammography is developed to achieve the image quality required to detect these abnormalities, ensuring the minimum risk to screened women.

Until now, many studies have been centred in the calculation of the absorbed mean glandular dose to the breast in mammography [1-3], as an estimator of the induced carcinogenesis. During last decade, Monte Carlo methods have been also extended to improve the image quality vs. patient dose ratio [4]. Monte Carlo methods have been applied before in image synthesis [5]. However, in this specific field, few studies have been presented [6-7].

In this paper the authors have developed a method to synthesize by Monte Carlo the images obtained in the exposure of a mammographic phantom. The aim of the simulation has been the reconstruction of the grey-scale images obtained from the distribution of photons on the mammographic film. The code used to simulate the images has been the MCNP, version 4c2 [8]. MCNP is a Monte Carlo transport code used for neutron, photon, electron or coupled neutron/photon/electron transport. The code is useful to model a large variety of geometry, and it allows different variance reduction techniques, that force particles to travel to the important regions, preventing an over or underbiased result.

In section 2, the authors present the methodology to achieve the reconstruction of the images of the mammographic phantom. These images have been synthesized assuming a discrete energy spectrum, emitted by the mammography unit X-ray tube. The results of the simulation, the limitations of the model and its range of validity have been discussed in section 3. The final section draws the conclusions and the directions of future research.
2. Methodology

2.1 Introduction

The photon flux, $\Phi$, can be scored by the MCNP code by means of point detectors or track length estimate tallies. The flux $\Phi_{ij}$ that reaches the pixel $(i,j)$ per emitted particle, due to a discrete spectral source, can be decomposed in two components, a scattered and an unscattered one, as showed

$$\Phi_{ij} = \frac{1}{N^0} \sum_{n=1}^{N^0} \phi^0_{ij}(n) + \frac{1}{N^c} \sum_{n=1}^{N^c} \phi^s_{ij}(n)$$

(1)

where $\phi^0_{ij}$ is the unscattered flux fraction or primary beam, $\phi^s_{ij}$ is the scattered flux fraction, both per emitted source particle and $N^0$ and $N^c$ are the number of photon tracks employed in the uncollided and collided Monte Carlo simulation, respectively.

2.2 Calculation of the primary flux with a point detector tally

If a particle is emitted by the source, a score is made to the point detector tally. The unscattered flux contribution $\phi^0_{ij}$ in the detector $(i,j)$ per emitted particle, due to a photon track with energy $E_n$ and weight $w_n$ has been estimated as

$$\phi^0_{ij}(n) = w_n E_n \frac{p^0(\theta_{ij},\varphi_{ij}) \exp(-\lambda_{ij}(E_n)}}{R_{ij}}$$

(2)

where $p^0(\theta_{ij},\varphi_{ij})$ = the value of the probability density function at $\theta_{ij}$ and $\varphi_{ij}$, the polar and azimuthal angles, respectively, between the source and the detector $(i,j)$. $\exp(-\lambda_{ij}(E_n))$ = the attenuation of the beam at energy $E_n$. $R_{ij}$ = the distance from the source to the detector $(i,j)$

The attenuation coefficient $\lambda_{ij}$, in (2), is calculated by the code as

$$\lambda_{ij} = \oint \Sigma_i(s_{ij},E) ds_{ij}$$

(3)

where $\Sigma_i(s_{ij},E)$ is the macroscopic total cross-section at energy $E$ in the point $s_{ij}$. The line integral is extended from the source or collision point to the detector $(i,j)$, through the different materials.

2.3 Calculation of the scattered flux with a track length estimate tally

The scattered fraction of the total photon flux has been estimated with a track length estimate tally. The MCNP code calculates this tally with the cumulative contribution of each particle that crosses the cell under study. The scattered flux contribution $\phi^s_{ij}$ in the detector cell $(i,j)$ per emitted particle, due to a photon track with energy $E_n$, and weight $w_n$ has been obtained by a Monte Carlo simulation as
\[ \phi_{ij}(n) = w_n \frac{E_n T_{ln}}{V_{ij}} \]  

(4)

where \( T_{ln} \) is the track length (transit time per photon velocity) and \( V_{ij} \) is the volume of the cell \((i,j)\).

In the case of an analogue simulation, the weight of all particles that reach the cell is unity. However, the number of photon tracks that make score in a tallied cell is too low. Therefore, some variance reduction techniques are needed to emit and transport photons to these cells.

2.4 MCNP-4c2 modelling

2.4.1 Geometry and materials

Two different MCNP-4c2 models have been developed to simulate the mammographic phantom during quality controls testing in the Valencian Breast Cancer Early Detection Program (VBCEDP). These models are comprised of a spectral point source, a mammographic phantom and the image receptor system, at a constant focus-film distance (FFD) of 60 cm. The image system has been simplified to represent only the air between the mammographic phantom and the film. The antiscatter grid has not been simulated, because the mammographic cassette was placed upon the breast support. The phantom is an approximation to the CIRS 11A (MAMMO PHANTOM SP01). This phantom is made of poly-methyl methacrilate and it contains a reference point, contrast and resolution detail targets, groups of microcalcifications and fibres. Fig 1 shows the digitized scanned image of the phantom modelled with MCNP code.

![Image](image.png)

**Fig 1.** Digitized scanned image of the CIRS 11A (MAMMO PHANTOM SP01), at 28 kV and 42 mAs (8 bits/pixel)

2.4.2 Source specifications

The source spectra have been obtained from the Catalogue of Diagnostic X-Ray Spectra and other Data (IPEM) [9] at 28 kV for a molybdenum tube anode with 0.03 mm molybdenum and 1 mm beryllium filter.

2.4.3 Tally specifications

A lattice of rectangular cells has been modelled under the phantom to represent each one of the pixels which will form the final image. A F5:p tally has been used to obtain the primary beam in the region of interest. The detectors have been placed on the centre of each one of the lattice cells. This tally is the most appropriate to score contributions from the uncollided fraction of the X-ray beam. However, the \( R_{ij}^2 \) term in the denominator in (2) produces a singularity that causes the variance of the scattering fraction to reach theoretically infinity, if a collision occurs near the
Fig 2. Scheme of the lattice cells used to reconstruct the mammographic phantom image. If a collision occurs, a pseudo-particle is transported upon the DXTRAN sphere and crosses the cell of interest.

detector. For this reason, a F4:p tally has been employed to calculate the scattered fraction of the X-ray beam that reaches each one of the cells of the lattice.

2.5 Estimation of Variance Reduction Parameters

2.5.1 Source biasing

It has been considered a biasing in the energy and in the angle of emission of initial photons. The Catalogue of Diagnostic X-Ray Spectra and other Data [IPEM] provides spectra discretized by energy groups.

The spectrum has been implemented with a $sp$ card and a $sb$ card. The $sb$ card is used to provide a probability distribution for sampling and the weight of each source particle is adjusted to compensate for this bias. The weights of the initial particles, $ω_0(k)$, are calculated with the expression

$$w_0(k) = \frac{ψ(k)}{\sum_k ψ(k)}$$

where $ψ(k)$ is the photon flux at energy $k$ supplied by the Catalogue in photons per mm$^2$ and tube load (mAs) at 75 cm. These weights are the entries of the $sb$ card.

The biasing in the angle of emission does not affect the weight of the initial photons, because particles emitted in other directions are annihilated (weight zero), and the total weight is conserved.

If we assume an azimuthal symmetry, the probability density function (PDF) of emitting a photon to the point $(i,j)$, $p^0(θ_y,φ_y)$, is calculated, as
where $\theta_{\text{max}}$ is the maximum anisotropic angle of source emission. This function is used by the code to sample the polar and the azimuthal angles, but it is employed in () to estimate the value of the unscattered flux in the point $(ij)$. 

### 2.5.2 Transport biasing

The system has been bounded by an infinite region of zero photon importance, annihilating particles that escape from the system, which not contribute to tallying purposes.

A DXTRAN angle biasing technique has been used, binding the geometry of the model, in order to stream particles to form the final image. Some DXTRAN spheres have been placed, upon the lattice cells. Each collision produces a pseudo-particle which is transported deterministically to the surface of the outer sphere. The radius of the inner sphere has been chosen to be zero. The weight of this particle is modified with the attenuation between the collision point and the DXTRAN sphere, and with the probability of scattering to the outer surface of the sphere, diminishing considerably, due to its small size, as showed in

$$w_{\text{DX}} = w_0 P(\theta_{ij}, \phi_{ij}) \exp(-\lambda_{ij})$$

where $P(\theta_{ij}, \phi_{ij})$ is the PDF of scattering with a polar and an azimuthal angles $(\theta_{ij}, \phi_{ij})$, depending on the kind of collision (coherent or incoherent).

To prevent that these low weight particles play Russian roulette, with a high probability to be killed, the weight cutoff has been eliminated. Once the pseudo-particle is transported to the sphere surface, it is again converted to a real one, but now, the proximity to the cell increases the probability of contributing to scoring.

The radius of the DXTRAN sphere has been estimated to the expression

$$R_{\text{DX}} = n_{\text{DX}} \frac{h}{2}$$

where $h$ is the pixel pitch and $n_{\text{DX}}$ is the number of pixels per each DXTRAN sphere.

Fig. 2 represents a scheme of the image region of the MCNP models. It shows the transport of a pseudo-particle, created in the collision point, to the DXTRAN sphere, the transformation to a real particle, and the flight to the cell of interest.

### 2.6 MCNP4c2 simulation and implementation

The code was compiled and run on a SGI Altix 3700 with 48 Intel Itanium II processors and Linux RedHat OS. The patchf file was modified to change the default values of NTALMX (maximum number of tallies), MXDT (maximum numbers of detectors), MXDX (maximum number of DXTRAN spheres) and MLGC (size of logical arrays for complicated cells) parameters, to allow the execution of the input file. Table 1 shows the default values and the values of these variables, used in the simulation.
Table 1. MCNP-4c2 internal variables

<table>
<thead>
<tr>
<th>Variable</th>
<th>Description</th>
<th>Default</th>
<th>Simulation</th>
</tr>
</thead>
<tbody>
<tr>
<td>NTALMX</td>
<td>Maximum number of tallies</td>
<td>99</td>
<td>5000</td>
</tr>
<tr>
<td>MXDT</td>
<td>Maximum number of detectors</td>
<td>20</td>
<td>5000</td>
</tr>
<tr>
<td>MXDX</td>
<td>Maximum number of DXTRAN spheres</td>
<td>10</td>
<td>1000</td>
</tr>
<tr>
<td>MLGC</td>
<td>Size of logical arrays for complicated cells</td>
<td>1000</td>
<td>1500</td>
</tr>
</tbody>
</table>

2.7 Image synthesis

The characteristic curve, which converts the output air kerma under the phantom to a grey-scale level, has been obtained employing several combinations of aluminium filters, at 28 kV and 5 mAs. The resulted mammogram has been scanned by an AGFA DUOSCAN F40 (1200 x 2400 dpi) at 8 bits (black = 0 and white = 255). The output air kerma has been calculated with a Monte Carlo simulation, reproducing in the model the experimental conditions. Fig 3 shows the digitized scanned image per aluminium step, whereas the fig 4 presents the grey-scale level of the digital image, as a function of the relative distance to the first aluminium filter. In table 2, it is summarized and presented the results derived from the Monte Carlo analysis, and the real values of optical density (OD), and grey-scale level.

Table 2. Measured and simulated values with MCNP-4c2

<table>
<thead>
<tr>
<th>Aluminium (cm)</th>
<th>OD</th>
<th>Grey-Scale Level</th>
<th>Air Kerma (MeV g⁻¹ photon⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>3.5</td>
<td>0</td>
<td>8.21 × 10⁻⁶</td>
</tr>
<tr>
<td>0.5</td>
<td>3.37</td>
<td>25.30</td>
<td>2.22 × 10⁻⁸</td>
</tr>
<tr>
<td>1</td>
<td>2.69</td>
<td>35.07</td>
<td>1.48 × 10⁻¹⁰</td>
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<tr>
<td>2</td>
<td>0.92</td>
<td>158.05</td>
<td>2.98 × 10⁻¹²</td>
</tr>
<tr>
<td>3</td>
<td>0.33</td>
<td>216.56</td>
<td>7.02 × 10⁻¹⁴</td>
</tr>
<tr>
<td>5</td>
<td>0.24</td>
<td>227.09</td>
<td>4.83 × 10⁻¹⁷</td>
</tr>
</tbody>
</table>

Fig3. Digitized scanned image, reproducing the experimental conditions, with five aluminium filters (28 kV and 5 mAs, 8 bits/pixel)
3. Results & discussion

Fig. 5 presents different reconstructed images of a semi-cylindrical phantom, through Monte Carlo simulation, with a step-wedge target (100 percent adipose/left and 100 percent glandular/right). Fig 6 shows the resulted digital image, where we can see the contrast detail targets, the line pair test horizontal target and the step wedge. This image is the result of a first approach to the reconstruction of mammographic films through Monte Carlo techniques. The synthesized image does not represent precisely the scanned film of the mammographic phantom. The grey-scale distribution of the image produced by simulation does not have the same contrast grading of the real image. Further studies are necessary to obtain more accurate characteristic response functions, for several tube loadings and voltages, focusing in several regions of the phantom, and obtaining the point spread function. A MTF analysis should be performed to compare the horizontal and vertical resolutions of the phantom.
The image synthesis has been performed in a IA-32 CLUSTER 1350, formed by 57 computation, 2 storage and 1 control nodes, with 2 Intel® Xeon® processors per node and Linux RedHat OS.

4. Summary and conclusions

The method has been applied to obtain reconstructed images by Monte Carlo techniques. The process is divided in two steps, the calibration of the characteristic curve and the simulation with the MCNP-4c2 code. The simulation follows two different ways, depending on the contribution (unscattered or scattered flux). Two tallies have been used, with different variance reductions techniques, focusing on the particularities of each one, to calculate the output air kerma.

In future work, the MCNP model will be completed, including the microcalcifications and fibers, and the vertical resolution target. Furthermore, due to that MCNP allows modelling different compositions and geometries, the generated mammograms could be used in the evaluation of the computer-aided techniques for the early detection of microcalcifications in breast.

5. Acknowledgements

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6. References