Breast CT image simulation framework for optimisation of lesion visualisation

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Abstract—Although X-ray mammography is the gold standard technique for breast cancer detection, it suffers from limitations due to tissue superposition which could either obscure or mimic a breast lesion. Dedicated breast computed-tomography (BrCT) represents an alternative technology with the potential to overcome these limitations. However, this technology is still under investigation in order to study and improve certain parameters (e.g. dose, scattered radiation, etc.). In this work, an image simulation framework is proposed to generate realistic BrCT images and spectral imaging analysis is explored to enhance the contrast of breast lesions. Results illustrated an improvement in contrast between 5 and 10% when the final image is reconstructed using X-ray photons with energies between 21 and 30 keV, in comparison with the reconstructed image from the polychromatic energy spectrum recorded within the image receptor.

Index Terms—Dedicated breast CT, contrast improvement, detector technology, image simulation, spectral analysis.

I. INTRODUCTION

Breast cancer is the second most commonly diagnosed cancer worldwide [1]. Although two-dimensional (2D) planar X-ray mammography is currently the most widely accepted modality for early breast cancer detection, it suffers from the superposition of three-dimensional (3D) anatomical structures onto the 2D projected image which can mimic the appearance of a breast lesion where it does not exist or it may obscure real breast lesions [2]. These effects contribute to a reduction of both the sensitivity and specificity of this technique.

Dedicated breast computed tomography (BrCT) is an emerging technology which generates tomographic 3D breast images, showing potential to overcome the tissue superposition limitations found in 2D planar mammography [3]. In dedicated BrCT, the patient lies in prone position on a bed table placing one breast into a scanning aperture. Underneath the bed table, the pendant breast is imaged using an X-ray tube and a detector which are rotated 360° around the breast. This allows to produce images with depth information of the breast, which may provide radiologists with more information to identify breast lesions. In spite of the reduction of the tissue superposition effects when using dedicated BrCT, further research is needed to investigate an optimal approach that reduce the radiation induced to the patient, as more X-ray projections are required in comparison with 2D planar X-ray mammography.

This is addressed here with the development of a fast non-Monte Carlo (MC) image simulation tool which can be used to rapidly compare BrCT simulated images from modelling different detector technologies, beam qualities, dose, etc. Such an approach may have applications in pre-clinical imaging.

II. IMAGE SIMULATION FRAMEWORK

In this work, an image simulation framework was developed to produce synthetich BrCT images. A diagram illustrating the proposed image simulation framework is shown in Fig. 1.

Firstly, the geometry of both the BrCT scanner and the breast phantom are specified in the image simulation framework. The BrCT scan geometry is described as:

1) source-to-isocentre distance (SID), where the isocentre corresponds to the central axis of the breast phantom;
2) iso-centre-to-detector distance (IDD);
3) detector’s dimensions;
4) detector’s pixel size.

Furthermore, a breast phantom (or test object) is inserted into the proposed image simulation framework using its mathematical 3D model, where each voxel represents a type of breast tissue. Once this preliminary information is defined and the breast phantom inserted, photon paths from an infinitesimal X-ray point source to the centre of each detector pixel \((x, y)\) is calculated using Siddon’s ray tracing methodology [4]. The intensity observed at each pixel, for each energy component

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E and a given projection, is then calculated analytically considering an idealised detector using Beer-Lambert’s law:

\[ I'(x, y; E) = e^{\sum_i (-\mu_i(E) t_i(x, y))}, \] (1)

where \( \mu_i(E) \) represents the linear attenuation coefficients of the different materials along the photon path \( t_i(x, y) \) between the X-ray source and detector pixels. For a particular geometry, this operation is calculated only once for energies \( E \) from 1 to 140 keV, in steps of 1 keV.

Then, this idealised primary projection \( I'(x, y; E) \) is ‘degraded’ to model a specific detector technology (e.g. CdTe photon counting detector, or a CsI energy integrating detector) using the detector efficiency \( \epsilon(E) \), energy resolution \( G(E) \) and noise characteristics \( N \). Finally, the energies are weighted accordingly, using the energy spectrum \( w(E) \) emitted from the X-ray tube, to simulate a given beam quality input using the X-ray spectra simulation tool described by Siewerdsen et al. [5]. Therefore, the final primary projection \( I(x, y) \) is calculated as:

\[ I(x, y) = \left( \sum_{E=0}^{E_{\text{max}}} \left( I'(x, y; E) \times w(E) \times \epsilon(E) \right) \ast G(E) \right) + N. \] (2)

Furthermore, if the BrCT scanner studied is expected to have a large scatter fraction in the detector (e.g. flat panel detector), scattered radiation can be added at this point. Due to the large computational time required by MC simulations, a kernel-based approach is proposed to estimate the scattered radiation [6]. On the other hand, when simulating a slat detector geometry (e.g. helical breast CT), the scatter field can be assumed to be negligible [7].

The above process is followed to calculate a given projection of the breast phantom. However, a large number of BrCT projections are needed in order to generate an image. Thus, this methodology is repeated while rotating the breast phantom at a number of angles across 360°. Once all the BrCT projections have been calculated, a sinogram is produced, from which reconstructed images can be generated using a reconstruction method such as filtered back projection. As will be seen in the results section, several BrCT images have been reconstructed for two breast phantoms using different energy bins.

III. SAMPLE GEOMETRY

The methodology presented above to simulate BrCT images can be adjusted to any scanner geometry. However, as an initial proof of concept, a simple fan-beam BrCT simulation was designed here as shown in Fig. 2.

This particular geometry includes an X-ray source located 200 mm from the isocentre (SID), where the isocentre corresponds to the central axis of the simulated breast phantom.

Two breast phantoms were simulated. The first one corresponds to a cylindrical phantom of radius 70 mm, which approximated a pendate breast filled with adipose tissue. Three spheres (1, 2 and 3 mm in diameter), simulating breast lesions, were included in the central slice of this cylindrical phantom as illustrated in Fig 3.

A second breast phantom used in this work corresponded to an anthropomorphic breast phantom developed at Duke University [8], with an approximate radius of 70 mm (see Fig. 4(a)). This breast phantom included five breast tissues, corresponding to (1) adipose tissue, (2) 25% glandular tissue, (3) 50% glandular tissue, (4) 75% glandular tissue and (5) skin. Furthermore, three realistic breast lesions simulated with a procedure developed at the University of Surrey [9] (see Fig. 4(b)) were inserted into the anthropomorphic breast phantom at different locations, corresponding to different breast tissue regions, as shown in Fig. 4(c).

The composition of the adipose tissue, glandular tissue and skin used in the above phantoms were taken from Hammerstein et al. [10]. In this work, all breast lesions were assumed to be 100% glandular tissue.

A photon counting CdTe slit detector with dimensions 20x400x1 mm³ (width x length x depth) was located at 200 mm from the isocentre (IDD) and a pixel size of 250μm was used.

A beam quality from a W anode and 60 kVp operating voltage, with a total Al filtration of 3 mm, was studied. As the BrCT scanner has a fan-beam geometry, the scattered radiation recorded in the detector was initially assumed to be negligible [7]. The detector quantum efficiency \( \epsilon(E) \) was calculated analytically as described by Beutel et al. [11]:

\[ \epsilon(E) = \frac{\Phi(E)(1 - e^{\mu(E)x})}{\Phi(E)}, \] (3)

where \( \Phi(E) \) represents the photon fluence per energy interval, \( \mu(E) \) the linear attenuation coefficients (taken from...
IV. RESULTS

In this section, results of the imaging spectra analysis are shown for (A) the cylindrical breast phantom and (B) the anthropomorphic breast phantom.

A. Cylindrical phantom

After inserting the cylindrical breast phantom illustrated in Fig. 3 into the BrCT geometry described in Fig. 2, 1,440 projections were calculated. The profiles observed for a given projection at different energy bins are shown in Fig. 5.

The 1,440 primary projection were used to generate the corresponding sinogram as illustrated in Fig. 6.

Then, this sinogram was used to reconstruct the projection images using filtered back projection. The reconstructed image of the central slice of the cylindrical phantom is shown in Fig. 7, where the three spheres are located within the yellow squares.

As explained above, several energy bins were explored in a preliminary attempt to maximise the contrast of the spheres (i.e. breast lesions). Fig. 8 illustrates a profile along the red line in Fig. 7 when using the polychromatic energy spectrum and selected energy window of 21-30 keV.

A region of interest (ROI) of 5x5 pixels was placed inside (ROI_{les}) and outside (ROI_{back}) the 3 mm diameter sphere (bottom) in order to calculate the contrast C at different energy bins ($C = \frac{ROI_{les} - ROI_{back}}{ROI_{les}}$). The contrast values calculated are illustrated in Table I.
TABLE I

<table>
<thead>
<tr>
<th>Energy bin (keV)</th>
<th>Contrast (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-60 (polychromatic)</td>
<td>22.3</td>
</tr>
<tr>
<td>21-30</td>
<td>31.0</td>
</tr>
<tr>
<td>31-40</td>
<td>22.2</td>
</tr>
<tr>
<td>41-50</td>
<td>17.3</td>
</tr>
<tr>
<td>51-60</td>
<td>14.7</td>
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B. Anthropomorphic breast phantom

As explained above for the cylindrical breast phantom, the anthropomorphic breast phantom was also imaged using 1,440 projection. The line integral of the linear attenuation coefficients along the central pixel array of a given projection at a fixed angle is shown in Fig. 9. This is illustrated for X-ray photon energies between 21 and 60 keV in bins of 10 keV each. Furthermore, the locations of the three realistic breast lesions inserted are illustrated with arrows.

After generating 1,440 primary projection of the anthropomorphic breast phantom (every 0.25°), the sinogram shown in Fig. 10 was created. This sinogram was generated for all the X-ray photon energies deposited in the detector (polychromatic energy spectrum) as well as for different energy bins. A sample of the reconstructed image using the anthropomorphic breast phantom for the 21-30 keV bin is shown in Fig. 11.

This has allowed to study the optimal energy bin in the detector that shows maximum contrast of the breast lesions for the simulated geometry. The line integral profiles along the red line illustrated in Fig. 11 are shown for two energy bins in Fig. 12.

The contrast was also measured for each of the energy bins studied using an ROI of 5x5 pixels and it is illustrated in Table II.
TABLE II
Contrast Values Calculated for Most Left Breast Lesion of Fig. 11.

<table>
<thead>
<tr>
<th>Energy bin (keV)</th>
<th>Contrast (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-60 (polychromatic)</td>
<td>9.2</td>
</tr>
<tr>
<td>21-30</td>
<td>13.7</td>
</tr>
<tr>
<td>31-40</td>
<td>9.2</td>
</tr>
<tr>
<td>41-50</td>
<td>6.8</td>
</tr>
<tr>
<td>51-60</td>
<td>5.7</td>
</tr>
</tbody>
</table>

V. DISCUSSION AND CONCLUSIONS

In this work, an image simulation framework was presented to generate images from a dedicated BrCT system. The proposed methodology has allowed us to produce synthetic images in approximately few minutes per projection, representing a fast and cheap method to investigate new BrCT scanners. This represents a key tool which can be used to investigate current geometries as in the development of new scanner geometries.

Furthermore, the energy spectrum recorded within a CdTe photon-counting detector was analysed for a cylindrical breast phantom and a more realistic anthropomorphic breast phantom, where different breast lesions were inserted. In both cases, it was found that an energy bin of 21-30 keV can improve the breast lesion contrast by approximately 9% and 5% for the cylindrical and anthropomorphic breast phantom respectively. This behaviour can be explained via the larger difference between the linear attenuation coefficients of adipose and glandular tissues at lower energies. On the other hand, as the energy bins contain greater X-ray photon energies, the contrast between the background and the lesions becomes smaller. In spite of this increment in contrast, breast cancer detection is very challenging when the lesions are located in glandular regions, as both the lesion and the background have similar attenuation coefficients [16].

Another utility of this proposed image simulation framework can be the investigation of the lesion detection performance when varying the detector technology (e.g. photon counting vs integrating detector).

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REFERENCES