Improvements in Monte Carlo Simulations For Mammographic Scatter Correction

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Introduction

Digital mammography is a specialized breast screening modality which uses X-ray imaging to probe for malignant diseases, most notoriously **breast cancer**. Maximizing contrast in mammographic images is critical to convey diagnostic information.

However, the quality of mammographic imaging is negatively impacted by noise caused by **scattering effects**, such as Compton and Rayleigh scattering. While anti-scattering grids are used, they have the drawbacks of lowering image resolution and exposing patients to higher radiation doses. Primary photons exclusively convey useful diagnostic information pertaining to anatomical attenuation of different soft tissues. Indeed, accurate estimations of the scattered photon spatial distributions thus unlock various algorithmic scatter correction methods.

Here, we harness the power of Monte Carlo simulations to perform grid-less scatter correction up to 200 times more efficiently (K. Sheriff *et al.*) than previous methods. Expanding on this **pixel-by-pixel scatter correction** technique, we present higher fidelity geometry modelling to efficiently extract more accurate primary kerma, suitable for clinical diagnostics.

Monte Carlo Mammography Simulations

Monte Carlo simulations are a mathematical technique used to estimate events governed by probability distributions. This makes them ideal estimators of radiation transport, where large quantities of incident particles can be characterized statistically. For our purposes, the EGSnrc Monte Carlo toolkit was used to model the Hologic Selenia dimensions mammography unit (Fig. 1).

Mammography devices have various properties to consider. For instance, systems use low energy X-rays to increase subject contrast. These are produced through intense electron bombardment of an anode, producing two different types of radiation. Furthermore, the heel effect also affects the uniformity of the produced phase space, as well as the geometry of the incident beam, which must intersect perpendicularly 4 cm away from the chest wall.

Fig. 1

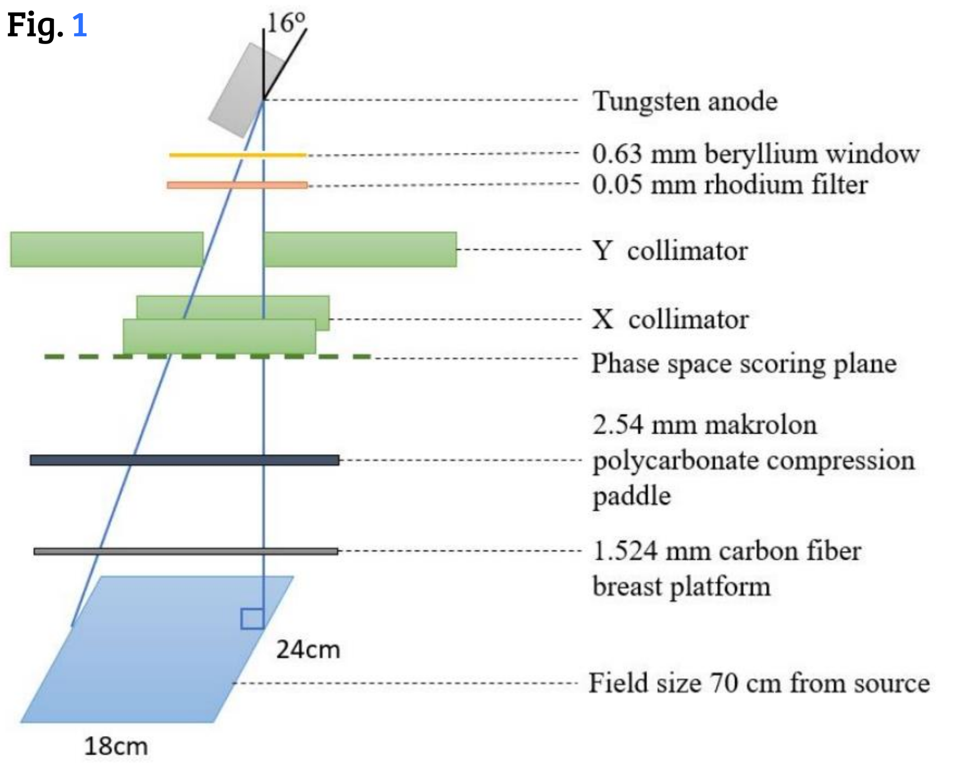
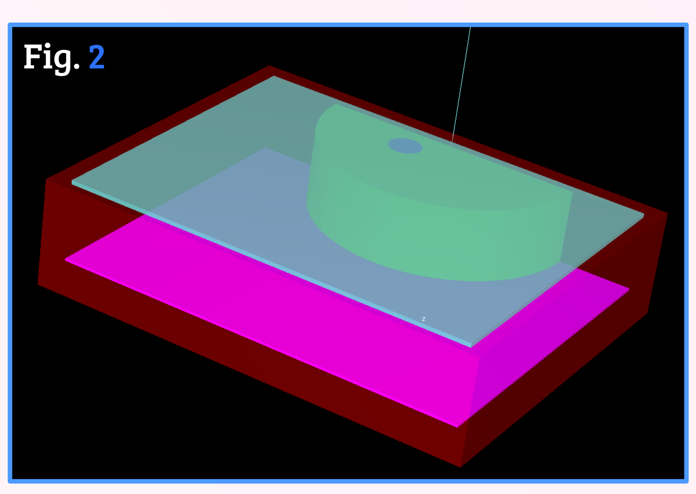


Fig. 2



Region		Material	
0		AIR	✓
1		MAKROLONPOLYCARBONATE	✓
2		ALUMINIUM	✓
3		BREAST_5050	✓
4		CARBON_FIBER	✓

Radiation Interactions

<u>**There are two main types of X-ray sources**</u>

- **Bremsstrahlung Radiation:**
  High energy electrons decelerate inside the anode metal, producing electromagnetic radiation to conserve momentum.
- **Characteristic Radiation:**
  Incoming particles (X-ray or electrons) knock orbital electrons out of the atom. As higher energy electrons fill these vacant spots, x-rays are emitted.

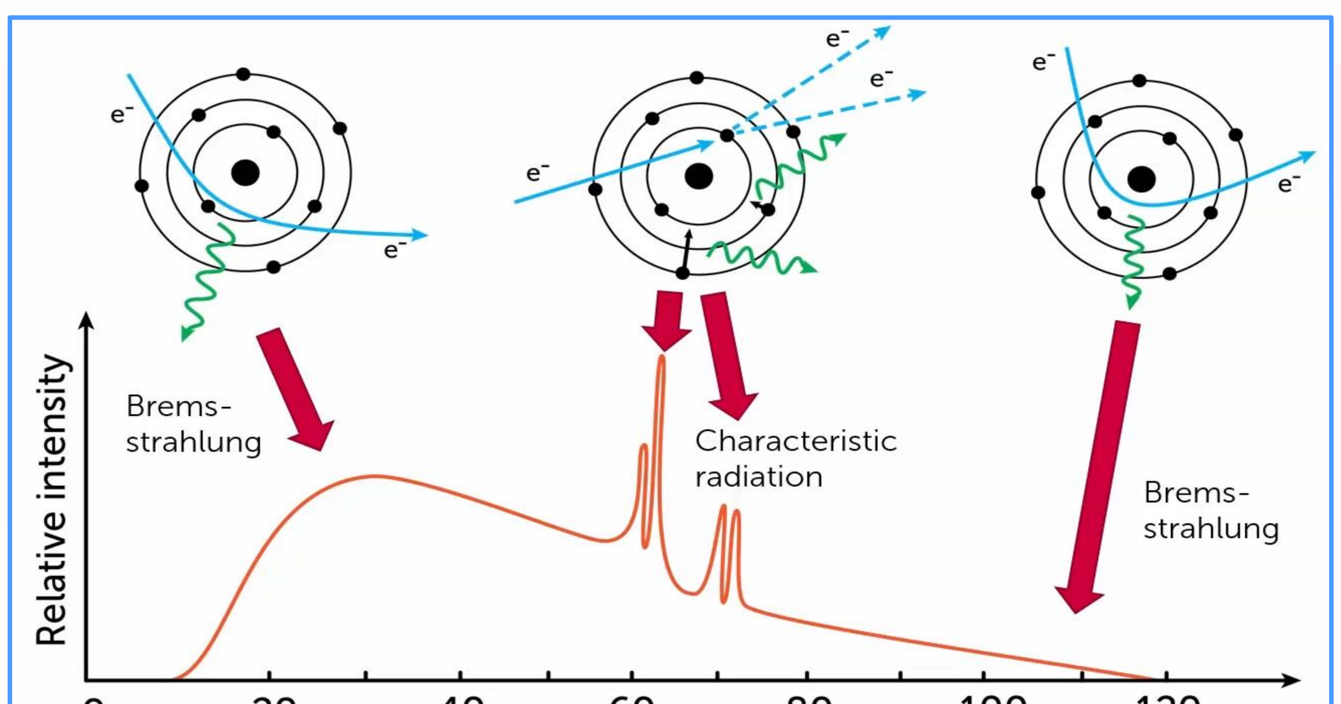
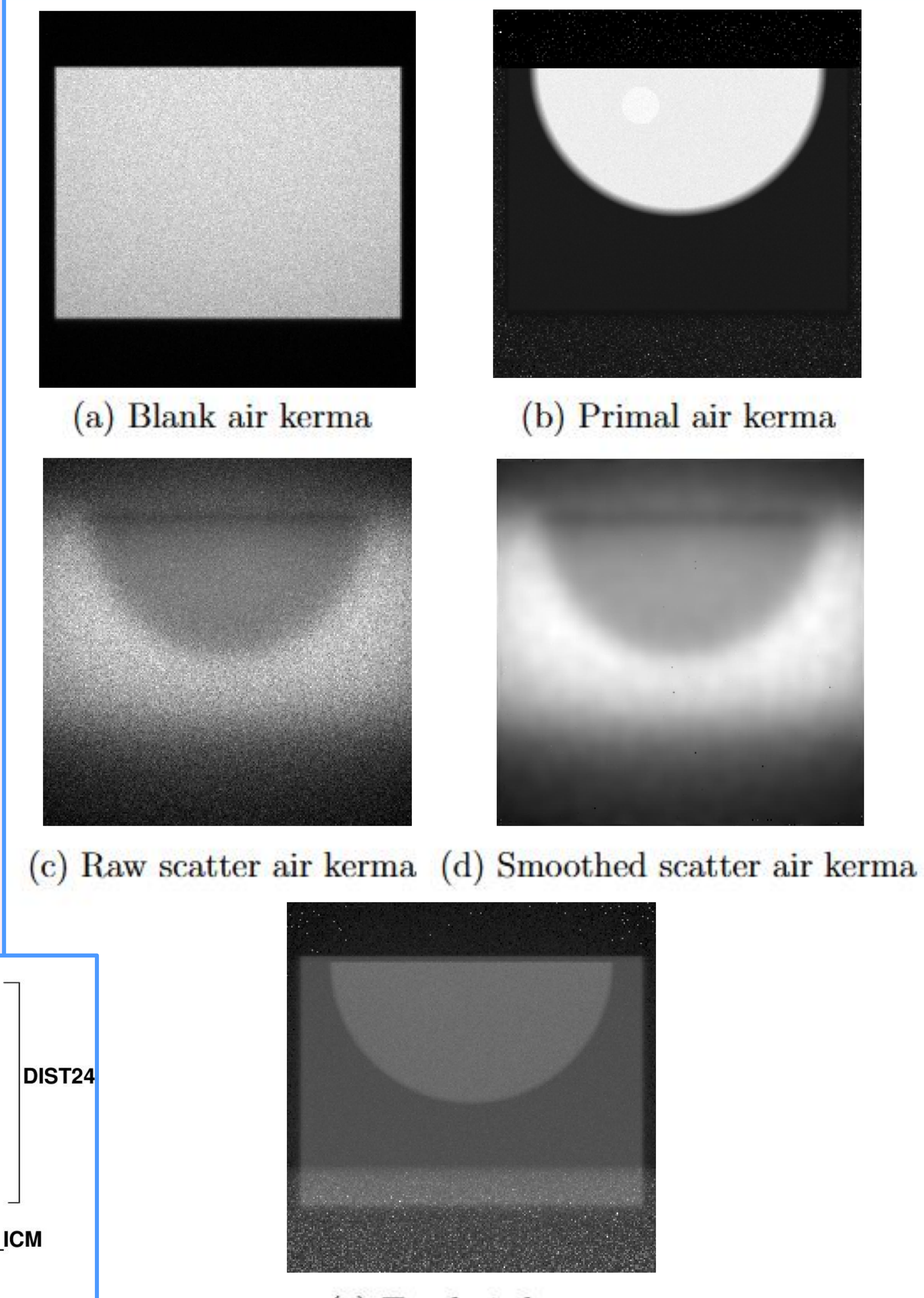


Fig. 3

Results

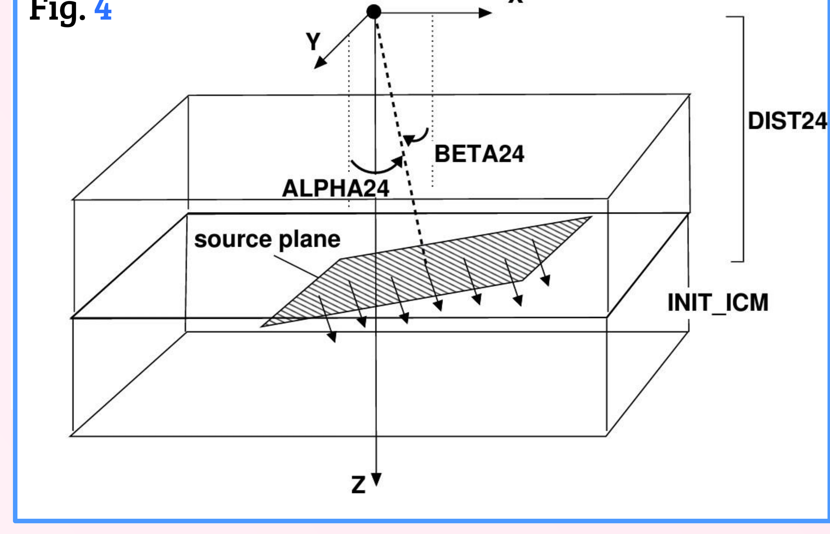
It was recently noticed that the M113T X-ray tube in the Hologic Selenia Dimensions mammography unit was slightly tilted by 6°. To be more accurate in our model, we had to modify the phase space acquisition method (Fig. 4). Indeed, the X-ray tube itself first had to be modelled on its own. Then, a composite input file tilted the phase space according to the specified measurements. On the left, we present the computed air kerma using the new phase space. These distributions are required for scatter correction.



(a) Blank air kerma (b) Primal air kerma (c) Raw scatter air kerma (d) Smoothed scatter air kerma (e) Total air kerma

Fig. 5

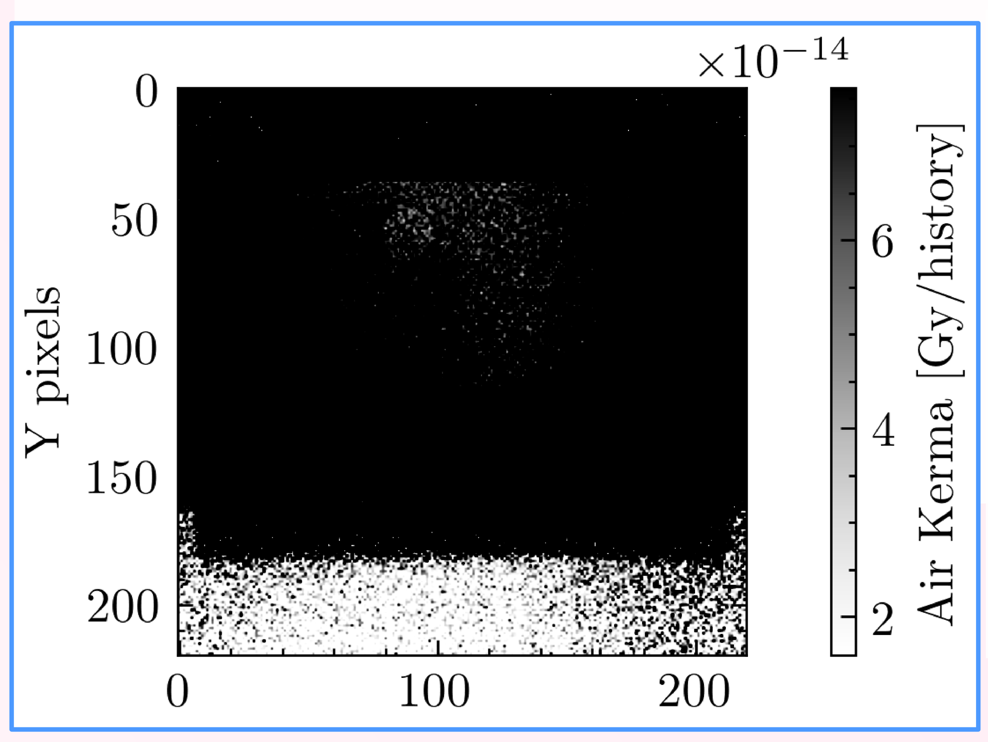
Fig. 4



Scatter Correction

The scatter correction is done on a pixel-by-pixel basis according to the following formula (1):

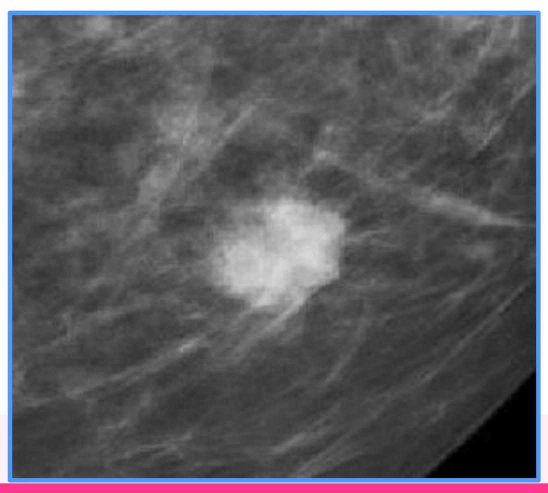
$$P_{real} = \frac{P_{mc}}{T_{mc}} \times T_{real} \quad (1)$$



The intuition behind this procedure is that the ratio between simulated kermas must be similar to the real clinical kerma ratio. This operates in a fashion reminiscent of ray-tracing, which is how such efficiency is acheived. In practice, the acquired mammographic image would replace the T_{real} air kerma when performing digital scatter correction.

Conclusion

Although more accurate simulation geometries should supposedly return higher contrast to noise ratio, the result of the algorithm is not quite as desired. Nonetheless, the primal kerma exhibits a higher concentration of fluence at the tip end (bottom of the image), which was the desired effect of the tilt. Another interesting observation of the tilted source is the larger presence of characteristic Radiation, which indicates larger overall photon fluence in the simulation. The culprit here seems to be the T_{real} file, which displays a large energy band towards the tip side.



Acknowledgements

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[5] Astra Toolbox, "Basic X-ray physics: X-ray generation": (https://www.youtube.com/watch?v=EzedRMAIGI0)