Digital Scatter Removal in Mammography to enable Patient Dose Reduction

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Aims of talk

Scatter in Digital Mammography
Overview of scatter removal algorithm
Advantages over current techniques
Validation using phantoms and clinical mammography equipment
Conclusions
Scatter in Digital Mammography

Scattered photons arise from photon interaction with breast tissue. Scatter reaching digital image detector reduces image contrast. Causes systematic low frequency blurring of primary image. **Grid** is therefore used to absorb and remove scatter.

However....

Grid attenuates primary fluence. Increase mA and breast dose for effective image quality.

*Scatter algorithm can be utilised by Digital Imaging instead of a grid*
Scatter Algorithm

Scatter algorithm is an integral part of an efficient model to calculate tissue densities similar to CT Hounsfield units.

Efficient model:
1. Model of photon energy spectrum (based on Cranley et al)
2. Ray tracer algorithm to obtain energy exiting the breast. Accounts for diverging beam and includes model of breast tissue attenuation.
3. **Scatter algorithm used to subtract scatter component.**
4. Model of image detector incorporating linear transfer functions.
Overall model: Standard Attenuation Rate (SAR)

SAR at an image location $x,y$:

$$\text{SAR}_{x,y} = m_{x,y} \ln(D^{-1}(I_{x,y}) - s_{x,y}) + c_{x,y}$$

$s_{x,y}$ is scatter

$D(i)$ is detector calibration transfer function relating recorded pixel intensity, $i$, to total photon energy absorbed by pixel detector

$m_{x,y}$ and $c_{x,y}$ are dependant on spatial location $x,y$
Scatter Algorithm

Builds on fundamental physical relations underlying Monte Carlo
Uses optimal information sampling and interpolations
Iterative refinement to calculate radiodensity and scatter field
Scatter kernels around primary ray combined to give scatter image.
Considers: Beam quality, position in field, presence of grid

Result:
Efficient scatter removal software which will run on any clinical system.
Scatter Algorithm

\[ \text{Scatter} = \text{Superpose} (\text{Primary}(x), I_{\text{detector}}(x)) \]

Superpose is non-linear. Accounts for breast periphery.

\( I_{\text{detector}}(x) \) is the spatially varying local scatter function which is interpolated using a small set of image locations.

Ability to use this algorithm in forward simulation, i.e. to model the scatter reaching image detector.
Scatter kernels

Division of primary ray into a number of sampled points, \( p \).
Sum the scatter from each point to get scatter kernels.
Sum scatter kernels to get scatter image.
Efficient ray-tracer calculates tissues traversal distances using fundamental scatter relations.
Attenuation \( p \) to \( C \) according to photoelectric absorption.
Validation
Validation of model

Various phantoms designed and built

Imaging carried out on standard digital unit used in NHS hospital:

*Digital Mammography X-ray Unit*

*GE Senographe Essential*

*Oxford Breast Imaging Centre*

Image detector:

aSi, 24 x 31 cms, pixel size 100µm
Scatter kernels

Shape of scatter kernels confirmed by using Pb apertures over scattering medium.

Variation in signal with aperture depends on angular scattering and attenuation.

Results:
Max. 5% error using 70mm aperture
Contrast and Noise

Sharp vertical discontinuity phantom

**Adipose** and **fibro glandular** tissue equivalents

Contrast = \( \frac{\text{mean(bg)} - \text{mean(object)}}{\text{mean(bg)}} \times 100\% \)

\[
\text{CNR} = \frac{\text{mean(bg)} - \text{mean(object)}}{\sqrt{(\text{sd(bg)}^2 + \text{sd(object)}^2)}} \times \frac{1}{2}
\]

(Young et al, BJR, 2006)

29kVp Mo-Rh 71mAs without an anti-scatter grid
Bucky/grid factor effect on CNR

Good CNR image without grid, processed with scatter algorithm

Dose saving of ~37% can be achieved and still maintain CNR.
Dose Reduction

The discontinuity phantom shows that the CNR for the scatter model images remains superior in the absence of a grid.

By using the scatter model a reduction in mAs from 71 to 45 can be achieved before CNR is compromised.
Step wedge test phantom

Tissue equivalent step wedge phantom. Adipose and fibroglandular tissues.
Clinical phantom evaluation

Tissue equivalent phantom – CIRS BR3D
Mimics clinically realistic heterogeneous breast tissue
Contains microcalcifications, fibrils, masses
Clinical Phantom images

Subtract ‘without grid’ images from ‘with grid’ images
Subtraction image shows solely random noise within 95% confidence limits.

Worst case occurs for Rhodium anode where focal spot position is different. Requires 2 pixel translation.
Conclusions

• Scatter algorithm can be computed in a clinically realistic computation time.
  Time to compute ~35 secs with parallelisation techniques.

• Differences between actual and simulated pixel intensities in breast tissue is:
  2.1% with grid
  3.4% without grid.

• Scatter model can restore image quality in absence of grid and at ~40% reduction in dose.

• Improved image quality is possible without having to increase dose.
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