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Effect of shaped filter design on dose and image quality in breast CT

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Abstract

The purpose of this study was to investigate the effect of shaped filters specifically designed for dedicated breast computed tomography (CT) scanners on dose and image quality. Optimization of filter shape and material in fan direction was performed using two different design methods, one aiming at homogeneous noise distributions in the CT images and the other aiming at a uniform dose distribution in the breast. The optimal filter thickness as a function of fan angle was determined iteratively to fulfil the above mentioned criteria for each breast diameter. Different filter materials (aluminium, copper, carbon, polytetrafluoroethylene) and breast phantoms with diameters between 80–180 mm were investigated. Noise uniformity in the reconstructed images, obtained from CT simulations based on ray-tracing methods, and dose in the breast, calculated with a Monte Carlo software tool, were used as figure of merit. Furthermore, CT-value homogeneity, the distribution of noise in cone direction, spatial resolution from centre to periphery and the contrast-to-noise ratio weighted by dose (CNRD) were evaluated. In addition, the decrease of scatter due to shaped filters was investigated. Since only few or one filter are practical in clinical CT systems, the effects of one shaped filter for different breast diameters were also investigated. In this case the filter, designed for the largest breast diameter, was simulated at variable source-to-filter distances depending on breast diameter. With the filter design method aiming at uniform noise distribution best results were obtained for aluminium as the filter material. Noise uniformity improved from 20% down to 5% and dose was reduced by about 30–40% for all breast diameters. No decrease of noise uniformity in cone direction, CT-value homogeneity, spatial resolution and the CNRD was detected with the shaped filter. However, a small improvement of CNRD was observed. Furthermore, a scatter reduction of about 20–30% and a more homogeneous scatter distribution were reached which led to reduced cupping artefacts. The simulations with one shaped filter at variable source-to-filter distance resulted in nearly homogeneous noise distributions and comparable dose reduction for
all breast diameters. In conclusion, by means of shaped filters designed for breast CT, significant dose reduction can be achieved at unimpaired image quality. One shaped filter designed for the largest breast diameter used with variable source-to-filter distance appears to be the best solution for breast CT.

(Some figures may appear in colour only in the online journal)

1. Introduction

In recent years, computed tomography (CT) technology has greatly advanced concerning spatial resolution and low contrast detectability. This opened the field for new applications in the area of breast cancer diagnosis leading to dedicated CT of the female breast (bCT) efforts (Boone et al. 2006, Ning et al. 2007, Kalender et al. 2011, Kalender 2011).

However, bCT systems, like any other CT system, suffer from a fan angle-dependent intensity distribution at the detector caused by the varying path lengths through the object. Such intensity differences from the centre to the periphery result in a non-uniform noise distribution in the reconstructed images. Furthermore, with the same beam intensity applied to the periphery of the object as to the centre an unnecessary high amount of dose is delivered. A solution for this is the use of shaped filters with an angle-dependent thickness to reduce the peripheral x-ray beam intensity. Shaped filters were already investigated for clinical CT systems concerning their effect on dose, image quality (Mail et al. 2009) and scatter (Bootsma et al. 2011). The reverse method to evaluate the attenuation profile of an already installed filter was also analysed (Boone 2010). For bCT systems as well, filter shape and thickness were investigated to achieve improved results regarding dose reduction (Boone et al. 2004) and noise uniformity (Cai et al. 2007).

The approach of respective prior work was to equalize the photon fluence at the detector. But due to the filtered backprojection (FBP) algorithm used for the reconstruction process, even a uniform fluence at the detector does not result in a homogeneous noise distribution in the reconstructed images. In particular, for small source-to-isocentre distances (SIDs) and large fan angles, the high-frequency noise components in the object periphery are amplified compared to the centre (Wunderlich and Noo 2007, Zeng 2004, Bennett and Byer 1986).

The aim of our work was to optimize shaped filters for bCT in fan direction based on a simulation study to provide dose reduction and optimization of CT image quality for different breast diameters. The study was executed as follows: noise uniformity and dose reduction were evaluated in simulations for different breast diameters. Also the influence of the shaped filter on the noise distribution in cone direction, CT-value homogeneity, spatial resolution and the contrast-to-noise ratio weighted by dose (CNRD) were investigated. Two different calculation methods for the filter design were compared, one aiming at uniform energy absorption inside the breast and another one aiming at homogeneous noise distribution in the CT images. For improvement of the second method a FBP correction factor was derived to equalize the spatially variant noise in the reconstructed images. Different filter materials were evaluated regarding beam hardening and filter thickness. Furthermore, scatter simulations were performed to investigate the scatter distribution at the detector and the scatter induced artefacts in the CT images. The potential of using only one shaped filter for all breast diameters and changing the source-to-filter distance (SFD) was investigated also.
2. Materials and methods

2.1. Simulation setup and evaluation

2.1.1. Breast CT setup. For the calculation of the shaped filter an example bCT scanner setup was assumed (figure 1, Kalender et al. 2011) with a SID of 300 mm and an isocentre-to-detector distance (IDD) of 150 mm. A breast phantom with radius $R_{\text{breast}}$ was located at the centre of rotation and the shaped filter at a SFD of 60 mm. The x-ray beam intensity $I_{\text{detector}}(\theta)$ is attenuated by the shaped filter thickness $d_{\text{filter}}(\theta)$ and the path length through the breast $d_{\text{breast}}(\theta)$ according to the fan angle $\theta$. $I_{\text{detector}}(\theta)$ was calculated by equation (1).

$$I_{\text{detector}}(\theta) = I_0 \cdot e^{-\mu_d(\theta) d_{\text{filter}}(\theta)} - e^{-\mu_d(\theta) d_{\text{breast}}(\theta)}$$  \hspace{1cm} (1)$$

The intersection length of x-rays through the breast was calculated depending on the radius and the SID (2).

$$d_{\text{breast}}(\theta) = 2 \cdot \sqrt{R_{\text{breast}}^2 - \text{SID}^2 \cdot \sin^2(\theta)}$$  \hspace{1cm} (2)$$

The spectrum of the x-ray source was calculated according to Tucker et al. (1991) with a tube voltage of 60 kV (Weigel et al. 2011), a filtration of 3.0 mm aluminium and an anode angle of 7°. The collimation in fan direction was adapted to cover the largest breast phantom (180 mm in diameter) with a maximum fan angle of 35°. In cone direction the collimation was set to an angle of 9.8°. For image acquisition a curved photon counting detector was simulated using cadmium telluride (CdTe) as sensor material with a thickness of 1.0 mm, a pixel matrix of 2816 $\times$ 768 and a pixel pitch of 100 $\mu$m. 2000 projections were taken over a 360° circular scan with a duration of 2 s. The shaped filter was taken into account by attenuating the x-ray beam intensity according to its angle-dependent thickness.

For simulations of dose and noise in the reconstructed images a homogeneous breast phantom was used (figure 2(a)). The volumes of interest (VOIs) for the noise evaluation (section 2.1.3) are highlighted. The phantom consisted of a cylinder made of breast-equivalent material (80% adipose and 20% glandular tissue, Yaffe et al. 2009) with a height of 100 mm and with variable diameters between 80 and 180 mm in steps of 10 mm.
Figure 2. Image of a homogeneous breast phantom with a diameter of 140 mm (a), a wire phantom of equal diameter and material with a peripheral tungsten wire (b) and a contrast insert phantom similar to the breast phantom with five iodine inserts (c). The VOIs important for evaluation are highlighted.

A wire phantom was simulated to determine spatial resolution with the use of the modulation transfer function (figure 2(b)). This phantom is equivalent to the breast phantom with an additional tungsten wire of 10 μm in diameter placed vertically in the centre or in the periphery at 80% of the phantom’s radius.

For the evaluation of the CNRD (Kalender et al. 2009, section 2.1.3) a contrast insert phantom with five iodine inserts at 10, 30, 50, 70 and 90% of the phantom’s radius was simulated (figure 2(c)). These inserts have a radius of 5% of $R_{\text{breast}}$ and are a mixture of water and 20 mg ml$^{-1}$ iodine (Weigel et al. 2011). The material and dimensions of the phantom are similar to the breast phantom. The VOIs important for the CNRD evaluation are highlighted.
2.1.2. Generation of CT raw data, reconstructed images and dose distributions. The simulation program ImpactSim (CT Imaging GmbH, Erlangen, Germany) was used to generate CT raw data. For each detector pixel the corresponding x-ray beam attenuation from source to detector was calculated taking the filter and breast phantom into account. The intensities at the detector were adapted to an air kerma at isocentre corresponding to an average glandular dose (AGD) of 3 mGy for the 140 mm breast diameter simulated without shaped filter (Kalender et al 2011). Poisson-distributed noise was added to the projections according to Press et al (2007).

Images were reconstructed with ImpactRecon (CT Imaging GmbH, Erlangen, Germany) with a standard Feldkamp algorithm (Feldkamp et al 1984) using Shepp–Logan convolution kernel (Shepp and Logan 1974) and voxel sizes of (75 μm)³. Furthermore an analytical precorrection was done to correct for beam hardening by the filter and the breast phantom.

3D dose distributions within the object were simulated with the validated Monte Carlo program ImpactMC (CT Imaging GmbH, Erlangen, Germany) (Deak et al 2008, Chen et al 2012). The deposited dose was calculated in mGy for 1 mm³ voxels.

2.1.3. Evaluation. The dose reduction for the breast with shaped filter compared to the simulation without was calculated in percent using the AGD according to equation (3).

\[
\text{Dose reduction} = \left(1 - \frac{\text{AGD}_{\text{with}}}{\text{AGD}_{\text{without}}}\right) \cdot 100\% \quad (3)
\]

For the noise distribution in fan direction of the reconstructed images the uniformity index (UI) (Bissonnette et al 2008) was calculated as the figure of merit (4). It represents the averaged percentage difference between the noise \( \sigma \) within four peripheral and one central VOI (figure 2(a)). The peripheral VOIs were positioned at 82.5% of \( R_{\text{breast}} \), with a radius of 7.5% of \( R_{\text{breast}} \) and a height equivalent to the reconstructed volume.

\[
\text{UI} = \frac{1}{4} \sum_{n=1}^{4} \left( \frac{\sigma_{\text{periphery}} - \sigma_{\text{centre}}}{\sigma_{\text{centre}}} \right) \cdot \frac{1}{4} \cdot 100\% \quad (4)
\]

To evaluate the noise distribution in cone direction one peripheral and one central VOI was used with the same dimensions as described above. A noise profile in z-direction was calculated inside these VOIs within the irradiated volume according to the opening angle in cone direction. This was investigated for a breast diameter of 140 mm with shaped filter.

Beside noise, also the homogeneity of the CT-values in the reconstructed images was investigated. The wire phantom was used to evaluate the influence of shaped filters on the spatial resolution in the centre and in the periphery for all breast diameters.

Furthermore, to determine the distribution of the CNRD from centre to periphery, the contrast insert phantom (figure 2(c)) was used. Five VOIs were positioned within the iodine inserts and further VOIs, used as reference, at the same distance from the centre rotated by 90°. The VOI diameter was set to 10% of the phantom’s radius. The contrast-to-noise ratio (CNR) was calculated between the iodine inserts \( i \) and the background defined by the reference VOIs \( j \) located at the same radius, respectively. The AGD within these reference VOIs was used to calculate the corresponding CNRD according to equation (5). This was done for a phantom diameter of 140 mm.

\[
C_{i-j,NRD} = \frac{C_{i-j,NR}}{\sqrt{\text{AGD}_j}} \quad (5)
\]
2.2. Filter design

The thickness of the shaped filter has to be varied as a function of fan angle $\theta$ and path length through the breast $d_{\text{breast}}(\theta)$ to adapt the x-ray beam attenuation. Two different design approaches for the angle-dependent filter thickness were investigated with respect to dose reduction and noise uniformity.

2.2.1. Design approach. Without a shaped filter the breast is irradiated with the same beam intensity independent of fan angle. According to the Beer–Lambert law the intensity is attenuated exponentially in relation to depth of penetration, which leads to a higher average dose in the periphery compared to the centre of the scanned object. Therefore, to reduce dose, one approach for the shaped filter design is to achieve a homogeneous energy absorption in the breast (HomAbsorption). This was realized by comparing the intensity $I_{\text{filter}}(\theta)$ in front of the breast and $I_{\text{detector}}(\theta)$ behind the breast for each fan angle. The loss of intensity relative to the intersection length $d_{\text{breast}}(\theta)$ has to be approximately the same compared to the energy absorption of the central ray (6) to compensate the effects of the inhomogeneous attenuation. This was obtained by iteratively adapting the angle-dependent filter thickness $d_{\text{filter}}(\theta)$ with the following constraint:

$$
\frac{I_{\text{filter}}(\theta) - I_{\text{detector}}(\theta)}{d_{\text{breast}}(\theta)} = \frac{I_{\text{filter}}(0^\circ) - I_{\text{detector}}(0^\circ)}{d_{\text{breast}}(0^\circ)}
$$

Another option for the shaped filter design is to reduce the noise inhomogeneity in the CT images that is caused by the high intensity differences at the detector. The simulation of a breast phantom without shaped filtration leads to a high peripheral photon fluence at the detector which results in lower noise compared to the centre. This lack of noise uniformity can also be seen in the reconstructed CT images. Therefore, the assumption for obtaining a homogeneous noise distribution in the CT images (HomNoise) is to equalize the angle-dependent noise at the detector (7a). For an ideal photon-counting detector the noise depends only on the Poisson-distributed detected number of photons $N_{\text{detector}}(\theta)$ (7b). With the variation of the filter thickness $d_{\text{filter}}(\theta)$ a homogeneous detector noise can be reached.

$$
N_{\text{detector}}(\theta) = N_{\text{detector}}(0^\circ)
$$

$$
N_{\text{detector}}(\theta) = N_0 \cdot e^{-(\mu d)(\theta)_{\text{breast}} - (\mu d)(\theta)_{\text{filter}}}
$$

However, due to the cone beam reconstruction with a standard FBP algorithm for a 3D volume (Herman 1995, Lewitt and McKay 1980), no noise homogeneity is reached in the reconstructed images (Wunderlich and Noo 2007, Zeng 2004, Bennett and Byer 1986). Instead a fixed cut-off frequency of the convolution kernel for all voxels leads to more high-frequency noise components in the periphery compared to the centre and consequently to a higher noise level. Therefore a correction factor $W_{\text{FBP}}(\theta)$ was derived to homogenize the noise distribution in the reconstructed images and to update the design method HomNoise (equations (8a) and (8b)). Details on this correction factor are given in the appendix.

$$
F_{\text{FBP}}(\theta) = \frac{1}{N_{x,y,z}} \cdot \sum_{i=0}^{N_x} \sum_{j=0}^{N_y} \sum_{k=0}^{N_z} U(x_i, y_j, \alpha_k) \frac{1}{\theta = \text{const}}
$$

$$
W_{\text{FBP}}(\theta) = \frac{F_{\text{FBP}}(\theta)}{F_{\text{FBP}}(0^\circ)}
$$
2.2.2. Filter material. Not only the filter shape but also the filter material influences the results regarding dose reduction and noise uniformity. Depending on the attenuation properties, beam hardening occurs to a greater or lesser extent which influences CT-values. Without taking into account the beam hardening produced by the shaped filter, capping artefacts and a loss of spatial resolution result in the reconstructed images (Benítez and Ning 2010). To limit the decrease of image quality the material for a shaped filter should be of high density to keep the thickness low and of low atomic number to limit beam hardening. Aluminium, carbon, copper and polytetrafluoroethylene (PTFE) were tested as filter materials (table 1, Bootsma et al 2011, Graham et al 2007, Boone et al 2004).

<table>
<thead>
<tr>
<th>Material</th>
<th>Effective atomic number $Z_{\text{eff}}$</th>
<th>Density $\rho$ (g cm$^{-3}$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Aluminium (Al)</td>
<td>13</td>
<td>2.7</td>
</tr>
<tr>
<td>Carbon (C)</td>
<td>6</td>
<td>2.3</td>
</tr>
<tr>
<td>Copper (Cu)</td>
<td>29</td>
<td>8.9</td>
</tr>
<tr>
<td>PTFE (C$_2$F$_4$)</td>
<td>8.4</td>
<td>2.2</td>
</tr>
</tbody>
</table>

2.2.3. Thickness calculation. For the evaluation of the angle-dependent filter thickness, $I_{\text{detector}}(\theta)$ and $N_{\text{detector}}(\theta)$ were determined for each fan angle (9), (10) and compared with the values for the central ray (6), (7a). The thickness of the shaped filter $d_{\text{filter}}(\theta)$ was increased iteratively to account for the lower attenuation in the breast periphery (11) until the constraint of the calculation method was achieved. This was done for each angle with an increment of 0.1° until the maximum fan angle was reached and repeated for each breast diameter separately. The calculation method HomNoise additionally takes the FBP correction factor $W_{\text{FBP}}(\theta)$ into account (10).

$$I_{\text{detector}}(\theta) = \sum_{i=1}^{E_{\text{max}}} N_i \cdot E_i \cdot e^{-\mu(E_i) \cdot d_{\text{filter}}(\theta)}$$ (9)

$$N_{\text{detector}}(\theta) = \sum_{i=1}^{E_{\text{max}}} N_i \cdot e^{-\mu(E_i) \cdot d_{\text{filter}}(\theta) \cdot W_{\text{FBP}}(\theta)}$$ (10)

$$\mu(E) d(\theta) = (\mu(E) d(\theta))_{\text{filter}} + (\mu(E) d(\theta))_{\text{breast}}$$ (11)

For the final filter design the calculated angle-dependent thickness was converted from cylinder to Cartesian coordinates to describe the filter parameters (12a), (12b). To simplify the calculations, it was assumed that the shaped filter has no variation in $z$-direction since $R_{\text{breast}}$ does not depend on $z$.

$$x = SFD \cdot \tan(\theta) + d_{\text{filter}}(\theta) \cdot \sin(\theta)$$ (12a)

$$y = d_{\text{filter}}(\theta) \cdot \cos(\theta)$$ (12b)

2.3. Scatter simulations

3D scatter simulations were performed using ImpactSim as shown in Kyriakou et al (2006) with the shaped filter treated as phantom. Due to the homogeneity of the phantom in $z$-direction and the resulting weak dependence of the scatter on $z$-position only the collected
scatter profile in the central detector row was evaluated; however, the scattered radiation from the whole phantom was regarded. The influence of the shaped filter and the breast phantom on scatter was investigated separately and in total. Simulations were done for the breast diameters 80, 140 and 180 mm with the corresponding shaped filter.

Also the CNR and CNRD was evaluated with the contrast insert phantom for a diameter of 140 mm to simulate the effects of scatter on contrast.

2.4. Variation of the source-to-filter distance

In the previous simulations each breast phantom was used with a shaped filter specifically designed for the breast diameter at a fixed SFD of 60 mm. For the assumed bCT system as for the clinical application of CT systems it is impossible to use the correct shaped filter for each object due to spacing reasons. An intention to overcome this disadvantage is to use only one shaped filter with a variable SFD depending on breast diameter.

Simulations were performed using the shaped filter designed for the largest breast diameter (180 mm) for all breast diameters. Each phantom was simulated at different SFDs between 50 and 100 mm in steps of 10 mm. The position and the modified attenuation profile of the filter were considered in the reconstruction. For small SFDs and large phantom diameters the fan may extend beyond the filter; these data were neglected.

The data obtained was used to identify the optimal SFD for each breast diameter aiming for highest dose reduction and noise uniformity.

3. Results

3.1. Filter design

3.1.1. Design approaches. In order to compare the two methods, aluminium was used as filter material for this part of the study. A detailed investigation of the optimal shaped filter material is presented in section 3.1.2.

Figures 3(a) and (b) show the noise and dose profiles for a 140 mm breast phantom that was simulated with the shaped filter designed according to the HomNoise and HomAbsorption method, respectively. For comparison, the results simulated without a shaped filter are also displayed. Figure 3(c) illustrates the dose reduction in percent for all breast diameters and for each design method in contrast to the results without a shaped filter. The noise UI was calculated in the same manner (figure 3(d)).

The HomNoise method proved to be the superior method concerning dose reduction and noise uniformity for all breast diameters and was therefore chosen to be the filter design method for the following simulations.

3.1.2. Filter material. The simulation results for each material and breast diameter, respectively, are given in figure 4. Materials with higher atomic number led to higher dose reduction, but also to an increased noise inhomogeneity. Materials with low atomic number generally resulted in better noise uniformity.

The effects of the different materials on the shaped filter size as well as beam hardening were also investigated. Figure 5(a)) shows the different filter dimensions for the 140 mm breast diameter and the design method HomNoise. Due to the different attenuation characteristics the filter shape varied strongly in thickness and width even for the same breast diameter and the same design method. Figure 5(b) displays the normalized spectra with the attenuation of the corresponding shaped filter at the maximum fan angle. The attenuation of the breast phantom was not included in these calculations. Table 2 summarizes the results simulated
Figure 3. Dose (a) and noise (b) profiles for the 140 mm breast diameter and the resulting dose reduction (c) and noise UI (d) for all breast diameters.

The results for the dose reduction and noise UI showed only small differences for all materials. Carbon for example led to a good noise UI but also to a lower dose reduction with the 140 mm breast diameter for all materials used in comparison to the results without a shaped filter.
Figure 5. Filter dimensions (a) for different materials designed for the 140 mm breast diameter and the resulting normalized spectra (b) for the maximum attenuation compared to the spectrum without shaped filter.

Table 2. Dose reduction and noise UI for the different filters designed for the breast diameter of 140 mm and the mean energy of the resulting spectra for the maximum attenuation compared to the results without a shaped filter.

<table>
<thead>
<tr>
<th>Material</th>
<th>Dose reduction</th>
<th>Noise UI</th>
<th>Mean energy (keV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Al</td>
<td>36.7%</td>
<td>3.9%</td>
<td>46.2</td>
</tr>
<tr>
<td>C</td>
<td>32.4%</td>
<td>1.0%</td>
<td>40.6</td>
</tr>
<tr>
<td>Cu</td>
<td>37.8%</td>
<td>5.2%</td>
<td>48.0</td>
</tr>
<tr>
<td>PTFE</td>
<td>34.4%</td>
<td>3.0%</td>
<td>43.3</td>
</tr>
<tr>
<td>w/o shaped filter</td>
<td>0.0%</td>
<td>−18.7%</td>
<td>37.4</td>
</tr>
</tbody>
</table>

compared to the other materials. However, due to spacing reasons, the filter thickness and width is a determining factor in the choice of the optimal filter material. The filter designed with carbon or PTFE with a maximum thickness of about 70 mm and 40 mm, respectively, would exceed the space for the given bCT setup. On the contrary, the filter made of copper with a maximum thickness less than 1 mm would be hard to produce. This leads to aluminium as the optimal choice among the considered materials for the shaped filter design. The beam hardening caused by this filter leads to an increase up to 10 keV in the mean energy of the spectrum as compared to the source spectrum (table 2). This effect can be compensated for by the analytical precorrection during reconstruction. For the following simulations aluminium was used as filter material.

3.1.3. Noise profiles in cone direction. Noise profiles in cone direction are shown in figure 6 scaled to the noise in the centre of the central slice. Noise was calculated within the central and peripheral VOIs over the whole volume. Due to the phantom diameter (140 mm) and the given cone angle a maximal volume of about 35 mm in z-direction was reconstructed. The use of a shaped filter showed a uniform noise distribution in z-direction.

3.1.4. Spatial resolution, CT-value homogeneity and CNRD. With the use of the shaped filter the spatial resolution for the central wire showed no deviation compared to the simulations without a filter and even for the peripheral wire only a degradation of about 3% was observed.
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Figure 6. Noise profiles in z-direction of the centre and the periphery for a phantom diameter of 140 mm.

Figure 7. CNR and CNRD simulated with a contrast insert phantom of 140 mm diameter.

No loss of CT-value homogeneity was detected for all breast diameters when an appropriate correction algorithm was applied.

The results for the CNR and CNRD simulations are shown in figure 7. Without shaped filter the CNR increased from the centre to the periphery due to decreasing noise. The simulations with shaped filter showed an almost constant CNR due to the more uniform noise distribution. As a result of the inhomogeneous dose distribution for the case without shaped filter the CNR weighted by dose was constant for both cases. With shaped filter the CNRD was higher across the whole phantom compared to the simulations without filter.

3.2. Scatter simulations

For the evaluation of the different scatter sources shaped filter, breast and the combination of both, the scatter intensity in the central detector row was determined as scattered photons per pixel for the simulation of the 140 mm breast diameter (figure 8(a)). The scatter to primary ratio (SPR) was calculated to illustrate the total fraction of scatter (figure 8(b)). Also the effect of different breast diameters was investigated on the total scatter (figure 8(c)) and the SPR (figure 8(d)).
Figure 8. Detected scatter in the central detector row for the standard scan protocol (a, c) and SPR in percent (b, d) simulated for the different scatter objects with the 140 mm breast diameter (a, b) and for different breast diameters including the shaped filter (c, d).

Figure 9. Total scatter reduction (a) for all breast diameters in relation to the simulation without a shaped filter and CT profiles (b) for the 140 mm breast phantom simulated with and without shaped filter.

A total scatter reduction of about 30% was reached for all breast diameters (figure 9(a)). Furthermore a more homogeneous scatter distribution was detected (figure 8(b)) which led to less cupping in the reconstructed images. This can be observed in figure 9(b) for the 140 mm breast diameter.
The results for the CNR and CNRD simulations with scatter are shown in figure 10. Towards the periphery the CNR without shaped filter increased equally to the results without scatter but with a greater magnitude due to higher CT-values as a result of scatter. With shaped filter CT-value homogeneity was improved which led to an almost constant CNR from the centre to the periphery. The results of the CNRD showed an almost uniform distribution for the case with shaped filter and a slightly increased CNRD in the periphery for the case without. However, the CNRD with shaped filter was improved across the whole phantom in contrast to the simulations without filter.

3.3. Variation of the source-to-filter distance

Figure 11 shows the dose reduction and noise UI for the shaped filter optimized for the largest breast diameter (180 mm) applied to different breast diameters by varying SFD.

Starting from the noise UI results of figure 11(b), more detailed simulations were done with small SFD increments aiming for optimal noise UI results. These results are shown in figure 12 and the corresponding SFDs are listed in table 3. For comparison the results of the
Figure 12. Dose reduction (a) and noise UI (b) for different breast diameters using the shaped filter optimized for the 180 mm breast diameter at a fixed SFD of 60 mm for which the filter was designed for and at the optimal SFDs between 54 and 112 mm.

Table 3. The optimal SFD for each simulated breast diameter (BD).

<table>
<thead>
<tr>
<th>BD (mm)</th>
<th>80</th>
<th>90</th>
<th>100</th>
<th>110</th>
<th>120</th>
<th>130</th>
<th>140</th>
<th>150</th>
<th>160</th>
<th>170</th>
<th>180</th>
</tr>
</thead>
<tbody>
<tr>
<td>SFD (mm)</td>
<td>112</td>
<td>104</td>
<td>98</td>
<td>92</td>
<td>82</td>
<td>78</td>
<td>72</td>
<td>68</td>
<td>64</td>
<td>60</td>
<td>54</td>
</tr>
</tbody>
</table>

same filter at a fixed SFD of 60 mm for which the filter was originally designed for are also shown.

With the variation of the SFD between 54 and 112 mm and the use of the shaped filter designed for a 180 mm breast diameter a dose reduction of about 30% and a nearly homogeneous noise uniformity was reached.

A visual illustration of the effects of a shaped filter on dose reduction, noise UI and CT-value homogeneity are displayed pictorially in figure 13. It shows the simulation results for the 140 mm breast diameter with and without shaped filter.

4. Discussion

The proposed filter design method HomNoise showed superior results in dose reduction and image quality compared to the method HomAbsorption. The HomNoise method is a combination of the widely used approach to equalize the intensity distribution at the detector and the intention to retain noise uniformity in the reconstructed images with the use of an FBP correction factor. However, even with the use of this correction factor no perfect noise uniformity was obtained in the reconstructed images. Nevertheless a noise UI of 3.9% with the HomNoise method for the 140 mm breast diameter was an improvement to the results without a shaped filter with a UI of −19%. Furthermore, a dose reduction of 37% was achieved with this method.

The results of the design method HomNoise were different to the design method HomAbsorption due to the aim of the constraints used. For the HomAbsorption method the total beam intensity was used which differs for polychromatic spectra compared to the HomNoise method for which the photon fluence at the detector was used. Furthermore, the loss of intensity in relation to the intersection length was calculated for HomAbsorption. This led to an inhomogeneous energy distribution at the detector that reflected the shape and thickness of
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Figure 13. Simulation of dose distribution (left), image noise (centre) and scatter induced cupping (right) for the 140 mm breast phantom without (top) and with shaped filter (bottom).

Concerning the choice of the filter material, carbon showed best results for noise uniformity and beam hardening (figures 4(b) and 5(b)). Although the dose reduction was slightly worse for this material compared to the others, it may be considered the best choice as a filter material. With a maximum thickness of 70 mm it is, however, too big for the bCT setup. Aluminium with a maximum thickness of about 20 mm presents a good choice as filter material. In general, there were only small differences in noise uniformity and dose reduction between the different materials, and the effect of the beam hardening can be compensated by considering the shaped filter in the reconstruction.

The use of a shaped filter led not only to advantages in dose reduction and noise uniformity, but also to a reduction of scattered radiation (figure 9(a)) and a more uniform scatter distribution (figure 8(b)) for all investigated breast diameters. A more homogeneous scatter reduced or even cancelled the cupping in the reconstructed images (figure 9(b)) which can be positive for image quality.

The magnitude of the calculated SPR for the 140 mm breast phantom corresponded well to similar simulations and measurements of a water cylinder with a diameter of 164 mm and an IDD of 300 mm which resulted in an SPR of 75% (Bootsma et al 2011). Due to the different parameters a discrepancy in the order of 15% compared to our simulations for the 140 mm breast phantom without a shaped filter appears reasonable.

A further improvement with respect to beam hardening and scatter reduction could be a combination of different materials for the shaped filter design (Benitez and Ning 2010). However, to evaluate the optimal combination and percentages of filter materials means extensive work and would be beyond the scope of this paper.

The variation of SFD instead of filter varying shapes led to good results in dose reduction and noise UI for each breast diameter (figure 12) compared to the simulations with filters
designed according to each breast diameter (figure 3). The results with the filter optimized for the 180 mm breast diameter and positioned at optimal SFDs between 54 and 112 mm for different diameters (figure 12(b)) resulted in nearly homogeneous noise distributions.

The investigated shaped filter design as well as the simulated breast phantom were only varied in fan direction. This may be a disadvantage in case of a change in shape of the breast from the chest to the nipple. One method to compensate this is the proposed spiral scan for the final bCT scanner (Kalender et al 2011). During acquisition the breast CT setup moves along a spiral track to cover the full breast in z-direction. According to the z-position and breast diameter the filter position can dynamically be adapted.

Nevertheless, with the results of the optimization in fan direction a good shaped filter design for bCT was found. Furthermore, an experimental validation of the final shaped filter at a variable SFD is necessary which will be part of further studies.

5. Conclusion

The proposed filter design method HomNoise provides significant improvements of noise homogeneity and dose reduction. This is achieved without degradation of spatial resolution, CT-value homogeneity or CNRD. Furthermore, the use of a shaped filter results in scatter reduction and more uniform scatter distributions for all breast diameters leading to a reduction of artefacts in CT images.

If the CT system does not allow for a variety of shaped filters to be used, a variable SFD and only one shaped filter designed for the largest breast diameter should be considered. It results in additional dose reduction and noise uniformity for all breast diameters.

In conclusion, the use of a shaped filter in bCT appears essential to keep patient dose as low as reasonably achievable and to improve image quality by reducing noise inhomogeneity as well as scatter-induced artefacts.

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Appendix. FBP correction factor

For the following calculations a simple setup was assumed with the bCT system rotating around the Cartesian coordinate system of the breast phantom with the rotation angle $\alpha$. Figure A1 displays the x-ray source with a ray interacting within a voxel $v(x, y)$ hitting the curved detector at the position $p'(x, y, \alpha)$.

The FBP algorithm used in the reconstruction is given in equation (A.1a) with $R(p', \alpha)$ as the projection raw data and $h(p' - p)$ as the convolution kernel (Turbell 2001). $U(x, y, \alpha)$ and $V(p)$ are the spatial weighting factors for each reconstructed voxel with SDD = SID + IDD as the source-to-detector distance (SDD) (A.1b), (A.1c). The position $p'(x, y, \alpha)$ on the curved detector can be calculated via the arc length of a circle with the SDD as the radius (A.1d). The angle $\theta$ can be expressed by $x$ and $y$ with the use of a rotation matrix with the angle $\alpha$ (A.1e).

$$f_{\text{FBP}}(x, y) = \int_{0}^{2\pi} \frac{1}{U(x, y, \alpha)^2} \int_{-\infty}^{+\infty} V(p)R(p', \alpha)h(p' - p) \, dp \, d\alpha$$  \hspace{1cm} (A.1a)

$$U(x, y, \alpha) = \frac{\text{SID} + x \sin \alpha - y \cos \alpha}{\text{SID}}$$  \hspace{1cm} (A.1b)
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Figure A1. Geometric parameters important for the derivation of the FBP correction factor.

\[ V(p) = \cos \left( \frac{p}{\text{SDD}} \right) \]  
(A.1c)

\[ p'(x, y, \alpha) = \text{SDD} \cdot \theta(x, y, \alpha) \]  
(A.1d)

\[ \theta(x, y, \alpha) = \arctan \left( \frac{x \cos \alpha + y \sin \alpha}{\text{SID} + x \sin \alpha - y \cos \alpha} \right) \]  
(A.1e)

According to Yu et al (2004) and Zeng (2004), the spatially variant noise in the reconstructed images has its origin in the weighting factor \( U(x, y, \alpha) \) of the FBP. Especially for small SID and large fan angles \( \theta \), the factor \( U(x, y, \alpha)^{-2} \) gains for peripheral voxels which increases the noise in the periphery of the reconstructed image. This inhomogeneity should be compensated by the convolution kernel \( h(p' - p) \) with a lower cut-off frequency for these regions. Instead, a fixed cut-off frequency is used for all voxels which amplifies the high-frequency noise components and as a result the total noise in the periphery.

This irregularity can be expressed by the function \( F_{\text{FBP}}(\theta) \) which averages the factor \( U(x, y, \alpha)^{-2} \) over all voxels in \( x \)- and \( y \)-direction \( N_x, N_y \) and over all rotation angles \( N_\alpha \) in relation to the fan angle and detector position respectively. \( F_{\text{FBP}}(\theta) \), normalized to the central ray, was used to calculate an angle-dependent correction factor \( W_{\text{FBP}}(\theta) \) (A.2b) for the shaped filter design method HomNoise (section 2.2.1). With an increasing correction factor, the resulting filter thickness is reduced in the periphery (A.2c) which led to a more homogeneous noise in the reconstructed image.

\[ F_{\text{FBP}}(\theta) = \frac{1}{N_{x,y,\alpha}} \cdot \sum_{i=0}^{N_x} \sum_{j=0}^{N_y} \sum_{k=0}^{N_\alpha} \frac{1}{U(x, y, \alpha)^2} \bigg|_{\theta=\text{const.}} \]  
(A.2a)

\[ W_{\text{FBP}}(\theta) = \frac{F_{\text{FBP}}(\theta)}{F_{\text{FBP}}(0)} \]  
(A.2b)

\[ N_{\text{detector}}(\theta) = \sum_{i=1}^{E_{\text{max}}} N_i \cdot e^{-\mu(E_i) \cdot d(\theta)} W_{\text{FBP}}(\theta) \]  
(A.2c)

Figure A2(a) illustrates the detector noise profiles for the simulation of the breast phantom with a diameter of 140 mm. With a shaped filter of aluminium designed according to the
HomNoise method but without FBP correction, a homogeneous noise distribution was achieved at the detector. However, the noise profiles in the CT images showed still a higher noise in the periphery compared to the centre (figure A2(b)). With the use of the same filter, designed with FBP correction factor, the noise homogeneity in the reconstructed images was improved, though no absolute uniformity was reached. The cause for this might be the fact that the factor $U(x, y, \alpha)^{-2}$ can only be compensated by the shaped filter through one correction factor for all voxels between the x-ray source and a given detector element. Only the correction factor averaged over all voxels and projection angles was calculated by the method described above. It should be noted that a different derivation of the correction factor might yield better results.

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Figure A2. Detector noise profiles (a) and the corresponding CT image noise profiles (b) of the 140 mm breast phantom simulated with a shaped filter of aluminium designed according to the HomNoise method with and without FBP correction, compared to the simulation without shaped filter.
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