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Abstract. Antiscatter grids are used in digital mammography to reduce the scattered radiation from the breast and improve image contrast. They are, however, imperfect and lead to partial absorption of primary radiation, as well as failing to absorb all scattered radiation. Nevertheless, the general consensus has been that antiscatter grids improve image quality for the majority of breast types and sizes. There is, however, inconsistency in the literature, and recent results show that a substantial image quality improvement can be achieved even for thick breasts if the grid is disposed of. The purpose of this study was to investigate if differences in the considered imaging task and experimental setup could explain the different outcomes. We estimated the dose reduction that can be achieved if the grid were to be removed as a function of breast thickness with varying geometries and experimental conditions. Image quality was quantified by the signal-difference-to-noise ratio (SDNR) measured using an aluminum (Al) filter on blocks of poly(methyl methacrylate) (PMMA), and images were acquired with and without grid at a constant exposure. We also used a theoretical model validated with Monte Carlo simulations. Both theoretically and experimentally, the main finding was that when a large $4 \times 8 \text{ cm}^2$ Al filter was used, the SDNR values for the gridless images were overestimated up to 25% compared to the values for the small $1 \times 1 \text{ cm}^2$ filter, and gridless imaging was superior for any PMMA thickness. For the small Al filter, gridless imaging was only superior for PMMAs thinner than 4 cm. This discrepancy can be explained by a different sensitivity to and sampling of the angular scatter spread function, depending on the size of the contrast object. The experimental differences were eliminated either by using a smaller region of interest close to the edge of the large filter or by applying a technique of scatter correction by subtracting the estimated scatter image. These results explain the different conclusions reported in the literature and show the importance of the selection of measurement methods. Since the interesting structures in mammography are below the 1-cm scale, we advocate the use of smaller contrast objects for assessment of antiscatter grid performance. © 2015 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: [10.1117/1.JMI.2.1.013501](https://doi.org/10.1117/1.JMI.2.1.013501)]

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1 Introduction

In mammography, image quality is degraded by an additional background caused by scattered radiation.^{1–3} It was found by Wagner that in the ideal observer model, the scattered radiation would reduce the signal-to-noise ratio by a factor of $(1 + \text{SPR})^{-1}$,⁴ where SPR is the scatter-to-primary ratio. Using slot- and multislit-scanning geometries offers the potential advantage of inherent rejection of scattered radiations.^{5,6} Challenges with these geometries are increased image acquisition time and higher tube load. Increasing the air gap for scatter reduction has been studied by many researchers,^{7,8} but the drawbacks of this technique include increased focal spot blurring due to the higher magnification, reduced effective field of view,⁹ and practical limitation on the separation between breast support and detector. For most commercial mammography systems, the common way to reject scattered radiation is to place a grid between the breast and the detector, which can reduce the SPR to 2 to 20%,^{10–13} depending on the thickness of the breast and the grid characteristics. However, the major drawback of the grid is that a considerable fraction of the primary radiation is

blocked from reaching the detector, thus losing important image information.¹⁴

In screen-film mammography, where a certain exposure of the film is needed, a grid is necessary to improve the image contrast. To compensate for the photons absorbed by the grid, the x-ray exposure has to be increased.¹⁵ Because the dynamic range in digital mammography is relatively large, scattered radiation does not affect the relative differences between signals in images, but only increases quantum noise. Therefore, image quality is instead quantified by signal-difference-to-noise ratio (SDNR), a.k.a. contrast-to-noise ratio (CNR). Removing the grid allows more primary radiation to reach the detector, which might compensate for the additional quantum noise caused by scattered radiation. If so, using a grid is not beneficial. As for the image inhomogeneity caused by scattered radiation, this can be corrected for by software-based methods applied in postprocessing.¹⁶

Åslund et al.¹⁷ investigated the usefulness of a grid with a theoretical model. In this model, the SDNRs with and without a grid were compared using published data from digitized screen-film mammograms. The authors concluded that a grid could not improve image quality for thinner breasts ($<5 \text{ cm}$) but was advantageous for the thicker ones. Similar results

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can be found in Gennaro et al.'s study, where the grid performance was directly evaluated in a full-field digital mammography system.¹¹ However, these results differ substantially from the conclusion of Fieselmann et al.,¹⁸ which is based on a phantom study in which SDNR was measured according to the fourth edition of the European Guidelines protocol,¹⁹ with and without a grid. Fieselmann et al. showed that by removing the grid and applying a software-based scatter correction technique, the dose could be saved for any breast thickness without degrading the SDNR. In this study, we reevaluate the usefulness of a grid by investigating the dose reduction that may be achieved when the grid is removed in digital mammography. In particular, the aim is to study how the experimental conditions influence the results and whether such differences could explain the divergence of earlier results. First, we theoretically analyzed the dose reduction by making use of the published data measured in digital mammography. For validation, we then carried out Monte Carlo simulations and phantom studies.

2 Materials and Methods

2.1 Theoretical Model

We use the theoretical model introduced by Wagner et al.²⁰ to analyze the SDNRs with and without a grid in digital mammography. Assuming a quantum-limited system, the total signal I received by a detector is the sum of scattered radiation S and primary radiation P . In the absence of a grid, SDNR is written as

$$\text{SDNR} = \frac{I_1 - I_2}{\sqrt{I_1}} = \frac{(P_1 + S_1) - (P_2 + S_2)}{\sqrt{P_1 + S_1}}, \quad (1)$$

where I_1 and I_2 refer to the total signals in the background and under a target, respectively.

When a grid is present, both the scattered and primary radiation would be partly attenuated, and then the SDNR value becomes

$$\text{SDNR}^{\text{grid}} = \frac{(T_p P_1 + T_s S_1) - (T_p P_2 + T_s S_2)}{\sqrt{T_p P_1 + T_s S_1}}, \quad (2)$$

where T_s and T_p are the transmission of scattered and primary radiation through the grid, respectively.

We define the SDNR improvement factor K as the ratio between the SDNRs obtained with and without a grid at the same patient dose. We assume that the amount of scattered radiation is the same beneath the target as in the background, i.e., $S_1 \approx S_2$. The factor K is then given by

$$K = \frac{\text{SDNR}^{\text{grid}}}{\text{SDNR}} = \frac{T_p \sqrt{P_1 + S_1}}{\sqrt{T_p P_1 + T_s S_1}} = \sqrt{T_p} \sqrt{\frac{1 + \text{SPR}}{1 + \text{SPR}^{\text{grid}}}}, \quad (3)$$

where SPR and SPR^{grid} are the scatter-to-primary ratios obtained without and with the grid, respectively.

We calculated the factor K with the values of SPR, SPR^{grid} , and T_p measured by Salvagnini et al.²¹ as a function of poly (methyl methacrylate) (PMMA) thickness using the beam-stop method on a digital mammography system, Siemens MAMMOMAT Inspiration. The system has a common linear grid, with a lead septa of 5:1 ratio and strip density of 31 lines/cm. For each PMMA thickness, the corresponding

SPR, SPR^{grid} , and T_p were averaged over the different potentials of 26, 28, 30, and 32 kV (Table 1).

We used the dose reduction factor D (i.e., the percent of the patient dose saved without degrading the SDNR) to quantify the benefit of removing the grid. Because SDNR is linearly proportional to the square root of the patient dose in a quantum-limited system,^{11,22} the resulting SDNR improvement factor was translated into the dose reduction factor D by

$$D = 1 - K^2, \quad (4)$$

where a negative value of D stands for the dose increasing as a result of grid removal.

2.2 Validation

2.2.1 Monte Carlo simulation

The scatter image $S(x, y)$ on the detector plane was constructed by convolving the point spread function (PSF) of scattered radiation with the image field, which is a method similar to that described by Boone and Cooper:²³

$$S(x, y) = (F_{\text{Al}} \otimes \text{PSF}_{\text{Al}})(x, y) + (F_{\text{bg}} \otimes \text{PSF}_{\text{bg}})(x, y) \times \begin{cases} F_{\text{Al}} = 0, F_{\text{bg}} = 1; & \text{in the background} \\ F_{\text{Al}} = 1, F_{\text{bg}} = 0; & \text{beneath the Al filter} \end{cases}, \quad (5)$$

where F_{bg} and F_{Al} refer to the background field and the field covered by an aluminum (Al) filter, respectively. PSF_{bg} and PSF_{Al} are their corresponding PSFs of scattered radiation.

The scattered PSFs were computed using PENELOPE Monte Carlo code.²⁴ To compare the results with phantom studies, the simulation was designed to use the settings in the following phantom studies. A range of 2 to 8 cm of PMMA layers was selected as the attenuation medium. The air gap between the bottom of the PMMA and detector was 1.5 cm. The interspace of the grid was assumed to be filled with PMMA 0.30 mm thick. Both sides of the grid were considered to be covered by a PMMA layer 1.25 mm thick.

PSF_{bg} for each PMMA thickness was obtained by tracking 10^8 x-ray photons that are normally incident at the center point of the PMMA surface (Fig. 1). Both coherent and incoherent scatters were considered in the simulation. If the photons exited from the bottom of the PMMA layers, their positions on the detector plane were recorded as they passed through the air gap. We assumed that the deposited energy of each photon

Table 1 The scatter-to-primary ratios with and without a grid (SPR^{grid} and SPR) and the primary transmission T_p , measured as a function of PMMA thickness by Salvagnini et al.²¹ on a full-field digital system. These values are the average results over the different tube potentials of 26, 28, 30, and 32 kV.

PMMA (mm)	SPR^{grid}	SPR	T_p
20	0.052	0.268	0.750
40	0.089	0.541	0.748
60	0.154	0.842	0.750
80	0.187	1.124	0.753

Fig. 1 The schematic of Monte Carlo simulation for a point spread function of scattered radiation.

reaching the detector was completely absorbed in an ideal energy-integrating system. When the grid was present in the simulation, the transmission of each exiting photon through the grid was calculated using the analytical equations derived by Day and Dance.²⁵ The secondary scattered radiation in the grid was neglected because it has only a small effect, according to Boone et al.⁹ Similarly, PSF_{AI} was generated by adding an extra 2-mm-thick Al filter on top of the PMMA in the simulation. For an Al filter of a given size, the scatter images $S(x, y)$ with and without grid were constructed by taking the resulting PSF_{bg} and PSF_{AI} into Eq. (5). To investigate the influence of the size of the Al filter, two different sizes of filters, 4×8 and 1×1 cm², were applied in the simulation.

Ultimately, the phantom image $I(x, y)$ was calculated by summing the scatter image $S(x, y)$ and its corresponding primary image $P(x, y)$ together, given by:

$$I(x, y) = S(x, y) + P(x, y); \begin{cases} P(x, y) = P_{bg}; & \text{in the background} \\ P(x, y) = P_{AI}; & \text{beneath the Al filter} \end{cases} \quad (6)$$

where P_{AI} and P_{bg} refer to the primary radiation under the Al filter and in the background, respectively.

With properly selected regions of interest (ROIs) in the simulated phantom images, the SDNR values for both grid-in and gridless cases were calculated as:

$$SDNR = \frac{|I_{AI} - I_{bg}|}{\sqrt{I_{bg}}}, \quad (7)$$

where I_{bg} and I_{AI} refer to the mean signal of the ROI in the background and under AI, respectively.

2.2.2 Phantom study

The phantom study was performed on a full-field digital mammography system (Philips MammoDiagnost DR), with the same grid properties as the system used by Fieselmann et al.¹⁸ The system uses a-Se as the detector material. The pixel matrix is 2816×3584 with an $85 \mu\text{m}$ pitch, giving an image size of $24 \text{ cm} \times 30 \text{ cm}$. The Philips MammoDiagnost DR system has two anode targets of Mo and W and two selectable filters of Mo and Rh. The grid in the system is the same one used in Salvagnini et al.'s study.²¹

PMMA layers 2 to 8 cm thick in 1-cm steps simulated breasts of different thicknesses. The SDNR values were determined by a 4×8 cm², 0.2 mm Al filter placed on top of the PMMA with its left edge superimposed on the center line of the PMMA layers (Fig. 2). Images were acquired in the manual exposure mode with a W-Rh anode-filter combination for all PMMA thicknesses. The tube potential and milliampere second (mAs) were appropriately selected to ensure an adequate exposure (in Table 2). For each PMMA thickness, the phantom images with and without the grid were acquired using the same exposure setting.

In each image, two 0.5×0.5 cm² ROIs on the Al filter were selected to compute the SDNR, one (ROI₁) at the center, as suggested by the fourth edition of the European Guidelines,¹⁹ and the other (ROI₂) at the edge for comparison. The reference ROIs in the background were symmetrically selected along the center line. For each pair of ROIs, the SDNR was measured by calculating the difference between the mean signal I_{AI} of the ROI under the Al filter and the mean signal I_{bg} of the ROI in the background, divided by the standard deviation of the ROIs in the background σ_{bg} :

$$SDNR = \frac{|I_{AI} - I_{bg}|}{\sigma_{bg}}. \quad (8)$$

Fig. 2 The experimental setup. (a) Front and (b) top view. A 4×8 cm², 0.2-mm-thick Al filter was placed on top of PMMA layers with a total thickness varying between 2 and 8 cm. The signals under the Al filter were determined by two 0.5×0.5 cm² regions of interest (ROIs), which are 6 cm from the chest wall. One (ROI₁) was centered on the Al filter, and the other one (ROI₂) at the edge. The reference ROIs in the background were symmetrically selected.

Table 2 Manually selected anode-filter combinations, tube potentials (kVp), and mAs values, according to phantom thicknesses.

PMMA (mm)	Anode/filter	kVp	mAs
20	W/Rh	24	45
30	W/Rh	25	71
40	W/Rh	27	100
50	W/Rh	29	140
60	W/Rh	31	200
70	W/Rh	32	280
80	W/Rh	33	450

The dose reduction factor D was calculated according to Eq. (4). The standard deviation of factor D was estimated by repeating the measurement three times.

The fourth edition of the European Guidelines does not specify the size of the Al filter, but in a recently published supplement, a $1 \times 1 \text{ cm}^2$ Al filter is suggested for SDNR measurement.²⁶ To study the effect of the size of the Al filter on SDNR, we repeated the experiment with a small $1 \times 1 \text{ cm}^2$ Al filter. The SDNR values were determined by the ROI centered on the Al filter.

Because scattered radiation is nonuniformly distributed in the phantom images, it might affect the SDNR values in two ways. First, the signal difference would be affected if the average amount of scattered radiation in the selected ROI is not the same on the Al filter as it is in the background. Second, the noise is probably overestimated, given the low-frequency varying background in the ROIs. Therefore, the nonuniformity of scattered radiation should be removed to correctly evaluate the SDNR. Here, we introduce a method to remove the nonuniformity of scattered radiation on images. The method has three steps:

Step 1: Subtract the grid-in image from the gridless image with the same PMMA thickness, given by:

$$I_s = \frac{T_p \times I^{\text{nongrid}} - I^{\text{grid}}}{T_p - T_s}, \quad (9)$$

where the primary transmission T_p and scatter transmission T_s were determined according to the simulation in Sec. 2.2.1.

Step 2: The low-frequency scattered radiation was obtained by smoothing the subtracted image I_s with a low-pass filter:

$$I'_s = (I_s)_{\text{lowpass}}. \quad (10)$$

Step 3: The low-frequency scattered radiation was removed from the original images. For gridless images, it was removed by:

$$I_c^{\text{nongrid}} = I^{\text{nongrid}} - I'_s. \quad (11)$$

Considering that a small fraction of the scattered photons still transmit through the grid, the images acquired with the grid were also corrected with

$$I_c^{\text{grid}} = I^{\text{grid}} - T_s \times I'_s. \quad (12)$$

For comparison, we have included the values of the dose reduction factor D before and after the scattered radiation was removed.

3 Results

3.1 Theoretical Result

In Fig. 3, the dose reduction factor D as a function of PMMA thickness, based on the theoretical model and the data published by Salvagnini et al.,²¹ is compared with the results by Åslund et al.,¹⁷ which were derived from the screen-film data. The breast thicknesses in Åslund et al.'s study were converted into the equivalent PMMA thicknesses using the conversion factors described by Dance et al.²⁷ The dose reduction factor D was fitted as a linear function of PMMA thickness. We found a cross-over of 3.3 cm in our study and 4.4 cm for Åslund et al.'s study. The theoretical analysis shows that for a thinner PMMA, <3.3 to 4.4 cm, removing the grid could reduce the dose without degrading image quality, whereas for a thicker PMMA, > ~4.4 cm, the grid provides benefit and should be kept.

3.2 Results from Simulations and Phantom Studies

Figure 4 shows the simulated primary and scattered radiation for a 7-cm-thick PMMA without a grid. It is found that the amount of scattered radiation under Al is much less than that in the background for the $4 \times 8 \text{ cm}^2$ Al filter. However, this result is not the case in examined breasts in which targets are usually small, so that the scattered radiation under targets is the same as that in the background. When the large Al filter is used, the resulting SDNR value for gridless images is overestimated as a result of a different amount of scattered radiation in the background region and in the region covered by the Al filter. Therefore, the dose reduction estimated from the simulation must be corrected by removing this effect.

Figure 5 shows an example of scatter removal. Figure 5(a) is the original image I^{nongrid} without a grid for a 7 cm PMMA. Figure 5(b) is the corresponding scatter field I_s obtained by subtracting the grid-in image from the gridless image according to Eq. (9). The primary and scatter transmissions, T_p and T_s , used

Fig. 3 The dose reduction factor D as a function of PMMA thickness, derived from the theoretical model and the digital data (circles). The results by Åslund et al.¹⁷ are based on screen-film data (diamonds).

Fig. 4 The simulated primary and scattered radiation for a 7-cm-thick PMMA without a grid, along a horizontal line 6 cm away from the chest wall, using Al filters of different sizes, 4×8 and 1×1 cm².

for the scatter removal were estimated from the simulation as a function of PMMA thickness, shown in Table 3. The average scatter and primary transmissions over different PMMA thicknesses are 0.137 and 0.756, respectively, which agree well with the results of 0.13 and 0.75 measured by Salvagnini et al.,²¹ with the same grid and anode-filter combination. Figure 5(c) is the corrected image I_c obtained by subtracting the scatter field from the original image. Figure 6 shows the signal distributions of the images I_{nongrid} , I_s , and I_c , along a horizontal line 6 cm away from the chest wall. The same approach was performed to remove the scattered radiation from the grid-in images, which is not reported in the paper.

The dose reduction factors calculated from the simulations are illustrated as stars in Fig. 7, where the nonuniformity of scattered radiation has been corrected for. It shows that the dose can be preserved without a grid for PMMAs thinner than ~ 4 cm, which agrees well with the theoretical results.

For the phantom study, the resulting dose reduction factors calculated from the large Al filter before scattered radiation was removed are plotted as squares and triangles for ROI₁ and ROI₂ in Fig. 7, respectively. The dose reduction factors of ROI₂ are consistent with the simulation results, whereas the values of ROI₁ are overestimated due to the additional contribution from the scattered radiation.

Fig. 5 An example of removing scattered radiation for a 7 cm PMMA phantom. (a) The original phantom image I_{nongrid} . (b) The scatter image I_s extracted from the original one. (c) The corrected image I_c has a more uniform distribution after the scatter has been removed (the gray scale in each image is adjusted for visualization).

4 Discussion

The theoretical model showed that removing the grid could reduce the patient dose for PMMA thicknesses up to 3.3 cm without decreasing the image quality, which can be compared to the earlier results by Åslund et al. based on screen-film data, where the cross-over was at a 4.4 cm PMMA thickness. This discrepancy can probably be attributed to different geometries and exposure parameters.

The phantom study confirmed the theoretical results and, except when using a large filter with a central evaluation ROI, all experimental cases showed that the thickness below which removing the grid is beneficial is between 4.5 and 5 cm PMMA (Fig. 7). This thickness is close to the center of the compressed breast thickness distribution of the female population. The Monte Carlo simulation corroborated the experiments and shows why the case with a large Al filter gives a different result (red squares in Fig. 7). Figure 4 illustrates that the scatter at the center of the large filter is lower than that close to the filter edges and in the background, resulting in an underestimated SPR without grid. This setup does not correspond well to the clinical imaging task, where targets and interesting structures are small, and the scattered radiation under those targets is very close to that of the surrounding background. Positioning the ROI at the edge of the large filter (Fig. 7, cyan triangles) resulted in the same outcome as for the small filter (Fig. 7, circles). Application of scatter correction to the images did not affect the results for the small filter, but, in this case, changed the result for the large filter to the same as for the small one (Fig. 7, crosses).

Software-based scatter correction methods can remove the low-frequency background noise caused by scatter and are useful for quantitative imaging, such as dual energy imaging and breast density estimation.¹⁶ But the procedure does not affect the properties of the local signal difference and the noise and, hence, does not affect SDNR, except for regions in the image with steep gradients. Ducote and Molloy also show that scatter correction can improve CNR by up to 23%. It should be noted, though, that they define CNR as contrast divided by noise (as the acronym suggests), which differs from the conventional definition of CNR or SDNR as the signal difference divided by the noise, which is the definition used in this work as well as the European Guidelines and is based on the Rose model of visual perception. An analysis of the data provided in their article shows that when the conventional definition is used, the result corroborates the findings in this work, but we omit the details here.

Table 3 The primary and scatter transmissions (T_p and T_s) through the grid for different PMMA thicknesses, obtained from the simulation with the same exposure conditions in the phantom study.

PMMA (mm)	Primary transmission T_p	Scatter transmission T_s
20	0.720	0.121
30	0.728	0.127
40	0.741	0.135
50	0.756	0.138
60	0.770	0.142
70	0.782	0.148
80	0.793	0.151

The fourth edition of the European Guidelines, on which the work by Fieselman et al. was based, does not specify the size of the Al filter. In a recently published supplement,²⁶ though, a $1 \times 1 \text{ cm}^2$ Al filter is specified for SDNR measurement. While the guidelines do not mention or advocate the measurement method for scatter quantification, we still believe that this is an important change in that context.

While grids are standard for two-dimensional mammography, the usage for tomosynthesis or three-dimensional imaging is more technically challenging, since the source is moving in relation to the breast support and the detector.²⁸ Some commercially available systems use grids in the tomosynthesis mode while others do not.²⁹ Because the signal difference and noise in the projection images should be translated to the reconstructed image volume, we believe the results presented in this work are also valid for tomosynthesis, but this remains to be investigated in detail and is outside of the scope here.

It has to be noted that the results presented in this paper were obtained based on a linear grid with a grid ratio of 5 and an air gap of 1.5 cm, both of which are typically used in commercial mammography systems. The SDNR can be further improved by

Fig. 6 The measured signal distributions along a horizontal line 6 cm away from the chest wall, for the images I_{nongrid} , I_s , and I_c in Fig. 5. **Fig. 7** The dose reduction factor D as a function of PMMA thickness, obtained from the simulation and the phantom studies.

using a grid with a higher efficiency of scatter reduction, such as a cellular grid. In the simulation, we observed that there is little variation of SPR when varying the air gap between 0 and 3 cm, consistent with the conclusion by Boone and Cooper for a field of view of a 15-cm-diameter circle.²³ In the future, it will be interesting to perform a more comprehensive study considering a wide variety of parameters, such as breast composition and field of view.

5 Conclusions

We re-evaluated the usefulness of a grid by investigating the possible dose reduction resulting from the grid removal in digital mammography. The dose reduction was determined as a function of PMMA thickness by comparing the SDNRs with and without a grid at the same dose level. With the theoretical model, we found that removing the grid could reduce the dose for PMMA thicknesses up to 3.3 cm without decreasing the SDNR, confirming the results by Åslund et al. derived from screen-film data.

The theoretical results were validated by Monte Carlo simulations and phantom studies. We concluded that keeping the grid is optimal for PMMAs above ~ 4 cm, whereas the dose can be reduced without a grid for thinner PMMAs. We also found that the size of the Al filter is an important factor in measuring SDNR without a grid, and that using a large Al filter overestimates the value because the amount of scattered radiation in the background differs from that under the filter. This finding provides new insights for explaining the conflicting conclusions of earlier investigations, and results showing that removal of the grid is beneficial in the whole breast thickness range can probably be attributed to the large Al filter used. Such a setup does not correspond well to the clinical imaging task and we advocate the use of small targets for quantification of scatter and scatter removal schemes.

Acknowledgments

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