White paper

Optimal grids for advanced digital radiography

State-of-the-art components
for perfect outcomes

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1. Introduction

Reducing scattered radiation is crucial to ensuring good image quality in radiography. But while X-ray systems have evolved dramatically, anti-scatter devices have often not kept pace and remained essentially unchanged for decades. The shortcomings of the standard anti-scatter grids that have been in use since the introduction of digital radiography have become evident with the recent advent of high-resolution detectors and their improved spatial resolution. This has triggered a reconsideration of the geometrical construction principles used to design anti-scatter grids and led to the development of a new generation of grids with low line density that are more efficient in filtering out scattered radiation.

In this white paper, we will explain the fundamentals of scatter reduction and show how the redesigned grids significantly enhance signal-to-noise ratio (SNR) by reducing the fraction of scattered radiation to increase the share of primary radiation that reaches the detector. This allows radiography departments and radiology practices to fully benefit from the higher resolution of the new HD detectors, and it enables the use of the same protocol database for both normal and high-resolution detectors.

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2. Scattered radiation and its impact on image quality

Scattered radiation is a major image-degrading factor in radiography [1]. It's produced when X-rays move through the human body and are deflected at an angle from their original direction. The more tissue the X-rays penetrate, the more of the primary radiation is turned into scatter. In the abdominal region of an adult, for example, 60 to 70 % of the radiation is deflected in all directions [2]. This creates a fog of obliquely incident X-rays [3] that's spread more or less evenly throughout the field of view (FoV) of the detector behind the patient [Fig. 1]. The scattered radiation obscures the differences in primary radiation that is distributed unevenly depending on the anatomy of the patient (very little primary radiation behind solid structures like bones, but high primary radiation behind soft tissue). Scattered radiation therefore counteracts the very principle of X-ray imaging.

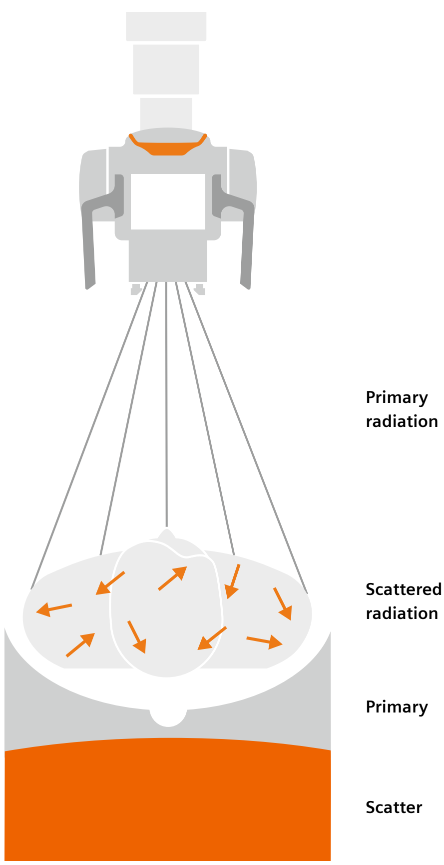

Figure 1:
Typical distribution of radiation in the abdominal region. The visibility of the X-ray image created by the primary radiation is obscured by the evenly distributed scattered radiation that's generated in the irradiated parts of the body and emitted in all directions.

It's clear that the image quality, usually defined by the signal-to-noise ratio (SNR), is directly dependent on the amount of primary photon fluence (φ_P) and the corresponding scattered photon fluence (φ_S) that add up to the total photon fluence behind the patient (φ_0). In mathematical terms, the relationship can be expressed as:

$$SNR^2 = \frac{{\varphi_P}^2}{\varphi_P + \varphi_S} = (1\text{-}S)^2 \cdot \varphi_O$$

In this equation, S represents the fraction of scattered radiation in the image $1/\varphi_S$. $1 - S$ is the scatter degradation factor. This is a simplified equation that supposes parallel primary radiation throughout the FoV and doesn't take de-centering and de-focusing into account. The angulation created by the source-to-image distance (SID) is neglected, as is the increased primary radiation toward the edges of the body due to the reduction of irradiated tissue. However, the general principle obviously holds true: A smaller amount of scattered radiation (φ_S) significantly improves the SNR and creates a much better image.

3. Scatter reduction with an anti-scatter grid

The negative impact of scattered radiation in X-ray imaging has been known for a long time, which is why radiographers have been experimenting with scatter reduction techniques for more than 100 years. In 1913, Gustav Peter Bucky invented a wafer-shaped grid that was placed behind the patient to filter out the deflected scatter from the linear primary radiation. Seven years later, Hollis E. Potter improved this with a grid made up of parallel unidirectional lead strips. The Bucky-Potter grid [Fig. 2] is still in use today and has proven to be the most effective method for reducing scattered radiation [4]. The structure of the grid, however, has been repeatedly redesigned, because the selective absorption of scatter has become more and more important with the perfection of the X-ray imaging process [5].

The most recent step in the evolution of the radiographic imaging chain was the introduction of HD detectors like the X.wi-D from Siemens Healthineers that reduce the pixel size to 99 µm and enable a spatial resolution of 5.06 lp/mm. Especially at large zoom factors, this has led to problems with the existing standard grids, which have a large number of lead strips per cm. As a result, the older reference grids have come under scrutiny, and a new generation of optimized grids has been developed.

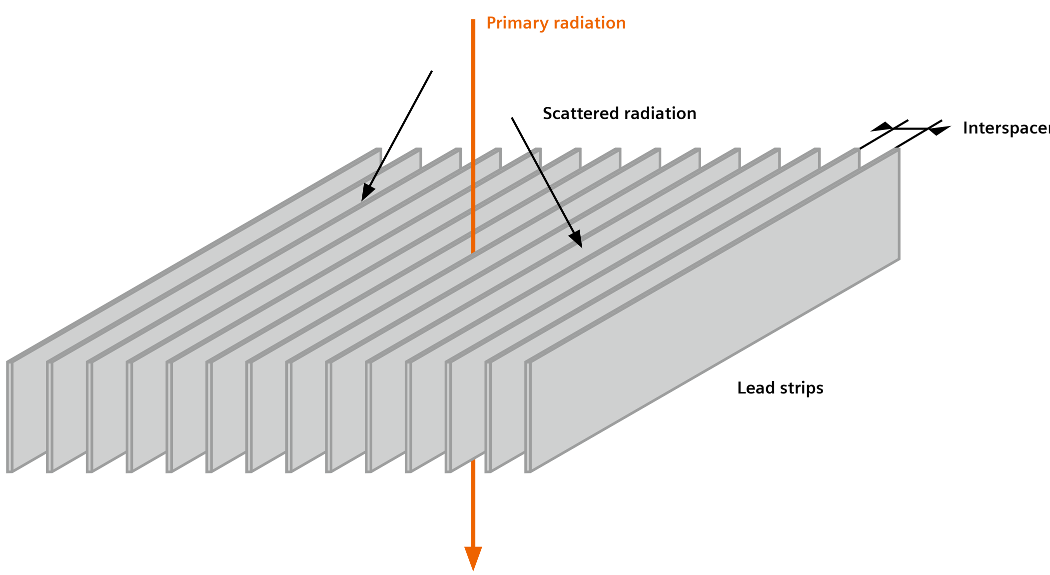

Figure 2:
Anti-scatter grid design. The basic design of the Bucky-Potter grid has remained unchanged. As the primary radiation passes through the interspacer material of the grid, the scattered X-rays are absorbed by the lead strips.

4. Features of anti-scatter grids

To understand how anti-scatter grids have been improved, we first need to look at the geometric parameters that define them [Fig. 3]. The main elements that characterize grids are the height of the lead strips (h) and their distance apart (D). This is represented by the aspect ratio (r), which is calculated as $r = h/D$. This is the most frequently cited quality factor, because a higher aspect ratio in two otherwise identical grids yields a greater contrast improvement [6].

What's missing in this consideration is the line density that also takes the thickness of the strips (d) into account, as this also has an influence on scatter reduction. The line density (N) is usually described as the number of strips per cm: $N = 1/(D+d)$. To find the optimal distribution for a given grid, the aspect ratio needs to be checked against the line density. Anti-scatter grids are therefore notated with a combination of r and N: for example, r13N92. This describes a standard grid with a high aspect ratio and a high line density that's been used extensively as a reference in digital radiography because its grid lines don't interfere with the pixel size used in the detectors.

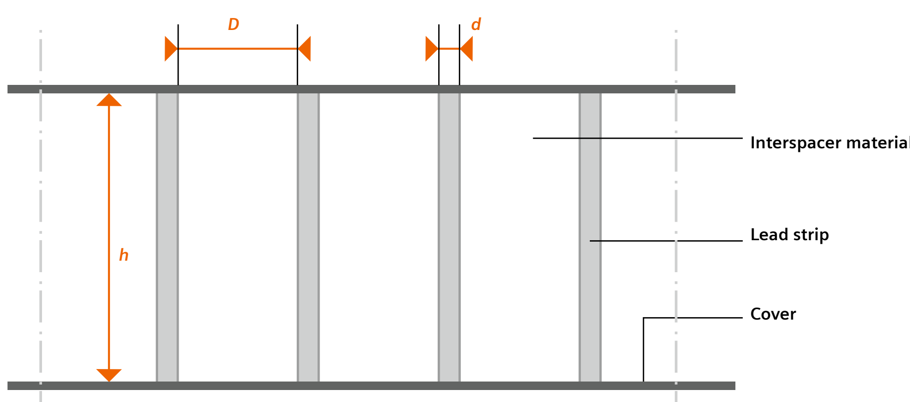

Figure 3:
The main parameters that characterize anti-scatter grids. This is a schematic depiction that doesn't show that the lead strips need to be tilted slightly toward the X-ray tube based on the source-to-image distance (SID) in order to provide a clear focus.

5. Signal-to-noise ratio improvement due to lower line density

We might expect that a high line density results in greater absorption of scattered radiation, resulting in better image quality. However, for a certain aspect ratio, the exact opposite is the case [7]. Siemens Healthineers has used this principle to optimize its anti-scatter grids. The redesigned r13N40 grid developed in 2024 keeps the aspect ratio of the older grids but has less than half the number of lead strips per cm.

This means that the improved grids are much thicker than the r13N92 grids. This results in an enhanced SNR that can be measured by the quality factor Q ($Q = SNR_{without\ grid} / SNR_{with\ grid}$). A quality factor of 1 means that the image quality isn't improved at all, while a higher quality factor indicates how much the SNR is improved by the grid.

If we compare typical values of the two grids [Table 1], we can see that at roughly the same amount of transmitted primary radiation (T_P), the r13N40 grid transmits a much smaller amount scattered radiation (T_S) than the r13N92 grid. The total radiation transmission can be calculated as follows: $T_T=T_P{\cdot}(1\text{-}S)+T_S{\cdot}S$

Grid type	Primary (T_P)	Scatter (T_S)	Total (T_T)	Quality Factor (Q)
r13N92	64.8%	19.6%	32.4%	1.75
r13N40	64%	6%	16%	2.50

Table 1:
Typical values for different grids (according to IEC 60627) (Data on file)

We can use the formula from page 3 to calculate a grid's SNR:

$$SNR^2_{Grid} = \frac{\varphi_P^{\,2}}{\varphi_P+\varphi_S} = \frac{T_P}{T_T}\,(1\text{-}S)^2\cdot\varphi_O$$

In this equation, $(1\text{-}S)^2{\cdot}\varphi_0$ represents the SNR without a grid, which allows us to calculate the quality factor for both grids. The current grid increases the SNR by a factor of 1.75, while the new grid raises it by a factor of 2.50. In relative terms, this corresponds to a 75% increase versus a 150% increase – meaning the new grid delivers twice the SNR improvement compared to the current one.

How this advantage manifests in clinical practice can be seen in the example of whole-spine exams that cover a broad range of anatomical regions with different densities [Fig. 4]. The image shows the enhanced image quality that can be achieved with the r13N40 grid throughout the entire anatomy examined.

Note that the r13N40 grid also typically absorbs more radiation than the older grid, and this is reflected in the lower T_T. Whether this needs to be compensated by a higher patient dose in order to achieve a diagnostic image depends on the specific exam circumstances. The r13N40 grid's higher quality factor can also be traded for a lower entrance dose to correspond with the risk profile of the patient and achieve only the contrast improvement required by the specific exam.

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Figure 4:
Whole-spine exam (a.p.) of a 16-year-old female patient
Acquired with YSIO X.pree, a r13N40 grid, and a X.wi-D detector
Dose: 3.731 µGy*m²

5. Summary

Anti-scatter grids are widely used as the most effective means for reducing scattered radiation and improving image quality in radiography. However, their geometrical design needs to be continually upgraded in order to realize the full potential of improvements in the radiographic imaging chain. To match the high spatial resolution of the new X.wi-D detectors, Siemens Healthineers has optimized the standard grid by reducing the line frequency. This significantly improves the SNR and allows radiography departments and radiology practices to use the same protocol database for both normal and HD detectors.

List of abbreviations and symbols

D	distance of two lead strips (= thickness of the interspacer) in a scatter grid
d	thickness of the lead strips in a scatter grid
FoV	field of view
h	height of the lead strips in a scatter grid
HD	high-definition
N	line density of a scatter grid (= number of lead strips per cm): $N = 1/(D+d)$
Q	quality factor: $Q = SNR_{without\ grid}/SNR_{with\ grid}$
r	aspect ratio, also called grid ratio (= ratio of strip height to distance): $r = h/D$
S	fraction of scattered radiation
SID	source-to-image distance
SNR	signal-to-noise ratio
T_P	transmission of primary radiation
T_S	transmission of scattered radiation
T	total radiation transmission: $T_T = T_p \cdot (1-S) + T_S \cdot S$
φ_0	total photon fluence behind the patient: $\varphi_0 = \varphi_P + \varphi_S$
φ_P	primary photon fluence behind the patient
φ_S	scattered photon fluence behind the patient

References

[1] Chan HP, Doi K. Physical characteristics of scattered radiation and the performance of antiscatter grids in diagnostic radiology. RadioGraphics 2.3, August 1982. 378-406. p. 378.

[2] Data on file.

[3] Hondius Boldingh W. Grids to reduce scattered X-rays in medical radiography. [Phd Thesis 1 (Research TU/e/Graduation TU/e), Applied Physics and Science Education]. Technische Hogeschool Eindhoven, 1964. https://doi.org/10.6100/IR62689. p.1.

[4] Shah GA, Hassam G, Newman DL. The effectiveness of antiscatter techniques. Radiography 2, 1996. 191-197.

[5] Hondius Boldingh, p. 3.

[6] Chan and Doi, p. 402, Fig. 27.

[7] Ibid.

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