Abstract

In X-ray imaging, anti-scatter grids are used to reduce scatter radiation reaching image receptors, hence improving image quality. Optimization of grid performance is essential for improving image diagnostic quality and minimizing radiation doses to patients. This work investigated the performance of a series of grid designs modeled from the design of typically focused grid with grid ratio 8:1 (r8) and strip height 1.7 mm (h1.7) for high-energy radiographic applications. Monte Carlo simulation was used to evaluate designs (r8h1.7) which had the strip thickness changed from 6 to 150 μm in 2 μm increments and the interspace distance fixed at 214 μm. The transmissions of radiation in grid materials were modeled by using a regression with radial-basis-function-networks (RBFNS). K_{SNR} was then determined from RBFNS models of radiation transmissions. The optimal strip-thickness was obtained at the maximum signal-to-noise ratio (SNR) improvement factor (K_{SNR}). For high-energy applications at 100 peak-kilo-voltage (kVp) and 30 cm PMMA thickness, the optimal lead-stripe-thickness was found approximately 74 μm resulting in a strip-frequency approximately 35 per cm (N35). Using the optimal thickness for imaging condition at 100 kVp and 30 cm thickness, the K_{SNR} would increase by approximately 5.3%. This work showed the existence of optimal strip-thickness for a series of grids with a given grid-ratio, strip-height, strip-, and interspace materials. The findings are useful and provide guidance to improve grid designs for better performance that will essentially lead to better image quality and better radiation protection for patients.

Keywords: anti-scatter grids; Monte Carlo simulation; optimal strip-thickness; signal-to-noise ratio improvement factor

1 INTRODUCTION

In digital radiography, anti-scatter grids are essentially important for improving the image quality of large anatomical regions, such as the abdomen examination of large patient sizes above 30 cm thickness. Grids are used to reduce scattering radiation reaching the image receptor and hence to improve the image quality.[1] Linear focused grids are the common grids that consist of radiopaque strips and radiolucent interspaces and that have the strips aligned to focus on a common line at a finite distance (Figure 1). Grids can be used either in a stationary or moving situation. An X-ray imaging system can be equipped with a Bucky device[2] which supports the grid and provides a moving mechanism during image acquisition to eliminate gridlines (also known as grid-line artifacts) which are the projections of grid strips appearing in the final radiograph. In screen-film radiography and computed radiography, a grid is preferably used in a Bucky with a moving mechanism. With digital image receptors, the operation of Bucky devices during the image acquisition often interferes with the receptors and leads to an increase in noise.[3, 4] High-resolution digital image receptors are used with a stationary grid as grid lines become more apparent for grids with low strip frequency than for grids with high strip frequency.[3, 4] However, gridlines resulted from grids with high-grid-frequency, such as 70 lines cm^{-1}, are still significant and interfere with the visibility of low contrast details.[4] When stationary grids are used with digital image receptors, grid-line artifacts may be removed by image processing techniques.[3, 5–9]
The principal purpose of grids is to reduce scatter radiation reaching the image receptor. However, grids also reduce primary radiation (the useful information or signal) reaching the image receptor and hence when used, an increase in patient exposures is needed to maintain image diagnostic quality.[10, 11] Grid performance can be expressed with grid-physical parameters, such as the transmission of the primary radiation (Tp); the transmission of scatter radiation (Ts); and the transmission of the total radiation (Tt).[12, 13] Tp, Ts, and Tt vary from grid designs; and for the same grid-ratio and the same grid materials, Tp, Ts, and Tt vary a great extent, up to 13%, 23%, and 21%, respectively.[14–16] There is no information found about grid design to optimize grid performance. It is accepted that forward scatter radiation in the narrow-beam method[12] influences Tp and forward scatter radiation which should arise in the materials underneath the blocker affects Ts.[13, 17] Tp, Ts, and Tt are used to calculate the combined effects of the differential attenuation of the primary radiation and scatter reduction (noise reduction) on image diagnostic quality. Some measures of these combined effects are the image contrast improvement factor (K)[12] and the quantum signal-to-noise ratio (SNR) improvement factor (K_{SNR}).[13, 15, 18, 19]

$K_{SNR}$ is an important measure of the combination of image diagnostic quality improvement and radiation protection. A better radiation protection can be indicated by a greater Tp. A better diagnostic quality improvement can be indicated by a smaller Ts. The quality of a grid is measured in a compromization between radiation protection and image quality improvement. The quality of digital images depends mainly on the differential attenuation of primary radiation reaching the image receptor, not the total radiation exposure to the image receptor. For a given imaging system used with a grid, the grid's $K_{SNR}$ can be used as a suitable indicator for the optimization of digital images only if the X-ray tube peak-kilo-voltage (kVp) and the imaging object are not changed; otherwise, the differential attenuation of the primary radiation determined by the radiation beam quality and the physical components of the imaged object would confound the results as shown by Moore, Wood.[20]

Grid design optimization is essential for improving image diagnostic quality and minimizing radiation exposures to patients. The Tp of focused grids can be increased by reducing the strip thickness and increasing the grid ratio.[21] This will lead to the best grid being the one with the largest grid ratio which then has the greatest grid application limits.[22] In screen-film imaging systems, the strip thickness 100 µm for a grid with grid ratio 12:1 was recommended for a general grid.[23] A 5 µm strip-thickness was found optimal for a mammographic grid with grid-ratio 5:1.[24] Grid optimization for digital general planar radiography has not been systematically investigated. Grid design optimization would involve the evaluation of many grids and is impractical by experimental measurements due to time-consuming and economically high cost. Monte Carlo simulation is a feasible method for the study of grid design optimization.
This work is an investigation of grid strip-thickness optimization for linear-focused grids used in digital planar radiographic systems and aims to provide a method to determine an optimal strip-thickness to maximize the grid performance.

2 THEORY OF OPTIMAL THICKNESSES OF GRID STRIPS

In planar radiography, a notation of a linear-focused grid is often given as its strip frequency \((N)\) and grid ratio \((r)\), for example, \(Nxry\) where \(x\) and \(y\) are their respective values.\([13–16]\) The grid-strip frequency, \(N = 1/(d + D)\), is a measure of the number of strips per unit length, and the grid ratio, \(r = h/D\), is the ratio of the strip height \((h)\) to the interspace distance \((D)\).\([25]\) These relationships are shown in Figure 1. For the same grid notation of the form \(Nxry\), two grids can have different strip heights. In this work, to differentiate grids with different strip heights, the notation, \(Nxryhz\), was used to denote a grid; where \(x\), \(y\), and \(z\) are the respective values for strip frequency in line-pairs cm\(^{-1}\), grid ratio, and strip height in mm.

Image diagnostic quality improvement due to a grid depends on the grid’s constructions being the: grid ratio, strip thickness, strip height, strip, and interspace materials. Grid cover materials are designed to minimize attenuating the radiation. In digital radiography, the \(K_{SNR}\) is a grid performance indicator specifying the image diagnostic quality improvement.

In this work, \(K_{SNR}\) was used to investigate the relationship between strip thickness and grid performance. \(K_{SNR}\) can be expressed by using Equation (1).\([15]\) Setting an extreme condition, if radiation attenuation in the interspace materials (eg, air) is negligible and the strip thickness approaches zero, no radiation is being attenuated by the strips hence \(T_p\) and \(T_t\) would approach unity, and \(K_{SNR}\) will approach unity. In another extreme, if the strip height is enough to fully attenuate the radiation, for example, 3 mm, and the interspace size approaches zero, the grid would be a single piece of strip material; hence \(T_p\) and \(T_t\) approach zero. \(T_t\) is a combination effect of \(T_p\) and \(T_s\) and can be expressed as Equation (2). This equation indicates that \(T_t\) would not go to zero faster than \(T_p\) because \(T_t\) depends on \(T_p\) and the amount of scatter radiation with grid, noting that without a grid, primary and scatter radiations would reach the image receptor without attenuation. Under this extreme, \(K_{SNR}\) will approach zero because \(T_p\) approaches zero faster than the square root of \(T_t\).

\[
K_{SNR} = \frac{T_p}{\sqrt{T_t}}
\]  

(1)

Where \(K_{SNR}\) is the quantum signal-to-noise ratio improvement factor; \(T_p\) and \(T_t\) are the transmissions of primary and total radiation, respectively.

\[
T_t = \frac{T_p \times P_{radiation} + S_{radiation\ with\ grid}}{P_{radiation} + S_{radiation}}
\]  

(2)

Where \(S_{radiation}\) and \(S_{radiation\ with\ grid}\) are the measurements of scatter radiation without and with a grid, respectively; \(P_{radiation}\) is the measurement of primary radiation without a grid.

The \(K_{SNR}\) of contemporary grids as set out in the literature is greater than unity for general grids.\([15, 16]\) Under extreme conditions where strip thicknesses are zero, \(K_{SNR}\) is unity, and where strip thicknesses are the grid size, \(K_{SNR}\) is zero. This consequently leads to a hypothesis that a maximum value of \(K_{SNR}\) exists when the strip thickness changes from zero to the grid size. For a series of grids with a given grid ratio, strip height, strip, and interspace materials, the optimal strip-thickness of these grids can be defined as the thickness at which a grid has the highest \(K_{SNR}\) values.

Given a series of grids described as \(Nxryhz\) with \(x\) as a variable, \(y\) and \(z\) fixed, the optimal strip-thickness can be found at the maximum of \(K_{SNR}\).
3 MATERIALS AND METHODS

3.1 Grid-design details

For this work, the design of a single general grid for high-energy (typically 100 kVp) radiographic applications was chosen. The grid's construction details were selected from Reference [15]. This selected grid had grid ratio 8:1, lead strip thickness 36 μm, grid height 1.7 mm, strip frequency approximately 40 cm⁻¹, carbon fiber interspace distance 214 μm, and 100 cm focal distance. This grid is identified as N4or8h1.7. Modeled on this grid design with the interspace carbon fiber thickness fixed at 214 μm, 72 additional general grid designs of this form, Nxr8h1.7, were able to be created for computer-based simulation in this work. These simulated-grid designs had their strip thicknesses varying stepwise from 6 to 150 μm at 2 μm intervals and were modeled with lead (Pb) strips and carbon interspace materials. Carbon fiber cover for each side was set as 300 μm which would provide necessary protection to the grid and the same time minimize the attenuation of radiation.

3.2 Grid evaluation setups in simulation

The simulation determinations of Tp, Ts, and Tt were in the broad-beam condition without a blocker in accordance with the IEC 60627 Standards.[25] This simulation setup identified forward scatter arising in a narrow-beam condition as well as in blocked primary radiation in a broad-beam condition, hence avoiding the influence of forward scatter on Tp and Ts.[13]

The Monte Carlo code system used in this investigation has been previously validated.[13] This Monte Carlo code system adapted the PENEOPE code[26–28] and kept the physical datasets, for example, the atomic cross-section datasets, Rayleigh scattering, Compton scattering, and photoelectric absorption. The interference effect on Rayleigh scattering[29–31] was added in this Monte Carlo code system. This code system was validated for photon energy approximately from 1 keV to 200 keV under appropriate X-ray imaging settings for the evaluation of grids with overall uncertainties in radiation transmissions within 2.0% between simulations and experiments for tube voltages 80 kVp and above.[13] In this work, the simulation included the interference effect on Rayleigh scattering which has an effect on scatter-to-primary ratio (SPR).[13, 29, 32–35] and Ts.[13, 35, 36] The simulation of radiation transport in the grid materials were performed using the method of Zhou et al.[17] in which the interaction probability of each photon in the grid materials was determined by the path lengths in each type of grid materials. These path lengths were presumably determined as if the photon would traverse the grid materials without an interaction with them.

Evaluation of each simulated-grid-design, Nxr8h1.7, was repeated 10 times to determine mean values and statistical uncertainties. In the simulation, thicknesses of Polymethyl methacrylate (PMMA) material were added as scattering objects. The simulated PMMA had a 30 cm × 30 cm cross-section and the thicknesses were varied from 10 cm to 30 cm in 5 cm increments. The tube voltage selection for the simulation was varied from 60 kVp to 100 kVp in 10 kVp increments as the PMMA thickness increased from 10 cm to 30 cm. X-ray beams were calculated using the model of Poludniowski et al.[37] and were approximately the qualities of RQR 4 to RQR 8 beams.[25] The spectrum of 100 kVp beam is given in Figure 2. The total aluminum (Al) filtration, the photon mean energy, and the first and second Al half-value layer (HVL) of these beams are given in Table 1. The number of photons tracked in each simulation with 30 cm thickness of PMMA was approximately 50 billion.
Figure 2 The spectrum of RQR 8 beam\textsuperscript{[25]}\textsuperscript{[38]} used in the simulation. This X-ray beam spectrum was calculated using the model of Poludniowski et al\textsuperscript{[37]}

<table>
<thead>
<tr>
<th>kVp</th>
<th>Total Al filtration (mm)</th>
<th>Mean photon energy (keV)</th>
<th>first Al HVL (mm)</th>
<th>second Al HVL (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>60</td>
<td>2.5</td>
<td>38.8</td>
<td>2.19</td>
<td>3.22</td>
</tr>
<tr>
<td>70</td>
<td>3.2</td>
<td>40.1</td>
<td>2.56</td>
<td>3.62</td>
</tr>
<tr>
<td>80</td>
<td>3.4</td>
<td>43.9</td>
<td>2.99</td>
<td>4.33</td>
</tr>
<tr>
<td>90</td>
<td>3.6</td>
<td>47.5</td>
<td>3.45</td>
<td>5.10</td>
</tr>
<tr>
<td>100</td>
<td>3.8</td>
<td>50.7</td>
<td>3.93</td>
<td>5.86</td>
</tr>
</tbody>
</table>

Results from previous simulation investigations\textsuperscript{[13, 17]} showed that radiation transmissions in grids were affected by the virtual image receptors stimulated with a physical image receptor without all the construction details. In both the grid evaluation and the determination of SPR, this work necessitated the use of a simulated perfectly efficient detector of energy-integration to avoid any potential bias from the construction of a physical image receptor. In the simulation, the SPRs were determined using the same approach described in section 3.3 for the experimental measurements of SPRs.

3.3 Experimental measurement of scatter-to-primary ratio

Experimental measurements of SPRs were undertaken to validate SPRs obtained in the simulation. A digital radiographic system, DRX Evolution Plus (Carestream Health, Rochester, New York) with a DRX Plus 3543 Gd\textsubscript{2}O\textsubscript{2}S:Tb Scintillator image receptor was used in the experimental setup to evaluate SPRs. X-ray
beam qualities were verified with a Cobia Sense (RTI Group AB, Molndal, Sweden) calibrated in Nov 2018. Additional aluminum filtrations (0.1 mm aluminum sheet) were added appropriately to achieve approximately equivalent first total aluminum half-value-layer (HVL) given in Table 1, except for 60 kVp the tube built-in total filtration was used which produced a beam quality of an equivalent first aluminum HVL of 2.5 mm.

The linearity of the imaging system response to radiation exposures were validated by measuring the mean pixel values with the tube current varied and the tube voltage unchanged. The repeatability of the imaging system response to radiation exposures were validated by repeating the measurements of the mean pixel values with both the tube current and voltage unchanged. The imaging system noise was measured when the image receptor was kept away from the irradiation field while a radiation exposure was engaged.<Query: Reference [38] was not cited in the text. It has been inserted here. Please confirm if this is correct; or delete this insertion and indicate where it should be cited; or delete from the Reference List and renumber the References in the text and Reference List. Ans: delete the citation [38] at this place. No citation is needed at this place. Citation [38] should be used in the Caption of figure 2 to replace citation [25].>>[38]

SPRs were determined using a broad beam with a blocker method modified from the graduated beam stop method used in Fetterly and Schueler[15] and Floyd et al.[39] The lead-blocker was appropriately 2 mm diameter and 5 mm height. Verification of radiation attenuation in the blocker was made without the imaging objects. The region of the blocker’s projection was determined on the image receptor. This region excluded pixels that were underneath the blocker and that might have been exposed to the primary beam due to penumbra effect of the geometry of the finite X-ray focal spot.[40] Scatter radiation was measured within this region and total radiation was measured in an area enclosed by two squares: a 4-mm inner-square and a 6-mm outer-square and centered at the blocker’s projection. SPRs were calculated by using Equation (3).

\[
SPR = \frac{S_{\text{radiation}} - N}{T_{\text{radiation}} - S_{\text{radiation}} - N}
\]  

(3)

Where SPR is the scatter-to-primary ratio; \(S_{\text{radiation}}\) is the measurement of the scatter radiation; \(T_{\text{radiation}}\) is the measurement of the total radiation; \(N\) is the measurement of the imaging-system-noise determined with the image receptor without exposure to the irradiation field.

3.4 Model of the relationship between \(K_{\text{SNR}}\) and strip thickness

The model of the relationship between \(K_{\text{SNR}}\) and strip thickness was determined by the models of Tp and Tt. The \(K_{\text{SNR}}\) model was then used to predict \(K_{\text{SNR}}\) for novel strip thicknesses. In the Monte Carlo simulation, radiation reaching the image receptor was used to calculate Tp, Ts, and Tt for each simulated grid. Models of Tp, Ts, and Tt were determined by regression analysis that followed the exponential attenuation law[41] which uses an effective linear-attenuation-coefficient of \(\mu_{\text{eff}}\), and effective thickness of radiation-attenuation materials, \(x_{\text{eff}}\) [see Equation (4)].

\[
I(x) = I_0 e^{-\mu_{\text{effective}} \times x_{\text{effective}}}
\]  

(4)

Where \(I(x)\) is the intensity of the transmitted beam through the grid materials; \(I_0\) is the intensity of the incident beam on the grid entrance surface; \(\mu_{\text{effective}}\) is the effective linear-attenuation-coefficient and \(x_{\text{effective}}\) is the effective thickness of the grid materials.

In this work, the relationship between strip thickness and Tp, Ts, and Tt was modeled by using Equation (5), a regression with radial basis function networks (RBFNS).[42, 43] which, when it uses the Gaussian function as the radial basis functions, is the sum of multiple Gaussian functions. The model of Ts was not needed in the determination of \(K_{\text{SNR}}\) and was included for references only. In Equation (5), each Gaussian term, which can be solved through the method of Guo.[44] was an exponential attenuation. Thus, for a radiation attenuation
medium, the product of an effective attenuation depth and effective linear attenuation coefficient was modeled as a function of the strip thickness.

\[
f(x) = \sum_{i=1}^{N} a_i \times e^{-\left(\frac{x-b_i}{c_i}\right)^2}
\]  \hspace{1cm} (5)

Where \(f(x)\) is a model of radiation transmissions of grids; \(e = 2.71828\ldots\) being the Euler’s number; \(x\) is the strip thickness variable; \(a_i, b_i, \) and \(c_i\) are coefficients; and \(i\) is the natural number taking values from 1 to \(N\).

The model of \(T_p\) consists of two Gaussian terms, one for modeling the radiation transmission in the interspace materials and another for the radiation transmission in the strips. For \(T_s\) and \(T_t\), radiation traveling in the grid materials had a wide range of entry angles to the surface of the grid. The effects of wide entry-angles on \(T_s\) and \(T_t\) were modeled using seven and eight Gaussian terms, respectively. The fitting code of multiple terms of this Gaussian function were provided in MatLab Ver. R2018b (The MathWorks Inc., Natick, Massachusetts).

Model fittings were iteratively performed to achieve a minimum sum of locally weighted nonlinear least square values, the sum of squared error (SSE), calculated by using Equation (6)—which is also known as the summed square of residuals.[45, 46] The calculation of nonlinear least square was performed using a trust-region-reflective method based on the interior-reflective Newton method described in Coleman and Li.[45, 46]

\[
\text{SSE} = \sum_{i=1}^{N} w_i \times |x_i - y_i|^2
\]  \hspace{1cm} (6)

Where \(\text{SSE}\) is the sum of squared error; \(x_i\) is the value determined in simulation; \(y_i\) is the predicted value from the fit and \(w_i\) is the weighting applied to the data point; and \(i\) is the natural number taking values from 1 to \(N\).

The \(T_p, T_s\) and \(T_t\) models are summarized in the fitting statistics: SSE and the coefficient of determination (R-squared).[47] SSE is a measure of the total deviation from the fitted values to the samples. A value of SSE closer to 0 indicates that the model has a smaller random error component and that the fit will be more useful for prediction. In the iterative fitting, when Equation (6) converged, SSE was calculated by using Equation (6); otherwise, SSE was determined by using Equation (6) at a predefined iterative condition.[45, 46] The R-squared, always between zero and 1, is a measure of how close the sample data are to the fitted model, the closer the R-squared to 1, the better the fitting (the greater proportion of variance is accounted for by the model).

4 RESULTS

4.1 SPR vs radiographic conditions

Table 2 shows the SPRs determined in this investigation. SPRs determined in the simulation were less than SPRs determined in the experiment. The difference of these SPRs at the same imaging condition is smaller at greater SPR and changes from appropriately less than 1% to 18%; the absolute difference of the SPR is from 0.07 to 0.78. The difference of the SPRs could be due to differences in the image receptors as digital image receptors are have higher detection efficiency for lower energy radiations and scatter radiations have lower average energy than the primary radiations.[48] Grids are generally used for large imaging anatomical parts.
The large SPR discrepancy found at 10 and 15 cm PMMA thicknesses should not affect the results for PMMA thicknesses at 20 cm and above.

**Table 2** Scatter-to-primary ratio (SPR) under five radiographic conditions determined by the polymethyl methacrylate (PMMA) thicknesses and peak tube voltage (kVp)

<table>
<thead>
<tr>
<th>Radiographic conditions (PMMA thickness/tube voltage)</th>
<th>SPR (± SE of mean)</th>
</tr>
</thead>
<tbody>
<tr>
<td>10 cm/60 kVp</td>
<td>2.31±0.01</td>
</tr>
<tr>
<td>15 cm/70 kVp</td>
<td>4.34±0.01</td>
</tr>
<tr>
<td>20 cm/80 kVp</td>
<td>7.22±0.01</td>
</tr>
<tr>
<td>25 cm/90 kVp</td>
<td>11.00±0.01</td>
</tr>
<tr>
<td>30 cm/100 kVp</td>
<td>15.76±0.02</td>
</tr>
</tbody>
</table>

SPRs obtained through this work simulation are average 11% greater than SPRs determined in the simulation of Zhou et al.[13] SPRs experimentally measured in this work are average 15% greater than SPRs that was experimentally determined in Zhou et al.[13] The difference in SPRs between this work simulation and the simulation of Zhou et al.[13] is caused by the difference in the radiation detectors. The difference in the SPRs between this work experiment and the experiment of Zhou et al[13] is likely due to the combination of the radiation detectors, the imaging-system-noise, and the difference in the radiation blocker sizes.

**4.2 Tp, Ts, and Tt models of grid designs**

The models of Tp, Ts, and Tt determined by using Equation (5) were shown in Figures 3–5, respectively. For all the models, the fitting statistics of the SSE values are less than $1 \times 10^{-4}$ and R-squared values are >0.9995.
five imaging conditions given by the tube peak-kilo-voltage (kVp) and the polymethyl methacrylate (PMMA) thicknesses.

**Figure 4** The transmission of scatter radiation (Ts) vs grid strip thickness (d) for five imaging conditions given by the tube peak-kilo-voltage (kVp) and the polymethyl methacrylate (PMMA) thicknesses. The Ts of a series of grids of the form Nxr8h1.7 decreases as the strip thickness increases for all the five imaging conditions, where N is the strip frequency and x is the value; r is the grid ratio and 8:1 is the ratio; h is the strip height in mm and 1.7 is the value.

**Figure 5** A plot of primary radiation transmission (Tt) vs strip thickness (d) for five imaging conditions given by the tube peak-kilo-voltage (kVp) and the polymethyl methacrylate (PMMA) thicknesses. The Tt of a series of grids of the form Nxr8h1.7 decreases as the strip thickness increases, where N is the strip frequency and x is the value; r is the grid ratio and 8:1 is the ratio; h is the strip height in mm and 1.7 is the value.
As shown in Figure 3, the Tp exponentially decreases from ~0.9 to ~0.5 in a very similar way for all the radiographic conditions as the strip thickness increases from 6 to 150 μm for the simulated grid designs of the form Nxr8h1.7. When the simulated kVp and PMMA thickness changed, at the same strip thickness, the Tp is higher when the kVp is higher due to the higher average photon energies. However, the relative difference of Tp for the various kVp and PMMA thickness settings at the same strip thickness is less than 1.2% and the absolute difference of Tp is less than 0.009.

Figure 4 shows that the Ts, in all simulated radiographic conditions, decreases exponentially from ~0.6 to ~0.07 as the simulated-strip-thickness increases from 6 to 150 μm. Again, at a constant simulated strip thickness, the Ts is larger in a simulated thicker PMMA material and higher kVp due to the higher average-scatter-photon-energies. The range of the absolute difference in Ts is from 0.019 to 0.177.

Because of the combination effect of the Tp, Ts, and SPR, the Tt decreases from appropriately 0.6 to 0.12 as the strip thickness increases from 6 to 150 μm (Figure 5). The decreasing rate of Tt is faster when strip thickness is thinner, and the decreasing rate gradually slows as the strip thickness increases. Unlike for the variation of Tp and Ts with strip thickness, the variations of Tt values are not the same for all the kVp and PMMA-thickness conditions.

The coefficients of the Tp, Ts, and Tt models expressed in Equation (5) are available from the authors upon request.

4.3 The optimal strip-thickness for Nxr8h1.7 designs

The $K_{SNR}$ of Nxr8h1.7 grid designs for the five simulated kVp and PMMA-thickness conditions were calculated by using Equation (1) and the values of Tp and Tt were determined from their respective models. Figure 6 shows the values of the $K_{SNR}$ at these conditions. As the strip thickness changes from 6 to 150 μm, the $K_{SNR}$ first increases to the maximum and then decreases. Figure 6 also shows the maximum values (Table 3) of the $K_{SNR}$ for each of these conditions.

![Figure 6](image-url)  
**Figure 6** The signal-to-noise ratio (SNR) improvement factor ($K_{SNR}$) of a series of grids of the form Nxr8h1.7 as a function of strip thickness (d); where N is strip frequency and x is the value; r is the grid ratio and 8:1 is the ratio; h is the strip height in mm and 1.7 is the value. For all the five imaging conditions given by the tube peak-kilo-voltage (kVp) and the Polymethyl methacrylate (PMMA) thicknesses, the
$K_{\text{SNR}}$ increases and then decrease as the strip thickness increases from 6 to 150 μm. The maximum $K_{\text{SNR}}$ (asterisk) exists for each of the imaging conditions.
<table>
<thead>
<tr>
<th>Tube voltage/PMMA thickness</th>
<th>60 kVp/10 cm</th>
<th>70 kVp/15 cm</th>
<th>80 kVp/20 cm</th>
<th>90 kVp/25 cm</th>
<th>100 kVp/30 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Optimal strip thickness (μm)</td>
<td>27.5</td>
<td>41.2</td>
<td>54.6</td>
<td>67.2</td>
<td>74.2</td>
</tr>
<tr>
<td>$T_p$</td>
<td>0.828</td>
<td>0.787</td>
<td>0.749</td>
<td>0.717</td>
<td>0.701</td>
</tr>
<tr>
<td>$T_s$</td>
<td>0.159</td>
<td>0.146</td>
<td>0.141</td>
<td>0.138</td>
<td>0.144</td>
</tr>
<tr>
<td>$T_t$</td>
<td>0.362</td>
<td>0.266</td>
<td>0.215</td>
<td>0.186</td>
<td>0.177</td>
</tr>
<tr>
<td>$K_{SNR}$</td>
<td>1.377</td>
<td>1.526</td>
<td>1.615</td>
<td>1.660</td>
<td>1.667</td>
</tr>
</tbody>
</table>
The optimal strip-thicknesses, for Tp, Ts, Tt, and \( K_{\text{SNR}} \) at these thicknesses are presented in Table 3. The optimal strip-thicknesses increase from approximately 27.5 to 74.2 \( \mu \)m as the kVp and PMMA thickness conditions change from 60 kVp and 10 cm to 100 kVp and 30 cm. The changes in Tp, Ts, and Tt are likely due to the combination effect of the increase in the strip thickness, the kVp, and the PMMA thickness. The Ts varies from approximately 0.138 to 0.159 with the minimum at 90 kVp and 25 cm PMMA thickness. For the five kVp and PMMA thickness conditions, the \( K_{\text{SNR}} \) increases from approximately 1.377 to 1.667 due to the combination effect of Tp and Tt.

5 DISCUSSION

The N40r8h1.7 grid, experimentally evaluated by Fetterly and Schueler,[15] is used for high-energy radiographic applications. At tube voltage 100 kVp and 30 cm PMMA thickness, Tp determined in this work simulation is the same as Tp obtained in the simulation of Zhou et al[13] which used water as the imaging materials. Ts determined in this work is 10% less than Tp determined in Zhou et al.[13] This suggests the difference in PMMA materials and water have negligible effects on Tp, but significant effects on Ts due to difference in the materials’ average atomic numbers, the higher the average atomic number, the greater the interaction probabilities between the photons and the mass.

The theoretical Tp and Ts of the N40r8h1.7 grid determined in this work are approximately 6% greater than Tp, and 4% less than Ts experimentally measured in Fetterly and Schueler,[15] respectively. The difference between measured Tp and the grid’s theoretical Tp has been previously noted[15] and could be due to manufacture precision. The difference of Ts between theoretical evaluation and experimental measurement can be due to the combination of manufacture precision and the difference of the imaging materials.

The optimal thickness of strips depends on the kVp and imaging material thicknesses (Figure 6 and Table 3). For radiographic conditions at 100 kVp and 30 cm thickness PMMA of 30 cm \( \times \) 30 cm cross-section, the optimal strip thickness of the N\text{xr}8h1.7 grids is approximately 74 \( \mu \)m, which corresponds to approximately 35 strip-lines cm\(^{-1}\), hence a N35r8h1.7 grid. The N35r8h1.7 grid has strip thickness which is approximately two times the strip thickness of the N40r8h1.7 grid. The theoretical performances of the N35r8h1.7 grid in terms of \( K_{\text{SNR}} \), Ts, and Tp are approximately 5.2% greater than, 36% and 13% less than the \( K_{\text{SNR}} \), Ts, and Tp of the N40r8h1.7 grid’s theoretical performance, respectively. The changes in optimal strip-thicknesses at each PMMA thickness indicate that one grid cannot suit all imaging conditions. A better approach might be to have multiple grids covering a range of imaging conditions and select the most appropriate one for each individual patient.

The theoretical \( K_{\text{SNR}} \) of N35r8h1.7 grid at 20 cm PMMA thickness and 80 kVp is 15% greater than the experimentally measured \( K_{\text{SNR}} \) of the N40r8h1.7 grid at 20 cm water thickness and 104 kVp, or at 30 cm PMMA thickness and 100 kVp is 9% greater than the measured \( K_{\text{SNR}} \) of the N40r8h1.7 grid at 50 cm water thickness and 104 kVp. This suggests that the N35r8h1.7 grid would significantly outperform the N40r8h1.7 grid. The differences, however, are also affected by the difference in kVp, imaging materials, and thicknesses.

Compared with contemporary commercial grids with strip-frequencies from 40 to 80 lines cm\(^{-1}\), the N35r8h1.7 grid has a low strip frequency. When this grid would be used in stationary with digital image receptors, grid-line-removing techniques[3, 5–9] might be an optional solution to remove interference of grid-line artifacts on making diagnosis.

The finding of optimal strip-thickness is not only about the improvement of image diagnostic quality, also about radiation protection which should not be overlooked all the time in medical X-ray imaging. The linear-no-threshold (LNT) model for radiation exposure induced cancer risks is used to guide the practice and the management of radiation protection in medical diagnostic X-ray imaging.[49–51] An improvement of \( K_{\text{SNR}} \) can mean either improving the image diagnostic quality, improving radiation protection, or both. The LNT model indicates a small amount of radiation exposure reduction to each patient is going to be a significantly
large benefit to the health outcomes of the public. This suggests that the improvement in $K_{SNR}$ can lead to a huge benefit to the public health outcomes for the reduction of radiation-exposure-induced-cancer risks.

6 CONCLUSIONS

In conclusion, for a given grid-ratio, strip-height, and strip and interspace materials, the optimal strip-thickness exists and can be determined at the maximum $K_{SNR}$ for a given radiographic condition. Optimal strip-thicknesses change at each imaging condition. An imaging system can be equipped with multiple grids optimized for different imaging conditions and the most appropriate grid can then be selected for each individual patient.

REFERENCES


