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FULL PAPER

Investigating the use of an antiscatter grid in chest radiography for average adults with a computed radiography imaging system

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Objective: The aim of this study was to investigate via simulation a proposed change to clinical practice for chest radiography. The validity of using a scatter rejection grid across the diagnostic energy range (60–125 kVp), in conjunction with appropriate tube current–time product (mAs) for imaging with a computed radiography (CR) system was investigated.

Methods: A digitally reconstructed radiograph algorithm was used, which was capable of simulating CR chest radiographs with various tube voltages, receptor doses and scatter rejection methods. Four experienced image evaluators graded images with a grid ($n=80$) at tube voltages across the diagnostic energy range and varying detector air kermas. These were scored against corresponding images reconstructed without a grid, as per current clinical protocol.

Results: For all patients, diagnostic image quality improved with the use of a grid, without the need to increase tube mAs (and therefore patient dose), irrespective of the tube voltage used. Increasing tube mAs by an amount determined by the Bucky factor made little difference to image quality.

Conclusion: A virtual clinical trial has been performed with simulated chest CR images. Results indicate that the use of a grid improves diagnostic image quality for average adults, without the need to increase tube mAs, even at low tube voltages.

Advances in knowledge: Validated with images containing realistic anatomical noise, it is possible to improve image quality by utilizing grids for chest radiography with CR systems without increasing patient exposure. Increasing tube mAs by an amount determined by the Bucky factor is not justified.

Radiography of the chest is one of the most frequently performed diagnostic radiographic examinations in the UK. In 2010, the Health Protection Agency (now Public Health England) reported¹ that chest radiographs represented 19.6% of all radiographic examinations in 2008 (although the contribution to collective dose was small at about 0.5%), so optimization of radiation dose (*i.e.* ensuring dose is as low as reasonably practicable) and image quality (*i.e.* ensuring all required clinical structures are visible to the reporting healthcare professional so that an acceptable diagnosis is possible) in chest radiography is an important research area, especially since digital imaging has all but replaced its film-screen counterpart. It is also a legal requirement in the UK under the Ionising Radiation (Medical Exposure) Regulations 2000² to optimize all medical exposures, consistent with the intended purpose.

One such technique to optimize image quality in chest radiography is to use a scatter rejection grid. They work by preferentially removing radiation scattered by the body prior to reaching the detector, and their improvement of image quality in film-screen imaging has been recognized for decades.^{3–6} This improvement was described by the contrast improvement factor⁷ but came with a cost. Film requires a given level of incident exposure to ensure adequate optical density (OD), and because a grid attenuates most of the scattered radiation (as well as some primary radiation), this necessitates an increase in tube current–time product (mAs). The Bucky factor describes the necessary multiplication by which exposure parameters must be increased, and for film-screen can be anything between 2–6 times the “non-gridded” exposure.^{8,9} Regardless of this, there are clear guidelines

recommending the use of scatter rejection grids with film-screen systems for adult chest radiography,¹⁰ but none for digital imaging modalities, although Fritz and Jones¹¹ have recently published guidelines for scatter rejection techniques in paediatric digital radiology.

Digital image detectors, such as computed radiography (CR) photostimulable powder phosphors, have a larger dynamic range than does film,¹² and grey levels in the resulting image are usually adjusted, irrespective of incident detector dose, to match the output of the display monitor. Therefore, unlike film, digital imaging is not contrast (OD) limited. At doses used clinically (e.g. air kerma of approximately 2–15 μGy at the receptor), digital images are dominated by quantum noise (*i.e.* other noise sources such as electronic and structural are typically $\leq 2\%$ of the total noise as consistently demonstrated through in-house routine quality assurance testing of this CR system), which depends on the level of air kerma incident on the detector (signal) and the detector's detective quantum efficiency. Therefore, when a scatter rejection grid is used, appropriate exposure factors (tube's peak kilo-voltage and/or mAs) are required to maintain a level of image signal-to-noise ratio (SNR)^{13,14} acceptable to the image evaluator, although there is no agreement as to what these appropriate exposure factors should be. A common school of thought, described in a considerable resource by Carlton and Adler,¹⁵ suggests that the lower limit on the increase in mAs should be the reciprocal of the primary transmission (T_p) of the grid and the upper limit, the Bucky factor. The reciprocal of T_p for modern grids is typically 1.2–1.4, which suggests that an increase in mAs of at least 20% is required. However, Tanaka et al¹⁶ have recently demonstrated that the use of grids (grid ratios 5:1 to 14:1) without increasing exposure factors (compared with "non-gridded" exposures), actually improved the effective noise equivalent quanta (eNEQ) when acquiring images of 20 cm of polymethylmethacrylate (PMMA). They concluded that the improvement to image quality owing to removal of scatter outweighs the increase in quantum noise when a grid is used, although they acknowledged that their work did not use any images with anatomical structure. Similarly, Fetterly and Schueler¹⁷ studied numerous grids with different thicknesses of uniform solid water and suggested that a scatter rejection grid can provide improvement in SNR without increasing exposure for large patients.

Given the scarcity of evidence/guidelines in the literature, the aim of this study was two-fold: firstly, to investigate the validity of using a scatter rejection grid for chest radiography of average adults with an Agfa CR imaging system (Agfa, Peissenberg, Germany) across the diagnostic energy range (60–125 kVp); and secondly, to investigate appropriate tube mAs to identify what increase in patient dose (if any) there needs to be. Many recent studies have demonstrated that anatomical noise (the influence of projected anatomy on the image evaluator's ability to detect potential abnormalities and provide an accurate diagnosis) is the limiting factor in chest radiography,^{18–27} so it was felt that the inclusion of realistic anatomy in the images used in this study was of particular importance, rather than using uniform PMMA or solid water only. Therefore, computer-simulated chest radiographs (each containing realistic projected anatomical noise and

lung abnormalities/nodules), reconstructed with a scatter rejection grid, were compared by expert image evaluators with images reconstructed without a grid (as per current clinical protocol in our radiology department).

METHODS AND MATERIALS

Simulated chest images

A computer algorithm capable of simulating CR chest images has been developed and validated by our group²⁸ and has been used in recent research applications to derive optimum exposure parameters for chest CR imaging.^{29–31} The algorithm can simulate diagnostic tube voltages, varying detector air kerma (DAK) values and scatter rejection techniques by simulation of a conventional two-dimensional radiograph created from CT data [a digitally reconstructed radiograph (DRR)]. Artificial lesions ("nodules") can also be added to the DRRs to simulate malignant and non-malignant disease. A brief synopsis of the computer model and how the images are produced is described below:

- (1) The virtual phantom is derived from the chest portion of real patient CT data sets. The voxel resolution of the phantom is $0.34 \times 0.80 \times 0.34$ mm (width $\times$ height $\times$ depth).
- (2) CT number is converted into linear attenuation coefficient using formulae derived from the Gammex RMI[®] (Broadway Business Centre, Nottingham, UK) tissue-equivalent phantom (model no. 467). This is a solid water cylinder that contains 17 inserts, the attenuation properties of which mimic the range of attenuations of the various tissues found *in vivo*.
- (3) X-ray spectra are generated using the spectrum processor prepared by Sutton and Reilly in IPEM report 78.³²
- (4) X-ray pencil beams are projected through the CT data set using a ray-casting method of DRR production. X-rays are attenuated as they move through the CT data in an exponential manner. The intensity of photons at each energy emerging from the virtual phantom is calculated.
- (5) Energy absorbed in the virtual CR phosphor is then calculated and converted to CR pixel value. This is the raw DRR.
- (6) Frequency-dependent noise (using images acquired on the clinical CR system) is added to the raw DRR using a slightly adapted method described by Bth et al.³³
- (7) Scatter measured physically ("scatter masks") on the clinical CR system is added to the raw DRR. The following separate scatter masks were acquired:
 - (i) *With no clinical scatter rejection (i.e. non-grid).* The ratio of scattered radiation absorbed in the CR phosphor to that of the total radiation at 60 kVp [the scatter factor (SF)] ranged from 0.33 to 0.47 and 0.66 to 0.85 in the lung and spine/diaphragm regions, respectively. SFs at 150 kVp ranged from 0.39 to 0.53 in the lung and from 0.69 to 0.88 in the spine/diaphragm, respectively. These values are in general agreement with SFs measured by Floyd et al³⁴ in humans. There is little change in SF in the spine/diaphragm regions with a change in tube voltage, but the effect is slightly more pronounced in the lung. This is similar to that reported by Bonenkamp and Boldingh.³
 - (ii) *With an antiscatter grid (strips per millimetre = 4; grid ratio = 12).* SFs were (on average) 40% lower in the lungs

(i.e. SF = 0.13–0.19) and 48% lower in the spine/diaphragm (i.e. SF = 0.26–0.34) regions when compared with SFs derived without scatter rejection (non-grid). This correlates with the scatter transmission factor of 0.14 derived using a 10-cm thick solid water phantom reported by Fetterly and Schueler.¹⁷

As described in our previous work,²⁸ the resulting DRR images were validated quantitatively with real-patient CR images (local ethics approval was obtained to allow the use of patient data), using SNR and histogram measurements. SNR values measured in the DRRs were always within 15% of the corresponding real images, and the dynamic ranges of the DRR histograms (minimum and maximum pixel values) were within two standard deviations (SDs) of the corresponding values in the CR images. Qualitative validation was carried out by expert image evaluators, and they all agreed that the DRRs adequately simulated real CR images and that they were acceptable to use for optimization studies. Each of the evaluators was asked “Do the images adequately mimic real CR images (1 = definitely not, 10 = definitely)?” The mean (± 1 SD) score was 7.5 ± 2.0 demonstrating the algorithm’s capability of producing chest radiographs. However, each evaluator did comment that DRR image spatial resolution was poorer than CR, but they agreed that this was not a limitation, and that optimization studies would be possible using these images.

To assess the clinical validity of using scatter rejection grids (Philips Medical Systems, Guildford, UK) of a typical specification for chest imaging with this CR system, as well as appropriate exposure factors, simulated images were reconstructed with an oscillating scatter rejection grid (strips per millimetre = 4; grid ratio = 12; lead absorbing material; primary transmission $T_p = 0.60$) across the diagnostic energy range with various DAKs through the lung region (the “test images”), and these were scored against the corresponding “non-gridded” image (the “reference image”) reconstructed with exposure factors in line with our radiology department’s current clinical protocol. It should be noted that in this manuscript, DAK always refers to air kerma reaching the CR receptor (i.e. if a grid is used in a simulated image, DAK refers to the air kerma reaching the receptor behind the grid). Figure 1 shows simulated images of a chest reconstructed with and without a scatter rejection grid, respectively. Figure 1a clearly demonstrates improved quality and contrast in the dense regions, such as the spine, heart and diaphragm, compared with Figure 1b.

Images of a given simulated patient at a given tube voltage were attached in series to a single study and given the name Grid_Patient_N, where N was the sequential simulated patient number. For example, Patient 1 contained eight images, all simulated at 60 kVp from a given real patient CT data set, where images 1–7 were designated as test images and reconstructed with DAKs of 1.0, 2.2, 2.8, 3.4, 5.6, 16.8 and 22.4 μGy , respectively; all with a scatter rejection grid. Image 8 was designated the reference image and was reconstructed with a DAK of 5.6 μGy without a scatter rejection grid. This patient was called Grid_Patient_1 and sent to the hospital’s picture archiving and communication system (PACS) for scoring by expert image

evaluators. A DAK at the CR receptor of 5.6 μGy was chosen for the reference image because this matches the average DAK at the CR receptor through the lung region of real clinical non-gridded images acquired with our radiology department’s clinical protocol (60 kVp, 8 mAs for average adults). The DAKs of the test images at the CR receptor behind the grid were chosen based on the primary transmission of the scatter

Figure 1. Simulated chest images of an average-sized patient reconstructed with an antiscatter grid (a), and without a grid (b).

(a)

(b)

rejection grid ($T_p = 0.6$); $3.4 \mu\text{Gy}$ is 60% of $5.6 \mu\text{Gy}$, and therefore it was possible to reconstruct this particular gridded image with the same exposure factors (60 kVp and 8 mAs) as that of the non-gridded reference image. DAKs $< 3.4 \mu\text{Gy}$ were chosen to assess the effect of lower exposure mAs (compared with the reference) on image quality. The three higher DAKs (5.6, 16.8 and $22.4 \mu\text{Gy}$) were chosen to reflect the traditional view that exposure mAs should be increased by at least the reciprocal of the primary transmission of the grid (in this case, to ensure that a DAK of $5.6 \mu\text{Gy}$ is still registered through the grid) and up to the Bucky factor (3–4 times the non-gridded exposure).

Simulated Patient 2 contained eight images simulated at 81 kVp (all reconstructed from the same real CT data set as Patient 1), where images 1–8 (1–7 gridded, 8 non-gridded) were reconstructed with the same DAKs as simulated Patient 1; this patient was called Grid_Patient_2. Simulated Patients 3 and 4 consisted entirely of 102 and 125 kVp images, respectively, simulated from the same real CT data set as simulated Patients 1 and 2, and with the same DAKs. These simulated patients were named Grid_Patient_3 and Grid_Patient_4, respectively. The process was repeated, with simulated Patient 5 made up of 60 kVp images, simulated Patient 6 made of 81 kVp images, simulated Patient 7 made of 102 kVp images and simulated Patient 8 made of 125 kVp images (simulated Patients 5–8 all produced from a given real CT data set but different to the one used for simulated Patients 1–4), until a total of 80 simulated patients were produced using a total of 20 real CT data sets for the simulations (*i.e.* 20 per tube voltage). It should be noted that all simulated patients used in this study were reconstructed from real CT data sets of average-sized male and female patients (patient width $33 \pm 2 \text{ cm}$) based on the advice of the expert scanning radiographer.

Simulated image quality (visual grading analysis score) scoring

Four experienced image evaluators (two radiologists and two reporting radiographers) graded the simulated images on reporting PACS workstations with a dual monitor configuration (Barco Ltd, Brussels, Belgium). The monitors were calibrated to national standards³⁵ and were kept in dedicated viewing rooms with lighting levels maintained at an acceptable level. The evaluators were asked to keep image 8 of each patient on the right-hand screen (the reference image for grading). All other images (test images) were displayed at random on the left-hand screen and graded against the reference image. None of the evaluators had any knowledge of what tube voltage or DAK, or if a grid had been used in any of the test or reference images. Evaluators were allowed to change the window and level settings of each image prior to grading to optimize the appearance of each, as per clinical practice.

Image scoring criteria were based on the Council of European Communities Quality Criteria,¹⁰ slightly revised to reflect modern diagnostic requirements and experiences of our group and others.^{36–38} We also asked the image evaluators to score the quality of simulated lung nodules/abnormalities, given these are important markers for both malignant and non-malignant chest disease. Image criteria can be seen in Table 1.

Negative and positive scores indicated inferior and superior image quality of the test images compared with the reference. A visual grading analysis score (VGAS) was calculated using the equation described by Tingberg and Sjöström:³⁷

$$\text{VGAS} = \frac{\sum_{i=1}^I \sum_{s=1}^S \sum_{o=1}^O G_{i,s,o}}{I \times S \times O} \quad (1)$$

where $G_{i,s,o}$ is the grading ($-3, -2, -1, 0, +1, +2, +3$) given by the observer o for the image i and the image criteria s ; I is the number of images; S is the total number of image criteria scored; and O is the number of evaluators. Table 2 demonstrates the scoring system employed; for example, if the “quality of the lung region” is “definitely inferior to the reference image”, then this structure should be given a score of -3 and so on. As well as grading the images, to gauge a qualitative measure of image quality, evaluators were asked to answer yes/no to the following question: “is the overall level of noise acceptable for clinical diagnosis?”

Statistical analysis

Interobserver variability was tested for statistical significance using the analysis of variance (ANOVA) test. A p -value of < 0.05 was considered as a statistically significant difference between the data sets. Error bars shown in the graphical data of the results were calculated by averaging the SDs of each VGAS for each chest structure (Table 1) over all patients.

Real radiographs of a physical phantom

In the event that the results demonstrated superior image quality with a scatter rejection grid for equal (or lower) exposure mAs compared with the non-gridded technique, real radiographs of a physical phantom were used to assess the image quality implications prior to issuing recommendations to our radiology department. As such, the chest portion of the Alderson RANDO[®] anthropomorphic phantom (The Phantom Laboratory, Salem, NY) consisting of a natural human skeleton embedded in a synthetic isocyanate rubber with lung substitute and air cavities, simulating the average male (approximately 73 kg) was set up on a Philips X-ray system (Philips Medical Systems) as per local clinical chest protocol. Although primarily used for radiation therapy, RANDO has been shown to attenuate diagnostic energy radiation similar to that of water, which in turn has very similar properties to that of human muscle.³⁹ The phantom was

Table 1. The image criteria used for visual grading analysis

Image criteria
Overall quality of image compared with reference
Quality of lung region
Quality of hilar/mediastinum
Quality of spine
Quality of ribs
Quality of diaphragm/retrodiaphragmatic
Quality of lung nodules/abnormalities if visible in image

Table 2. The grading system for visual grading analysis score

Grading	Visibility of structure
-3	Definitely inferior to the reference image
-2	Reasonably inferior to the reference image
-1	Slightly inferior to the reference image
0	Equal to the reference image
+1	Slightly better than the reference image
+2	Reasonably better than the reference image
+3	Definitely better than the reference image

exposed (60 kVp, 8 mAs) with and without the use of the scatter rejection grid. After each acquisition, the cassette was digitized in the CR reader with a fixed sensitivity of 400 (an Agfa specific pre-processing setting), with no clinical post-processing applied.

RESULTS

Simulated image quality (visual grading analysis score) scoring

For all images evaluated in this study, the ANOVA test demonstrated a *p*-value of 0.34, and therefore, it was deemed there was no significant difference between the scores of each image evaluator.

The results of the non-grid vs grid technique are shown graphically in Figure 2. It is clear from Figure 2 that the VGAS is higher for images reconstructed with a grid for all DAKs $>1 \mu\text{Gy}$, irrespective of the tube voltage. This suggests that the image quality is superior with the use of a scatter rejection grid even when using the same (or decreased) exposure mAs as the non-gridded technique, even for low tube voltage techniques (*i.e.* 60 kVp) such as that used in our radiology department. VGAS scores ranged from -0.81 for $1.00 \mu\text{Gy}$ to $+0.89$ for $22.4 \mu\text{Gy}$. It

is clear that increasing the exposure mAs by the Bucky factor (as required with film-screen) makes very little difference to the image quality, given that the VGAS at $22.4 \mu\text{Gy}$ is only 0.05 points higher than that at $5.6 \mu\text{Gy}$.

Table 3 shows the response of the evaluators when asked if the overall level of noise in each image was acceptable. These responses not only clearly demonstrate that images reconstructed with $1 \mu\text{Gy}$ are not good enough (as per the VGAS results), but also indicate that a significant minority of the 2.2 and $2.8 \mu\text{Gy}$ (17.5% and 8%, respectively) are also too “noisy” for clinical diagnosis (even though VGAS results demonstrate superior image quality at these DAKs). It is likely that the number of rejects resulting from this fraction of non-diagnostic images would be too high for a radiology department to tolerate. Images reconstructed with DAKs $\geq 3.4 \mu\text{Gy}$ were all deemed acceptable in terms of image noise.

Real radiographs of a physical phantom

Figure 3a,b show real radiographs of the RANDO phantom acquired with and without a scatter rejection grid, respectively, both with the same exposure factors (60 kVp, 8 mAs). It is clear that the image acquired with a grid demonstrates superior detail in the dense regions of the phantom, such as the diaphragm and spine regions, without loss of detail in the lungs. As per the results discussed in simulated image quality (VGAS) scoring section, this raises the possibility that chest images may be acquired with a scatter rejection grid on this CR system with the same exposure factors as would have been used without the grid.

DISCUSSION

In the UK, the prime role of the medical physics expert (MPE) in radiological imaging is to ensure optimization of patient radiation exposure and image quality (although there must be input from other members of the multidisciplinary team, such as radiologists and radiographers). The results of this study will

Figure 2. Image quality [visual grading analysis score (VGAS)] results for simulated images reconstructed with an antiscatter grid at each tube voltage investigated and increasing levels of detector air kerma.

Table 3. Response of image evaluators when asked if overall level of noise in the images was acceptable or not

Image detector air kerma (μGy)	Evaluators' response
1	0% of images acceptable
2.2	82.5% of images acceptable
2.8	92% of images acceptable
3.4	100% of images acceptable
5.6	100% of images acceptable
16.8	100% of images acceptable
22.4	100% of images acceptable

be useful to the MPE charged with the task of optimization, given they indicate it is possible to use a scatter rejection grid without having to increase exposure (and therefore patient dose), irrespective of the tube voltage used. It is also clear from the results that increasing dose to the level required for film-screen (*i.e.* an increase in proportion with the Bucky factor) is not required. This has been demonstrated through expert grading of computer-simulated images as well as basic visual inspection of real radiographs of the RANDO phantom. The results presented in this manuscript, which uses images containing realistic anatomical noise and lung abnormalities/nodules, are also consistent with those of Tanaka *et al*¹⁶ and Fetterly and Schueler,¹⁷ even though they acquired images of uniform phantoms only.

Based on the results of a previous optimization study,²⁹ in 2012, our radiology department adopted a standard operating protocol for average chest exposures of 60 kVp, 8 mAs. This represented a reduction from 85 kVp, 5 mAs, as the results demonstrated that the image quality was superior at low tube voltages for a given effective dose (and therefore risk). In the 3 years that the new exposure protocol has been in place, there have been no complaints about image quality. Although this is a relatively low tube voltage, annual large-scale patient dose audits, following the method by Wood *et al*,⁴⁰ have consistently shown that median dose–area product for standard-sized patients are no higher than the national diagnostic reference level of 0.12 Gy cm^2 for posteroanterior chest exposures. Attempts at reducing mAs further gave images that were deemed too noisy, so it may be considered that our patient doses are at or near an optimum level. This is important in the context of this study, given that the reference images used were all simulated as per clinical protocol, that is, simulated gridded images were benchmarked against optimized non-gridded images. Based on the conclusive evidence presented in this article, chest images of six patients were acquired using a scatter rejection grid without increasing patient exposure from the current protocol of 60 kVp, 8 mAs (these gridded exposures were therefore “dose neutral”). Figure 4 shows the resulting image Figure 4a, as well as a radiograph acquired 2 years previously without a grid Figure 4b. It is clear that Figure 4a demonstrates improved visibility in the denser regions (diaphragm, retrocardiac and spine), as well as increased contrast in the lung. These findings were consistent

with the other five patients' images. None of the radiographs needed repeating.

Owing to use of a grid, DAKs through all chest regions are lower than their non-gridded counterpart (by a factor equal to the T_p), which results in lower SNR. This is evident in all of the images acquired with a grid (they look “noisier” in the dense regions). Nevertheless, this work has shown that improved visibility of structures in the dense areas of the chest clearly outweighs the decrease in SNR (and therefore increase in perceived noise). This backs up the argument of Tanaka *et al*¹⁶ in that an increase in normalized noise power does not outweigh the increase in eNEQ.

It is acknowledged that some hospitals still use high tube voltage techniques (even without a grid) primarily to (1) minimize the

Figure 3. RANDO® phantom (The Phantom Laboratory, Salem, NY) chest images acquired with an antiscatter grid (a), and without a grid (b). Both exposures were carried out with the same exposure factors (60 kVp, 8 mAs).

(a)

(b)

Figure 4. Real patient chest images acquired with an anti-scatter grid (a), and without a grid (b). Both exposures were carried out with the same exposure factors (60 kVp, 8 mAs).

(a)

(b)

appearance of ribs and (2) to minimize patient entrance dose, but the natural conclusion of this work is that chest CR imaging can be performed at low tube voltages with a grid. Optimal tube voltage is dependent on the anatomical region of interest—a difficulty that will only be overcome with the implementation of dual-energy imaging. However, using a scatter rejection grid helps overcome this problem to an extent by increasing the visibility of structures in the denser chest regions. Nevertheless, one must not forget that to maintain an appropriate DAK at low tube voltages, patient entrance air kerma (*i.e.* mAs, assuming all other factors remain constant) must be increased compared with higher tube voltage techniques. This notwithstanding, the highest entrance air kerma encountered with chest radiography (typical values <0.1 mGy at 60 kVp with our current clinical protocol) will always be submilligray, so it should be effective dose (and

therefore risk) and not entrance dose that is the limiting factor in determining optimum tube voltage for chest radiography with digital imaging systems, even when using a scatter rejection grid. Furthermore, in the era of film-screen radiography, high tube voltages were required to overcome the restricted dynamic range of film, but this is no longer an issue with digital detectors. Owing to the results presented in this manuscript and in our previous work,²⁹ the use of low tube voltages with antiscatter grids for chest radiography is clearly justified.

It should be remembered, given the relatively large errors in the VGAS owing to the relatively limited simulated patient information set ($n = 80$), the acquired level of change one may expect to see in clinical image quality when using grids, for a given change in computer-simulated image quality, may not always be observed in reality. The results reported here must therefore be interpreted in the context of this limitation. Similarly, only one grid type has been investigated, so one must be careful in transferring the results derived here to grids with different specifications.

It should be remembered that the findings in this article are specific to Agfa CR receptor technology. However, given that most powder-based CR phosphors are based on the barium fluorohalide family, it is likely the conclusions of this work are transferable to other manufacturers, but further studies would be worthwhile. Furthermore, clinical post-processing used by this specific CR manufacturer was not used in any of the simulated images, possibly limiting the value of this work to this vendor only (however, it should be noted that real patient images acquired with a grid were subject to the manufacturer's clinical post-processing).

As discussed in previous work,²⁸ the DRR algorithm used in this study includes the noise properties of the CR system; uniform noise images are taken from the CR system at a receptor dose matching that through the lung region (reaching the CR receptor) of the DRR. For example, if a DRR is reconstructed with $5.6 \mu\text{Gy}$ reaching the receptor through the lung, a uniform image of $5.6 \mu\text{Gy}$ is acquired on the Agfa CR system and added to the DRR. However, the magnitude of noise is “corrected” in the denser regions of the chest using a square root relationship only (*i.e.* only quantum noise is addressed). This is a limitation of the simulation tool given that this may not adequately simulate electronic noise at very low doses, or structure noise at very high doses.

Finally, it would be interesting to investigate whether gridded exposures of denser anatomies, such as the abdomen and pelvis, can be acquired without having to increase dose as required by the Bucky factor; this is an area for future study.

CONCLUSIONS

Contrary to traditional convention, we have demonstrated with images containing realistic clinical content (normal and abnormal), that it is possible to use scatter rejection grids for chest imaging with an Agfa CR system, irrespective of tube voltage, without having to increase DAK and therefore patient dose. It is probable that an increase in visibility of structures, especially in

the dense regions of the chest, outweighs the decrease of SNR (and therefore perceived increase in noise) in these regions. There is no justification to increase exposure factors (and therefore patient dose) as required by the traditional Bucky factor. This is possible because CR (and digital imaging systems in general) has a larger dynamic range than does film-screen. This will allow the MPE, in conjunction with clinical imaging

experts, to optimize chest imaging by utilizing scatter rejection grids if this is deemed clinically appropriate at their institution.

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REFERENCES

- Hart D, Wall BF, Hillier MC, Shrimpton PC. *Frequency and collective dose for medical and dental x-ray examinations in the UK 2008*. Chilton, UK: Health Protection Agency; 2010.
- The Ionising Radiation (Medical Exposure) Regulations 2000*. Norwich, UK: The Stationary Office Limited; 2000.
- Bonenkamp JG, Boldingh WH. Quality and choice of Potter Bucky grids. I. A new method for the unambiguous determination of the quality of a grid. *Acta Radiol* 1959; **51**: 479–89.
- Bonenkamp JG, Boldingh WH. Quality and choice of Potter Bucky grids. II. Application of the criterion of quality to various types of grids. *Acta Radiol* 1959; **52**: 149–57.
- Bonenkamp JG, Golding WH. Quality and choice of Potter Bucky grids. III. The choice of a Bucky grid. *Acta Radiol* 1959; **52**: 241–53.
- Boldingh WH. Quality and choice of Potter Bucky grids: parts IV and V. *Acta Radiol* 1962; **55**: 225–35.
- Morgan RH. An analysis of the physical factors controlling the diagnostic quality of roentgen images; contrast and the intensity distribution function of roentgen images. *Am J Roentgenol Radium Ther* 1946; **55**: 67–89.
- Bushong SC. *Radiologic science for technologists: physics, biology, and protection*. St Louis, MO: Mosby/Elsevier; 2008.
- Allisy-Roberts P, Williams JR. Imaging with X-rays. In: Horne T, ed. *Farr's physics for medical imaging*. Philadelphia, PA: Saunders/Elsevier; 2010. pp. 49–63.
- Council of European Communities. European guidelines on quality criteria for diagnostic radiographic images. Office for Official Publications of the European Communities; 1996.
- Fritz S, Jones AK. Guidelines for anti-scatter grid use in pediatric digital radiography. *Pediatr Radiol* 2014; **44**: 313–21. doi: [10.1007/s00247-013-2824-9](https://doi.org/10.1007/s00247-013-2824-9)
- Rowlands JA. The physics of computed radiography. *Phys Med Biol* 2002; **47**: R123–66.
- Baydush AH, Ghem WC, Floyd CE Jr. Anthropomorphic versus geometric chest phantoms: a comparison of scatter properties. *Med Phys* 2000; **27**: 894–7.
- Chan HP, Lam KL, Wu YZ. Studies of performance of antiscatter grids in digital radiography: effect on signal-to-noise ratio. *Med Phys* 1990; **17**: 655–64.
- Carlton RR, Adler AM. The grid. In: Bellegarde M, ed. *Principles of radiographic imaging: an art and a science*. New York, NY: Delmar Cengage Learning; 2012. pp. 257–72.
- Tanaka N, Naka K, Saito A, Morishita J, Toyofuku F, Ohki M, et al. Investigation of optimum anti-scatter grid selection for digital radiography: physical imaging properties and detectability of low-contrast signals. *Radiol Phys Technol* 2013; **6**: 54–60. doi: [10.1007/s12194-012-0169-y](https://doi.org/10.1007/s12194-012-0169-y)
- Fetterly KA, Schueler BA. Experimental evaluation of fibre-interspaced antiscatter grids for large patient imaging with digital x-ray systems. *Phys Med Biol* 2007; **52**: 4863–80.
- Bochud FO, Valley JE, Verdun FR, Hessler C, Schnyder P. Estimation of the noisy component of anatomical backgrounds. *Med Phys* 1999; **26**: 1365–70.
- Samei E, Flynn MJ, Eyler WR. Detection of subtle lung nodules: relative influence of quantum and anatomical noise on chest radiographs. *Radiology* 1999; **213**: 727–34.
- Samei E, Eyler W, Baron L. Effects of anatomical structure on signal detection. In: Beutel J, Kundel HL, Van Metter RL, eds. *Handbook of medical imaging. Vol. 1. Physics and psychophysics*. Bellingham, WA: SPIE—The International Society for Optical Engineering; 2000. pp. 655–82.
- Keelan BW, Topfer K, Yorkston J, Sehnert WJ, Ellinwood JS. Relative impact of detector noise and anatomical structure on lung nodule detection. *Proc SPIE* 2004; **5372**: 230–41.
- Sund P, Båth M, Kheddache S, Månsson LG. Comparison of visual grading analysis and determination of detective quantum efficiency for evaluation system performance in digital chest radiography. *Eur Radiol* 2004; **14**: 48–58.
- Håkansson M, Båth M, Börjesson S, Kheddache S, Johnsson AA, Månsson LG. Nodule detection in digital chest radiography: effect of system noise. *Radiat Prot Dosimetry* 2005; **114**: 97–101.
- Håkansson M, Båth M, Börjesson S, Kheddache S, Grahm A, Ruschin M, et al. Nodule detection in digital chest radiography: summary of the RADIUS chest trial. *Radiat Prot Dosimetry* 2005; **114**: 114–20.
- Båth M, Håkansson M, Börjesson S, Kheddache S, Grahm A, Ruschin M, et al. Nodule detection in digital chest radiography: introduction to the RADIUS chest trial. *Radiat Prot Dosimetry* 2005; **114**: 85–91.
- Båth M, Håkansson M, Börjesson S, Kheddache S, Grahm A, Bochud FO, et al. Nodule detection in digital chest radiography: part of image background acting as pure noise. *Radiat Prot Dosimetry* 2005; **114**: 102–8.
- Båth M, Håkansson M, Börjesson S, Hoeschen C, Tischenko O, Kheddache S, et al. Nodule detection in digital chest radiography: effect of anatomical noise. *Radiat Prot Dosimetry* 2005; **114**: 109–13.
- Moore CS, Liney GP, Beavis AW, Saunderson JR. A method to produce and validate a digitally reconstructed radiograph-based computer simulation for optimisation of chest radiographs acquired with a computed radiography imaging system. *Br J Radiol* 2011; **84**: 890–902. doi: [10.1259/bjr/30125639](https://doi.org/10.1259/bjr/30125639)
- Moore CS, Avery G, Balcum S, Needler L, Swift A, Beavis AW, et al. Use of a digitally reconstructed radiograph-based computer simulation for the optimisation of chest radiographic techniques for computed radiography imaging systems. *Br J Radiol* 2012; **85**: e630–9. doi: [10.1259/bjr/47377285](https://doi.org/10.1259/bjr/47377285)
- Moore CS, Wood TJ, Beavis AW, Saunderson JR. Correlation of the clinical and physical image quality in chest radiography for average adults with a computed radiography imaging system. *Br J Radiol* 2013; **86**: 20130077. doi: [10.1259/bjr.20130077](https://doi.org/10.1259/bjr.20130077)
- Moore CS, Wood TJ, Avery G, Balcum S, Needler L, Beavis AW, et al. An investigation of automatic exposure control calibration for chest imaging with a computed radiography system. *Phys Med Biol* 2014; **59**: 2307–24. doi: [10.1088/0031-9155/59/9/2307](https://doi.org/10.1088/0031-9155/59/9/2307)

32. Sutton DG, Reilly AJ. *Report 78 spectrum processor*. York, UK: Institute of Physics and Engineering in Medicine; 1997.
33. Båth M, Håkansson M, Tingberg A, Månsson LG. Method of simulating dose reduction for digital radiographic systems. *Radiat Prot Dosimetry* 2005; **114**: 253–9.
34. Floyd CE Jr, Lo JY, Chotas HG, Ravin CE. Quantitative scatter measurement in digital radiography using a photostimulable phosphor imaging system. *Med Phys* 1991; **18**: 408–13.
35. Hiles PA, Mackenzie A, Scally A, Wall B, eds. *IPEM report 91: recommended standards for the routine performance testing of diagnostic x-ray imaging systems*. York, UK: Institute of Physics and Engineering in Medicine; 2005.
36. Tingberg A, Herrmann C, Lanhede B, Almén A, Sandborg M, McVey G, et al. Influence of the characteristic curve on the clinical image quality of lumbar spine and chest radiographs. *Br J Radiol* 2004; **77**: 204–15.
37. Tingberg A, Sjöström D. Optimisation of image plate radiography with respect to tube voltage. *Radiat Prot Dosimetry* 2005; **114**: 286–93.
38. Lanhede B, Båth M, Kheddache S, Sund P, Björnell L, Widell M, et al. The influence of different technique factors on image quality of chest radiographs as evaluated by modified CEC image quality criteria. *Br J Radiol* 2002; **75**: 38–49.
39. Shrimpton PC, Wall BF, Fisher ES. The tissue-equivalence of the Alderson Rando anthropomorphic phantom for x-rays of diagnostic qualities. *Phys Med Biol* 1981; **26**: 133–9.
40. Wood TJ, Davis AW, Moore CS, Beavis AW, Saunderson JR. Validation of a large-scale audit technique for CT dose optimisation. *Radiat Prot Dosimetry* 2012; **150**: 427–33. doi: [10.1093/rpd/ncr434](https://doi.org/10.1093/rpd/ncr434)