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Rejection and redistribution of scattered radiation in Scan Equalization Digital Radiography (SEDR): simulation with spot images

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Abstract

The anti-scatter grid has been widely used to reject scatter and increase the perceptibility of low contrast object in chest radiography; however it also attenuate the primary x-rays, resulting in a substantial loss of information and an increased relative noise level in heavily attenuated regions. A more dose efficient approach to scatter rejection is the slot-scan imaging technique. Another problem in chest radiography is the low transmitted x-ray intensity in heavily attenuating regions. It results in higher relative noise level and subject the contrast sensitivity to limitation by the system noises. A solution to this problem is through the exposure equalization technique with which the incident x-ray intensity is regionally modulated to compensate for the differences of x-ray attenuation due to the anatomic variation. We are in the process of implementing the scan equalization digital radiography (SEDR) technique which combines the advantages of slot-scan imaging and exposure equalization. However, associated with the use of exposure equalization is a re-distribution of scattered radiation at the detector which may impact on the benefit of using exposure equalization in conjunction with the slot-scan imaging geometry. In order to understand the scatter properties and their impact in SEDR, we have used spot collimated aSi FP images to synthesize simulated SEDR images with which scatter components, primary signals, and scatter-to-primary ratios (SPRs) were measured. Using a simple model for scatter effects, we have also estimated and compared the contrast-to-noise ratio degradation factors (CNRDFs). It was found that for quantum limited situations the slot-scan technique has resulted in a substantial improvement of the image quality as indicated by higher estimated CNRDFs. Compared to slot-scan imaging, SEDR resulted in higher SPRs in the lungs and lower SPRs in the mediastinum. In the sub-diaphragmatic regions, the SPRs remain about the same. This correspond to lower CNRDFs in the lungs, higher CNRDFs in the mediastinum, and about the same CNRDFs in the sub-diaphragmatic regions. It was shown that although SEDR has resulted in minimum improvement over slot-scan imaging in reducing the SPRs, it could improve the contrast sensitivity by raising the primary signal levels in heavily attenuating regions. This advantage needs to be further investigated in our continuing study of the SEDR technique.

Keywords

Scattered radiation; scatter rejection; slot-scan imaging; exposure equalization; digital radiography; flat-panel detector

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I. INTRODUCTION

Since the 1980s, various digital radiography techniques have been developed in the hope of improving both the quality and utilization of x-ray image data.¹⁻¹⁵ The rapid developments in advanced electronics and computer technology have made the concept of totally digital and filmless radiographic operation feasible. Recently, the flat-panel (FP) based digital radiography systems have rapidly gained acceptance for chest imaging as one of the candidates among many other digital imaging techniques such as charge-coupled-device (CCD) and computed radiography (CR) for implementing digital radiography. It is expected that the diagnostic performance of digital imaging will be superior to that of conventional screen/film (S/F) imaging and digital imaging will provide a dose efficiency that is similar to or better than that of S/F imaging. Improvements in digital radiography have been mainly focused on image acquisition speed, resolution, x-ray absorption efficiency, and software tools for image display. However, two problems intrinsic to projection radiography still remain and degrade the low contrast performance with the use of an area digital detector as x-ray receptor as is in S/F imaging: (1) lack of efficient methods for rejecting scattered radiation, and (2) low x-ray photon fluence at the detector in heavily attenuated regions.

Scattered radiation has long been recognized as a main cause for degraded image quality in projection imaging as it decreases the image contrast and contrast-to-noise ratio (CNR) for the object being detected or visualized.¹⁶⁻¹⁹ Various techniques have been investigated and implemented to reject scattered radiation during image acquisition, the most practical of which being the air gap and the anti-scatter grid methods.²⁰⁻²² By increasing the distance between the patient and the detector, the amount of detected scattered radiation can be reduced; however, air gaps may lead to an increased patient dose, reduced field-of-view, and increased focal spot blurring effect. The anti-scatter grid method has been widely used in many radiographic and fluoroscopic procedures with conventional S/F or image intensifier based imaging systems. Even after transitioning from analog to digital imaging techniques, the use of anti-scatter grid was carried over in most cases for scatter rejection. The main advantage of the anti-scatter grid is its simplicity. Grids reject scattered radiation effectively, but they also attenuates a substantial fraction of the primary x-rays (30–50%), resulting in a substantial loss of information and an increased relative noise level in the image.²³ To keep the relative noise at the same level, the exposure must be increased to compensate for the attenuation by the grid. Various slot-scan imaging techniques have been proposed and investigated to offer better scatter rejection without having to attenuate the primary x-rays. With these techniques, a collimated narrow fan beam was used in conjunction with a dynamically collimated area detector or a tracking detector array to scan across the patient.²⁴⁻³⁵ The main advantage of the slot-scan technique is its dose efficiency, as primary x-rays do not have to be attenuated. However, it requires accurate and precise alignment of the fan beam collimator with the x-ray tube focal spot.

A second issue is that heavy x-ray attenuation in regions like the retrocardia, mediastinum, and subdiaphragm results in poorer image signal-to-noise ratio (SNR), leading to poorer detectability and visualization of low contrast objects. To overcome this problem, the x-ray beam incident to the patient may be spatially modulated to compensate for variation of patient attenuation. Several exposure equalization techniques have been implemented and investigated in the past,³⁶⁻⁴⁹ all of which have demonstrated improved image quality particularly in the heavily attenuated regions. These techniques may be divided into two categories: (1) those based on open field image acquisition and (2) those based on slot-scan imaging geometry. The former usually relied on the use of a two dimensional regional beam attenuator to equalize the exposure on the detector. An anti-scatter grid is usually used to reduce the scattered radiation detected, as in conventional screen/film or digital radiography. Due to the use of beam attenuation for exposure equalization, the quality of the incident x-ray beam is non-uniform.

The slot-scan imaging based approach, often referred to as the scan equalization radiography (SER), employs a narrow fan beam to scan the patient. An area detector used in conjunction with an aft-collimator or a matching rectangular detector array are used to detect x-rays within the projected fan beam while preventing scattered x-rays outside the fan beam area from being detected. Spatially and temporarily varying beam width or pulse width modulation has been used to achieve exposure equalization within the fan beam area during the scan. Because these methods do not rely on the use of x-ray attenuation for exposure equalization, they do not alter the x-ray beam quality or the object contrast. The two dimensional beam attenuator approach has the advantage of allowing the area detector to be used in the regular way, enabling short exposures and lower tube loading in screen/film based imaging and fast framing rate in II based imaging. The SER techniques combine the advantages of scatter rejection without having to attenuate primary x-rays and exposure equalization without the need to harden the x-rays. On the other hand, they require a longer scanning time, although the effective exposure time is as short as in conventional radiography, and greater tube loading. Thus, the SER techniques can best be applied to stationary imaging modality, such as chest radiography or mammography, for improving the image quality.

Various SER techniques were initially designed, implemented and investigated for use with a screen/film combination mainly to improve image contrast in heavily attenuating regions. The SER technique can also be applied to digital radiography detector for image quality improvement, resulting in a technique referred to as the scan equalization digital radiography (SEDR). To explore and investigate the benefits of SEDR in chest imaging, we are in the process of implementing SEDR with an amorphous silicon (aSi) flat panel (FP) detector using the beam width modulation approach for exposure equalization. While the construction is still undergoing, we have used spot collimated aSi FP images to synthesize simulated SEDR images to investigate the rejection and re-distribution of scattered radiation. In this paper, we report on the scatter properties measured in two different regions in the simulated SEDR image of an anthropomorphic chest phantom and compare them with those of a slot-scan image and an open field image. Using a simple model for scatter effects, we have also estimated and compared the resulting image quality of the three different techniques.

II. EFFECTS OF SCATTERED RADIATION ON IMAGE QUALITY

Image contrast may be characterized by the contrast ratio (CR), defined as the difference between the mean signal of a low-contrast object and that of its surrounding area divided by the latter. In absence of scatter signals, the contrast ratio, CR_p , may be expressed as follows:

$$CR_p = \frac{I_p - I_p'}{I_p} \quad (1)$$

Where I_p' and I_p , are the primary only signals inside and outside the object. The noise properties of the image signals may be characterized by the signal-to-noise ratio (SNR). In absence of scatter signals, the SNR, SNR_p , may be computed as follows:

$$SNR_p = \frac{I_p}{\sigma_p} \quad (2)$$

where σ_p is the noise level in the scatter free (primary) image signal. The ability for a low contrast object to be detected and visualized in an image may be determined by the contrast-to-noise ratio (CNR) defined as the signal difference between contrast object and the

surrounding area divided by the noise level rather than the CR or SNR alone. In absence of scatter signals, the CNR, CNR_P , may be computed as follows:

$$\begin{aligned} CNR_P &= \frac{I_p - I_p'}{\sigma_p} = \frac{I_p - I_p'}{I_p} \cdot \frac{I_p}{\sigma_p} \\ &= CR_P \cdot SNR_P \end{aligned} \quad (3)$$

The scatter component in the image signals is a smooth function of the position and varies little across a reasonably small object (e.g. ≤ 1 cm). Thus, the scatter component may be assumed to be the same inside and outside the object with the value, I_S . Using the previous definitions, the scatter degraded CR and CNR, CR_{P+S} and CNR_{P+S} which may be computed again and linked to CR_P and CNR_P as follows:

$$\begin{aligned} CR_{P+S} &= \frac{(I_p + I_S) - (I_p' + I_S)}{I_p + I_S} \\ &= \frac{I_p - I_p'}{I_p + I_S} \\ &= \frac{(I_p - I_p')/I_p}{1 + I_S/I_p} \\ &= \frac{CR_P}{1 + I_S/I_p} \end{aligned} \quad (4)$$

$$\begin{aligned} CNR_{P+S} &= \frac{(I_p + I_S) - (I_p' + I_S)}{\sigma_{P+S}} \\ &= \frac{(I_p + I_S) - (I_p' + I_S)}{I_p + I_S} \cdot \frac{I_p + I_S}{\sigma_{P+S}} \\ &= CR_{P+S} \cdot SNR_{P+S} \end{aligned} \quad (5)$$

where σ_{P+S} is the noise level in the scatter added image signal. To simplify Eqs. 4 and 5, we assume that the x-rays are mono-energetic at effective energies characterizing the primary and scatter x-ray spectra. With this assumptions, the image signals, I_p and I_S , may be considered as proportional to the number of primary and scattered x-rays incident onto a detector pixel, P and S , respectively, as follows:

$$I_p = g_p \cdot P \quad (6)$$

$$I_S = g_S \cdot S \quad (7)$$

where g_p and g_S are the gain factors at the effective energies for the primary and scattered x-rays. Eqs. 4 and 5 contain the levels of image noises which consist of a quantum noise component originating from the detection of x-rays and a system noise component originating from conversion of the detected x-rays into the final image signals. Since the signals acquired were usually high inside and along the edges of the slot area, we further assume that the system noises are negligible. Thus, the noise level may be considered as proportional to the square root of the photon number and σ_{P+S} and SNR_{P+S} may be expressed as:

$$\begin{aligned}
 \sigma_{p+s} &= \sqrt{\sigma_p^2 + \sigma_s^2} = \sqrt{(g_p \sqrt{P})^2 + (g_s \sqrt{S})^2} \\
 &= (g_p \sqrt{P}) \sqrt{1 + \left(\frac{g_s \sqrt{S}}{g_p \sqrt{P}}\right)^2} \\
 &= \sigma_p \sqrt{1 + \left(\frac{I_s}{I_p}\right) \left(\frac{g_s}{g_p}\right)}
 \end{aligned} \tag{8}$$

$$\begin{aligned}
 SNR_{p+s} &= \frac{I_p + I_s}{\sigma_{p+s}} \\
 &= \frac{I_p (1 + I_s/I_p)}{\sigma_p \sqrt{1 + \left(\frac{I_s}{I_p}\right) \left(\frac{g_s}{g_p}\right)}} \\
 &= SNR_p \cdot \frac{1 + I_s/I_p}{\sqrt{1 + \left(\frac{I_s}{I_p}\right) \left(\frac{g_s}{g_p}\right)}}
 \end{aligned} \tag{9}$$

CNR_{p+s} can then be rewritten by substituting expressions in Eqs. 4 and 9 for CR_{p+s} and SNR_{p+s} in Eq. 5 as follows:

$$\begin{aligned}
 CNR_{p+s} &= \frac{CR_p}{1 + I_s/I_p} \cdot SNR_p \cdot \frac{1 + I_s/I_p}{\sqrt{1 + \left(\frac{I_s}{I_p}\right) \left(\frac{g_s}{g_p}\right)}} \\
 &= \frac{CNR_p}{\sqrt{1 + \left(\frac{I_s}{I_p}\right) \left(\frac{g_s}{g_p}\right)}}
 \end{aligned} \tag{10}$$

CR_{p+s} , SNR_{p+s} and CNR_{p+s} are all proportional to their scatter free counterparts, CR_p , SNR_p and CNR_p , with the factors of change dependent on the scatter-to-primary ratio (SPR) defined as I_s/I_p . Thus, the SPR is widely used to quantify the presence of scattered radiation in image signals. The detection and visualization of a low contrast object is mainly related to the CNR and the object size unless the overlapping anatomical structures become a limiting factor. Eq. 10 shows that in the presence of scattered radiation, the CNR is degraded by a factor which is referred to as the CNR degradation factor (CNRDF) in this manuscript, and may be used to quantify the effects of scattered radiation on the image quality:

$$\begin{aligned}
 CNRDF &= \frac{CNR_{p+s}}{CNR_p} \\
 &= \frac{1}{\sqrt{1 + \left(\frac{I_s}{I_p}\right) \cdot \left(\frac{g_s}{g_p}\right)}}
 \end{aligned} \tag{11}$$

Eq. 11 shows that with a large amount of scattered radiation present in chest imaging, the CNR may be substantially degraded, especially in heavily attenuated regions such as the

mediastinum and subdiaphragm where the primary signals are at a very low level while a large amount of scatter may spill from the neighboring lung areas, resulting in a substantially elevated SPRs.

Since g_S/g_P appears in the expressions for SNR_{P+S} , CNR_{P+S} , and CNR_{DF} , it may also affect the effects of scatter on these quantities. The gain factor is roughly proportional to the photon energy and the corresponding energy absorption ratio. As the photon energy increases from the effective energy of the scattered x-rays (e.g. 55.4 keV) to that of the primary x-rays (e.g. 69.3 keV for 120 kVp x-rays), the absorption ratio decreases, thus partially compensating for the energy increase and resulting in a largely unchanged gain factor. We have computed g_S/g_P for a range of different combinations of effective energies for primary and scattered x-rays. The results show that the values of g_S/g_P range from 1.02 for 10 cm tissue to 1.11 for 30 cm tissue at 120 kVp. Thus, with the same SPR, the CNR would be degraded by a larger factor while the SNR would increase by a smaller factor with a higher effective photon energy of the primary x-rays (higher kVp setting or larger patient thickness). The effect is equivalent to a 2–11% increase of the SPR.

III. MATERIALS AND METHODS

To investigate the rejection and redistribution of scattered radiation in SEDR, we have acquired and used spot collimated images to synthesize simulated SEDR images. To achieve the goal, a two-dimensional collimator system was mounted on a linear scanner to generate a rectangular shape spot x-ray beam and scan it across an anthropomorphic chest phantom. Images were acquired with a commercial flat-panel based digital radiographic system. Each spot image consists of largely primary signals inside a $1.4 \text{ cm} \times 1.0 \text{ cm}$ spot area and scatter only signals outside the spot area. Exposure modulation factors were computed from a slot-scan image. These factors were multiplied with the spot image data with the outside-spot data cut to obtain the simulated SEDR image data. SPRs, SRFs and CNRDFs were measured in the SEDR image and compared with those from an un-equalized slot-scanning image, and full-field images with and without anti-scatter grid in place for various regions of interest (ROI).

A. Image acquisition

The digital radiography system used in this investigation was an amorphous silicon/cesium iodide [a-Si:H/CsI(Tl)] flat panel (FP) based digital chest unit (Revolution XQ/i, GE Medical Systems, Milwaukee, WI) with an active image area of $40.4 \text{ cm} \times 40.4 \text{ cm}$ (15.9 in $\times$ 15.9 in). The detector has an active image matrix size of 2022×2022 pixels and a pixel pitch of 200 μm . However, in normal clinical mode (tube switch set to “normal”), the system generates images with a matrix size of 1736×2022 pixels, corresponding to an approximately 14 in $\times$ 16 in field of view for chest imaging. The system was manufactured as a stand-alone digital chest unit and integrated with a stationary but removable anti-scatter grid with a 13:1 grid ratio and 78 lines/cm (Mitaya MFG. Co., Ltd., Tokyo, Japan). The characteristics and performance of this system have been studied and reported on previously.^{1–3} The readout signals of the digital radiography system are digitized into 14-bit image data and stored both as linear exposure data (with dark current and gain corrected) and as processed (log-mapped) image data for archival, display, and printing. For this study, the unprocessed linear exposure data were transferred to a Pentium IV research computer via a network for further processing and analysis. Both spot and slot images were acquired with the anti-scatter grid removed, but an ionization chamber and detector cover were still in the x-ray path. The anti-scatter grid was only inserted to acquire full-field images for comparison with the slot-scan and SEDR images and necessary slot collimated images for estimating the primary and scatter signals in these full field images. Full field images were also acquired with the anti-scatter grid removed for comparison. An anthropomorphic chest phantom (Radiology Support Devices, Long Beach,

CA) with attenuation and scatter radiation properties similar to those of a typical medium-sized patient was imaged in the posterior-anterior (PA) position.

A computer-controlled scanning fore-collimator system was mounted at the output of the x-ray tube to generate spot or slot collimated images. The collimator was mounted on two high-precision linear drives (Velmex, Inc., Bloomfield, NY) integrated together to allow for two dimensional positioning or scanning. The collimators were made of 2 mm thick tungsten which attenuates nearly all x-rays at 120 kVp. To minimize the focal spot blurring effects, the tungsten edges were cut with a relief angle of 16° on each side of the collimator. The spot/slot width and position could be adjusted and positioned independently. The system setup for acquiring spot collimated images with a commercial flat-panel based digital radiography system is shown in Fig.1.

A series of slot collimated images were acquired to construct simulate slot-scan images corresponding to various fan beam widths: 1, 2, 3 and 4 cm. A horizontally oriented slot collimator was constructed and used to generate a 1 cm wide fan beam on the detector. The projected fan beam was moved downward by 0.5 cm (vertically) after each exposure to acquire a series of images to cover the entire detector area with 50% overlapping between each two consecutive images. Each slot collimated image consisted of primary plus scatter signals inside the slot area and scatter only signals outside the slot area. Scatter signals outside the slot area were removed during the superimposition process to simulate the slot-scanning geometry desired. Slot collimated images in the simulated beam areas were also acquired with the anti-scatter grid inserted to estimate the primary and scatter signals in full field images acquired with the grid inserted.

A series of spot collimated images were acquired to construct simulated SEDR image signals for two rectangular areas in the chest images: one at the middle lung level, the other in sub-diaphragmatic area. The dimensions of the areas were chosen to be 33.6 cm wide and 4 cm high to allow SEDR images to be simulated for a beam width of 1, 2, 3 and 4 cm. A spot collimator was constructed and used to generate a 1.4 cm wide and 1 cm high spot beam on the detector. 4 rows of 24 spot collimated images were acquired by first moving the projected spot area horizontally by 1.4 cm after each exposure and then vertically by 1 cm after acquisition a complete row of images. The horizontal edges of these images are aligned with slot collimated images to allow both slot-scan and SEDR images to be simulated for the same beam area and the scatter signals, primary signals and SPRs measured along top edges of all simulated images. Each spot collimated image consisted of primary plus scatter signals inside the spot area and scatter only signals outside the spot area. Scatter signals outside the slot area were removed during the superimposition process to simulate the slot-scanning geometry desired. Scatter signals outside the spot area but inside the slot area were retained and were added to the overall image signals as the in-slot scatter component.

All exposures were made at 120 kVp and 2 mAs, which are the typical technique settings for diagnostic chest radiography procedures. The source-to-image distance (SID) was kept at 183 cm (72 in) with the collimator placed at ~52 cm from the focal spot leading to a magnification factor of ~3.5. Exposures were made in the small focal spot (nominal 0.6 mm) mode to minimize focal spot blurring of the spot/slot edges. A 1-minute delay between two consecutive acquisitions was used to prevent overheating of the x-ray tube.

B. Simulation of slot-scan images for larger beam widths

The slot-scan images acquired were obtained with 1 cm wide beam width. Slot-scan images corresponding 2, 3 or 4 cm wide beam width were obtained by superimposing 2, 3 or 4 contiguous, non-overlapping 1 cm slot-scan images together. During superimposition, scatter signals from different slot collimated images were added into larger scatter components inside

and outside the simulated beam area, reflecting the fact that slot-scan imaging with wider beam area is less efficient in scatter rejection. The simulated slot-scan images contain primary signals only in the simulated slot area. They also contain scatter signals in and outside the slot area. Scatter signals outside the slot area were kept to allow the scatter components along the slot edge to be estimated. In actual implementation of slot-scan imaging, scatter signals outside the slot area would be rejected by using a slot shaped array detector or an area detector with an aft-collimator or dynamically regional resetting and activation. A largely artifact free full field slot-scan image was also constructed by extracting and superimposing only the central 0.5 cm wide part of the overlapping slot collimated images. This image is necessary for computing the exposure equalization factors as will be described in the next section. It should be noted that these images were generated specifically for this simulation study only. The image acquisition and processing methods were devised and used to measure the scatter components and SPRs along the top edges of the simulated beam area and they should not be confused with techniques for actual implementation of the slot-scanning or SEDR imaging which would involve many practical considerations like the scanning time, scanning artifacts, patient dose and efficiency in x-ray usage.

C. Simulation of SEDR images

The objective of exposure equalization are to regionally modulate the input exposure level to achieve coarsely (averaged over a large area) uniform exposure levels over the main part of the chest image while keeping the overall average input exposure level at the same level as in conventional chest radiography. In this study, the detector (image) area was divided into 24 (horizontally) by 40 (vertically) $1.4 \text{ cm} \times 1 \text{ cm}$ or 70×50 pixels sub-regions, indexed by (i, j) for exposure equalization. These regions were aligned with those over which spot collimated images were acquired so that the latter could be used to synthesize and compare the simulated SEDR images with the slot-scan images simulated for the two beam locations (middle lung and sub-diaphragm) and four different slot widths (1, 2, 3, and 4 cm). To compute the exposure equalization factors for these sub-regions, a central area which covers the lung, mediastinal, and sub-diaphragmatic regions was selected as the reference area (Fig. 2) and the exposure equalization factor for the sub-region (i, j), $E(i, j)$, was computed from a full-field slot-scan image as follows:

$$E(i, j) = \frac{\bar{I}}{I(i, j)} \quad (12)$$

where $\bar{I}$ is average image signal over the reference area and $\overline{I(i, j)}$ is average signal over the sub-region (i, j).

The SEDR image for a specific beam area, I_{SEDR} , was computed by first weighting the appropriate spot collimated image and then superimposing them as follows:

$$\begin{aligned} I_{SEDR} &= \sum_{i,j} E(i, j) \cdot I^{SP}(i, j) \\ &= \sum_{i,j} \frac{\bar{I}}{I(i, j)} \cdot I^{SP}(i, j) \end{aligned} \quad (13)$$

where $I^{SP}(i, j)$ is the spot collimated image for the region (i, j) and the summation is over the spot collimated images covering the simulated beam area. Since $I^{SP}(i, j)$'s include scatter signals outside the spot collimated area, I_{SEDR} contains primary signals covering the simulated

beam area as well as scatter signals outside the area. The latter were used to estimate the scatter component in signals inside the simulated beam area. Notice that in using beam width modulation to implement scan equalization radiography, the active detector area, defined by a scanning aft-collimator or a moving slot shaped detector, usually covers a fixed rectangular area coinciding with full width x-ray fan beam due to the difficulty to design an aft-collimator or detector to regionally alter the width of the active detector area. Thus, the scatter rejection geometry with such SEDR is essential the same as with slot-scanning imaging with uniform full slot width, which was simulated in this study. The simulated SEDR images contain primary signals only in the simulated slot area and were subject to artifacts along edges between the 1 cm slot images. Since the main use of the images is to measure and study the scatter properties along the edge of the simulated edge, complete elimination of these artifacts is desired but unnecessary. Signal smoothing was used to reduce these artifacts in the measurement and estimation of the scatter and primary signal profiles but could not totally eliminate their effects. Fig. 3 shows the superimposition of (a) weighted spot collimated images or (b) slot collimated images to simulate an SEDR or slot-scan images in lung regions, respectively, for a wider slot width.

D. Measurement of primary, scatter signals and SPRs

Various methods have been developed to measure scatter components, including the beam stop, slit (e.g. steel plate), and edge spread methods.^{17,51,52,53} The beam stop method can be used to precisely measure and estimate the scatter component at any location. However it is impractical to use this method to measure the spatial distribution of scatter on a pixel-by-pixel basis. The slit and edge spread methods, on the other hand, can be used to measure the scatter distribution.

In this study, we used a method similar to the edge spread method, referred to as the slot edge method in this manuscript, to estimate the scatter component in the image signals along the upper edge of a simulated beam area. With this method, the first derivative of a vertical signal profile across the simulated beam area is computed as a function of vertical position. Example of a signal profile and its derivative are shown in Fig. 4. Across the first (upper) edge in the signal profile, the primary signal increases abruptly from zero (totally blocked by the collimator) to a full strength (completely inside the simulated beam area) value. Across the same region, the value of the first derivative increases significantly from zero to a maximum value and then returns to zero again. Thus, the image signals at where the first derivative just begins to rise above zero may be measured as an estimate for the scatter component in the image signal at where the first derivative nearly returns to zero. These two signals may be measured along the (upper) edge of the simulated slot area on a pixel-by-pixel basis to obtain a scatter signal profile, $I_S(x)$, to estimate the scatter component in the edge image signal profile, $I_{P+S}(x)$, which contains both primary and scatter signals. By subtracting the estimated scatter component from the total (primary plus scatter) signal, the primary signal profile, $I_P(x)$, along the (top) slot/spot edge may be computed as follows:

$$I_P(x) = I_{P+S}(x) - I_S(x) \quad (14)$$

Due to focal spot blurring, the scatter component and image signal are measured at slightly offset locations. This may cause the scatter components to be slightly underestimated and the primary signals slightly overestimated. Notice that the estimated scatter components and total image signals may be similarly measured and used to compute the primary signals along the lower edge of the simulated beam area, corresponding to the right edge in Fig. 4. However, they should be measured at where the first derivative just begins to decrease and where it nearly returns to zero, respectively.

The primary signals estimated with Eq. 14 are the same as the primary signals in a full-field image. Thus, they may be subtracted from image signals in the full field image to estimate the scatter component in the full-field image, $I_s^{ff}(x)$, along the same locations (upper edge of the simulated beam area). Similarly, slot collimated images acquired with anti-scatter grid inserted were acquired in the simulated beam areas to determine the primary and scatter signals along the top edge of the simulated beam areas in a full field image acquired with the anti-scatter grid inserted.

Following the measurement of $I_S(x)$, $I_{P+S}(x)$, $I_P(x)$ and $I_s^{ff}(x)$, the position-dependent scatter-to-primary ratio (SPR) and scatter reduction fraction (SRF),⁵⁰ $SPR(x)$ and $SRF(x)$, may be computed as follows:

$$SPR(x) = \frac{I_s(x)}{I_p(x)} = \frac{I_s(x)}{I_{P+S}(x) - I_s(x)} \quad (15)$$

$$SRF(x) = \frac{I_s^{ff}(x) - I_s(x)}{I_s^{ff}(x)} \quad (16)$$

E. Contrast-to-noise ratio degradation factor (CNRDF)

The main objective of this study is to study and compare the scatter rejection properties for the slot-scan and SEDR images. The data acquired did not allow us to directly measure the signal-to-noise ratio (SNR) in the image. However, using Eq. 11, the CNRDF was estimated from the measured SPRs along the top edge of the simulated beam area on a pixel-by-pixel basis as follows:

$$CNRDF(x) = \frac{1}{\sqrt{1 + SPR(x) \cdot \left(\frac{g_S}{g_P}\right)}} \quad (17)$$

where g_S and g_P were assumed to be proportional to the photon energy and energy absorption ratio. The ratio, g_S/g_P , was computed by assumed that the 120 kVp primary x-rays transmitting through 20 cm of tissue has an effective energy of 69.3 keV while the scattered x-rays have an effective energy of 55.4 keV.^{54,55} The ratio was estimated to be around 1.103 for 400 μ m thick CsI layer.⁵⁶

IV. RESULTS

The primary signals of the chest phantom images acquired with the slot-scan, SEDR, and conventional (full-field with anti-scatter grid) techniques are plotted in Fig. 5(a) for the lung region and in Fig. 5(b) for the sub-diaphragmatic region. Both the slot-scan and SEDR images were simulated for a 1 cm wide beam area. The primary signals in the slot-scan images are the same as those in the full-field images acquired without the grid. Thus their plots may also represent the primary signals in the latter. Also, since the primary signals in the slot-scan images are the same for different beam widths, the plots for the slot-scan images simulated for 1 cm beam width may also represent those for 2, 3 and 4 cm beam widths. The plots show that full-field imaging with the anti-scatter grid in place reduced the primary signals by as much as 40%,

as compared to the slot-scan imaging or full field imaging without the anti-scatter grid. SEDR, on the other hand, produced largely uniform primary signals with a substantial increase in heavily attenuated regions (e.g. mediastinum, sub-diaphragm) and a substantial decrease in the lightly attenuating areas (e.g. lungs).

The scatter components in the four different types of images are plotted in Fig. 6(a) for the lung and Fig. 6(b) for the sub-diaphragm. Since the scatter components in a slot-scan image vary with the beam width, they were separately plotted for the four different beam widths: 1, 2, 3 and 4 cm. Both the anti-scatter grid and slot-scan methods substantially reduced scatter signals in all regions. However, even with a 4-cm slot width, the slot-scan method still performed better with much lower scatter signals than the anti-scatter grid method in sub-diaphragm. SEDR resulted in uniformly distributed scattered radiation but slightly increased intensity in heavily attenuated regions. The position-dependent SPRs are plotted for comparison, as shown in Fig. 7(a) for the lung and Fig. 7(b) for the sub-diaphragm. The slot-scan method was more effective in reducing the SPRs than the anti-scatter grid method mainly because the grid also reduced primary x-rays while rejecting the scattered x-rays. It can be made even more effective with the use of narrower slot widths, leading to even lower SPRs. The SEDR technique through equalizing the detector exposure, on the other hand, produced reduced and largely uniform SPRs. In the lightly attenuation regions, the SPRs with the SEDR images were higher than those with slot-scan images with the same imaging geometry, i.e. 1 cm slot width. In the heavily attenuating regions, they became slightly lower. This is due to the combined effects of exposure equalization and the resulting redistribution of scattered radiation.

The position-dependent SRFs, used to quantify the ability of scatter rejection for various imaging techniques, are plotted for comparison in Fig. 8(a) for the lung region and in Fig. 8 (b) for the sub-diaphragm region. The slot-scan technique with a slot width ranging from 1 to 4 cm resulted in overall better scatter rejection than the anti-scatter grid method except that in the mediastinum, the SRFs with the 4 cm slot width were similar to those with the anti-scatter grid technique. The SRFs improved as the slot width used became narrower. The SRFs with the slot-scan and anti-scatter grid techniques tended to be higher in heavily attenuated regions and lower in lightly attenuated regions. Those with the SEDR technique, on the other hand, were largely uniform. This was because with the SEDR technique, scatter signals in the SEDR and full field, no grid images were both largely uniform with only slight spatial variation.

The CNRDFs were plotted as a function of the position and compared in Fig. 9(a) for the lung and in Fig. 9(b) for the sub-diaphragm. With a slot width of 1 to 4 cm, the CNRDFs with the slot-scan method were higher than those with the anti-scatter grid method, indicating a significant improvement in the low-contrast performance through more effective scatter rejection. The improvement was more pronounced in the heavily attenuated regions than in the lightly attenuated regions because the image quality was already high in the lightly attenuated regions (due to less attenuation). A narrower slot width could also result in higher CNRDFs. The CNRDFs with the SEDR technique were largely uniform in both lung and sub-diaphragm. They were slightly higher than those with the slot-scan technique with the same (1 cm) slot width in heavily attenuated regions but slightly lower in the lightly attenuated regions. CNRDFs were plotted as a function of the position for SEDR images acquired with various slot widths: 1, 2, 3, and 4 cm in Fig. 10(a) for the lung and in Fig. 10(b) for the sub-diaphragm. The CNRDFs decreased as the slot width used increased.

V. DISCUSSION AND CONCLUSIONS

In this study, we simulated slot-scan and SEDR images for four different slot widths and used them to study their scatter rejection properties in the lung and sub-diaphragmatic regions. Using

a simple model for effects of scatter on image CNRs, we also estimated the CNRDFs for these images. The results were compared with those acquired with open field exposures with and without an anti-scatter grid. It was shown that approximately 80% of scattered radiation was rejected by the anti-scatter grid in heavily attenuated regions and 70% in lightly attenuated regions. However, the primary signals were also attenuated by approximately 40%, thus greatly reducing, if not eliminating, the benefit of using an anti-scatter grid. With the slot-scan method, scatter reduction by as high as 95% (with 1 cm wide slot) could be achieved without attenuation of the primary x-rays, hence more effective (depending on the slot width used) and efficient than using the anti-scatter grid. This resulted in better image quality as indicated by higher estimated CNRDFs with the slot-scan method.

SEDR combines slot-scan imaging with regional exposure equalization which helps increase transmitted x-ray intensity in heavily attenuating regions and decrease it in lightly attenuating regions, thus achieving more uniform exposures to the detector. However, it was found that exposure equalization was accompanied by a substantial increase of the scatter components in heavily attenuating regions and a decrease in lightly attenuating regions. Thus, a slight improvement of image quality, as measured by the estimated CNRDFs, was observed in the mediastinum at the expense of slightly degraded image quality in the lungs. In sub-diaphragmatic regions, the image quality remained about the same as with slot-scan imaging.

The above observations seem to indicate that there is only slight or no advantage to add regional exposure equalization to slot-scan imaging. However, our estimation of the CNRDF was based on the assumption that the image noises were quantum in nature and proportional to the square root of the number of photons detected. In a slot-scan or open field image, the primary signal levels in heavily attenuating regions are substantially lower. The system noises, which do not decrease with the input exposure level, becomes more dominant and poses a limit to the minimum contrast level that can be detected or visualized. For example, in Fig. 5(b) for the sub-diaphragmatic region, the primary and scatter signal levels at the location 900, are ~46 and 22, respectively, in the slot-scan image and ~32 and 94 in the open field image acquired with an anti-scatter grid. We have measured the noise level to be 5.7 for a mean image signal of 307 and 1.1 for a mean image signal of 0.6 (zero exposure) with 120 kVp x-rays filtered with 0.5 mm copper only with the anti-scatter grid in place. By interpolation, we may estimate the noise level to be 2.1 and 3.0 in slot-scan and grid images, respectively. The CNR (evaluated from pixel values) required for an object varies with the object size. Assuming that a CNR of 3 is required to see a low contrast object, the minimum detectable contrast signal levels are ~6.3 and 8.9, corresponding to a contrast level of ~14% of the primary signals in the slot-scan image and 27% in the grid image, respectively. Both represent rather low contrast sensitivity. With SEDR, the primary signal level and the total noise level were raised to ~278 and 6.7 at the same location with increased exposure level. Thus, the minimum detectable signal level becomes ~19.9, representing a contrast level of ~7.2% (of the primary signal). This illustrates that although SEDR has resulted in minimum improvement over slot-scan imaging in reducing the SPRs, it could improve the contrast sensitivity by raising the primary signal levels in heavily attenuating regions. This advantage needs to be further investigated in greater details and with observer performance studies in our continuing investigation of the SEDR technique.

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Figure 1. System setup for acquiring spot collimated images with a commercial flat-panel based digital radiography system.

Figure 2.

Diagram showing a chest phantom radiograph used to determine the exposure equalization factors for SEDR imaging. The central area (in gray, covered lungs, mediastinum, and subdiaphragm) was selected to compute the mean exposure to the detector. The chest phantom image was divided into 24 (horizontally) by 40 (vertically) sub-regions. These sub-regions were aligned with those over which spot collimated images were acquired.

Figure 3. Superimposition of (a) weighted spot collimated images or (b) slot collimated images to simulate an SEDR or slot-scan images in lung regions, respectively, for a wider slot width. Images were shown in a film-like fashion for demonstration purpose.

Figure 4.

Locations for scatter and primary plus scatter were determined by the first-order derivative of the signal profile for the slot or spot simulated images.

Figure 5. The position-dependent primary intensities of the chest phantom images at slot edge for simulated slot-scan, SEDR, and full-field with anti-scatter grid plotted for the (a) lung and (b) subdiaphragm.

Figure 6.

The position-dependent scatter components of the chest phantom images at slot edge for simulated slot-scan, SEDR, and full-field with anti-scatter grid plotted for the (a) lung and (b) subdiaphragm.

Figure 7.

The SNRs of the chest phantom images at slot edge for simulated slot-scan, SEDR, and full-field with anti-scatter grid plotted across the (a) lung and (b) subdiaphragm.

Figure 8.

The SRFs for the chest phantom images at slot edge for simulated slot-scan, SEDR, and full-field with anti-scatter grid plotted across the (a) lung and (b) subdiaphragm.

Figure 9.
The CNRDFs of the chest phantom images at slot edge for simulated slot-scan, SEDR, and full-field with anti-scatter grid plotted across the (a) lung and (b) subdiaphragm.

Figure 10.

The CNRDFs of the chest phantom images at slot edge for simulated SEDR images with different scanning beam width plotted across the (a) lung and (b) subdiaphragm.