Comparison of scatter rejection and low-contrast performance of scan equalization digital radiography (SEDR), slot-scan digital radiography, and full-field digital radiography systems for chest phantom imaging

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Purpose: To investigate and compare the scatter rejection properties and low-contrast performance of the scan equalization digital radiography (SEDR) technique to the slot-scan and conventional full-field digital radiography techniques for chest imaging.

Methods: A prototype SEDR system was designed and constructed with an a-Se flat-panel (FP) detector to improve image quality in heavily attenuating regions of an anthropomorphic chest phantom. Slot-scanning geometry was used to reject scattered radiation without attenuating primary x rays. The readout scheme of the FP was modified to erase accumulated scatter signals prior to image readout. A 24-segment beam width modulator was developed to regulate x-ray exposures regionally and compensate for the low x-ray flux in heavily attenuating regions. To measure the scatter-to-primary ratios (SPRs), a 2 mm thick lead plate with a 2-D array of aperture holes was used to measure the primary signals, which were then subtracted from those obtained without the lead plate to determine scatter components. A 2-D array of aluminum beads (3 mm in diameter) was used as the low-contrast objects to measure the contrast ratios (CRs) and contrast-to-noise ratios (CNRs) for evaluating the low-contrast performance in chest phantom images. A set of two images acquired with the same techniques were subtracted from each other to measure the noise levels. SPRs, CRs, and CNRs of the SEDR images were measured in four anatomical regions of chest phantom images and compared to those of slot-scan images and full-field images acquired with and without antiscatter grid.

Results: The percentage reduction of SPR (percentage of SPRs reduced with scatter removal/rejection methods relative to that for nongrid full-field imaging) averaged over four anatomical regions was measured to be 80%, 83%, and 71% for SEDR, slot-scan, and full-field with grid, respectively. The average CR over four regions was found to improve over that for nongrid full-field imaging by 259%, 279%, and 145% for SEDR, slot-scan, and full-field with grid, respectively. The average CNR over four regions was found to improve over that for nongrid full-field imaging by 201% for SEDR as compared to 133% for the slot-scan technique and 14% for the antiscatter grid method.

Conclusions: Both SEDR and slot-scan techniques outperformed the antiscatter grid method used in standard full-field radiography. For imaging with the same effective exposure, the SEDR technique offers no advantage over the slot-scan method in terms of SPRs and CRs. However, it improves CNRs significantly, especially in heavily attenuating regions. The improvement of low-contrast performance may help improve the detection of the lung nodules or other abnormalities and may offer SEDR the potential for dose reduction in chest radiography. © 2011 American Association of Physicists in Medicine. [DOI: [10.1118/1.3519903](https://doi.org/10.1118/1.3519903)]

Key words: digital radiography, exposure equalization, scattered radiation, scatter rejection, flat-panel detector, contrast-to-noise ratio, low-contrast performance

I. INTRODUCTION

Despite recent advances in cross-sectional imaging, chest radiography is the most commonly performed diagnostic x-ray examination among all imaging procedures for initial diagnosis of many pulmonary diseases. In 2006, there were nearly 130×10^6 chest radiography examinations (account for about 44% imaging procedures) performed in the United States^{1,2} and the number of chest x-ray exams is continuously growing annually. Over the past two decades, the

progress in the development of advanced electronics and computer technology has led to the replacement of the conventional screen/film (SF) based radiography by flat-panel (FP) based digital radiography in many radiology departments.^{3–10} However, the nature of projection chest radiography along with the use of an area detector as image receptor indicates that some fundamental problems still exist in the digital imaging format as it was in the conventional SF based radiography.

First, the chest is one of the largest and thickest body

parts that are imaged on a routine basis, requiring large detector coverage (typically $35\text{ cm} \times 43\text{ cm}$) and use of high energy (typically 100–150 kVp) x rays to penetrate thicker body parts. Intrinsic to the use of a large area detector is the acceptance of a significant amount of scatter x-ray as a part of image signals. It has long been recognized that scattered radiation is the main cause for image quality degradation in terms of reduced image contrast and contrast-to-noise ratio (CNR). Scattered radiation accounts for about 70% (in the lungs) up to 95% (in the mediastinum and subdiaphragm regions) of total detected x-ray for chest radiograph acquired without an antiscatter grid.^{11–13} Various techniques have been implemented to reduce the scattered radiation during image acquisition.^{14–19} The antiscatter grid is by far the simplest and most effective method that has been successfully used in clinical procedures. Although the antiscatter grid is effective in reducing scattered x rays, it also attenuates the primary x rays by as high as 50%, resulting in a significant degradation of the detection efficiency for primary photons. As an alternative approach, the slot-scan technique has shown the promise to reduce the scattered radiation significantly while maintaining the primary signals unchanged. Thus, the slot-scan technique has the advantage of dose efficiency and can notably enhance the effective detection efficiency of imaging system.^{20–32}

Second, a large variation in x-ray opacity of anatomical structures in chest radiography imposes a fundamental limit on the visualization of subtle abnormalities in the heavily attenuating regions. The regional variation in x-ray transmission through the chest could be as high as two orders of magnitude in typical clinical procedures.¹³ The low-contrast performance is usually preserved in most lightly attenuating regions. However, the detectability of low-contrast objects is often degraded significantly due to combined effects of lower primary signals and higher scatter component detected in the most heavily attenuating regions. The problem is even more notable with increased patient population of obesity in the modern world. To overcome this problem, x rays incident to the patient may be regionally modulated to compensate for the variation of anatomical structure instead of evenly distributed in conventional radiography and hence achieve a uniform detector exposure. Technologies have been implemented to increase the exposures in the heavily attenuating regions and reduce the exposures in the lightly attenuating regions for applications in thoracic radiology and interventional radiology, with both scanning and area exposure equalization methods, all of which have demonstrated improved image quality and detectability of low-contrast objects in heavily attenuating regions.^{13,24,33–47}

We have developed a prototype scan equalization digital radiography (SEDR) system implemented with amorphous selenium (a-Se) based flat-panel detector. The system combines slot-scanning geometry with regional beam width modulation to reject scatter radiation effectively while regulating x-ray exposure regionally.⁴⁸ In this paper, we will present the scatter rejection/redistribution and low-contrast performances of the prototype SEDR technique measured with an anthropomorphic chest phantom and compare the

results to those measured for the antiscatter grid method and slot-scan technique. The main differences between the current work and the previous study⁴⁸ are as follows: (1) The SEDR system has been redesigned/built with a 24-segment beam width modulation assembly instead of a seven-segment module presented in the previous publication to improve the equalization accuracy and resolution for chest imaging; (2) the beam aperture array method³² was used to separate the primary intensities from the total image signals and to estimate scatter-to-primary ratios (SPRs) in various anatomical regions of the chest phantom image instead of measuring image lines across the chest phantom image with edge device technique; (3) the low-contrast performance was directly estimated by measuring the contrast signals and noise levels in this study instead of using primary signal-to-noise ratio (PSNR) to indirectly estimate the CNR improvement in the chest phantom image; (4) and, finally, the effects of effective exposures on the CNR were studied to examine the potential dose reduction for SEDR technique.

II. MATERIALS AND METHODS

II.A. Imaging system

The SEDR system investigated in this study is based on an a-Se direct conversion FP detector system (DirectRay, Hologic, Inc., Newark, DE). The active image area of the detector panel measures $35.6\text{ cm} \times 42.7\text{ cm}$ (14 in. $\times$ 17 in.) with 2560×3072 image elements. The pixel pitch of the detector is $139\text{ }\mu\text{m} \times 139\text{ }\mu\text{m}$. Charges generated during x-ray exposure (proportional to the energy deposited in the a-Se layer) are stored in each individual pixel and converted to voltage signals via front-end charge amplifiers during image readout. The image signals are then digitized into 14-bit image data via on-board analog-to-digital converters (ADCs) and transferred to a PC based detector controller. The detector controller performs image data correction (for dark current biases, gain nonuniform, and defective image pixels/lines) and stores calibrated image data temporarily. It also provides and monitors low and high voltage powers to the detector panel.

The FP detector was originally designed for “single shot” radiographic application. The readout scheme of the FP detector has been modified in our laboratory to allow “electronic aft-collimation” (referred to as the alternate line erasure and readout or ALER technique) to be achieved and to eliminate the need of using a bulky, heavy aft-collimator for slot-scan imaging.^{31,32} With the ALER technique, each image line is reset to erase the scatter signals accumulated prior to the arrival of the moving fan-beam and image signals are read out immediately after the fan-beam passes. To achieve electronic aft-collimation during the slot-scanning, the control signals from a specially designed controller board are used to alternatively reset the image line on the leading edge of the moving fan-beam and to read out the image line on the trailing edge of the moving fan-beam. The reset and readout operations are repeated over the entire detector as the fan-beam moves from top to bottom on the FP detector. In the conventional single shot full-field imaging operation, a short

FIG. 1. Bench top SEDR system and experimental setup. The system is built based on an a-Se flat-panel detector and slot-scanning geometry.

x-ray exposure is made to the entire detector panel followed by image readout; the x-ray exposure and the image readout are operated at different time periods. In the slot-scan imaging mode, the detector is exposed by a collimated fan-beam scanning across the object and synchronized with the speed of image readout in a line-by-line basis; both x-ray exposure and image readout are operated at the same time period. An image readout mode controller was custom designed and built to allow the detector system to operate in either the conventional single shot full-field imaging mode or the slot-scan imaging mode.³⁰

A three-phase, high-frequency x-ray generator (INDICO 100, CPI Canada Inc., Ontario, Canada) coupled to a dual-focal spot x-ray tube (G1593BI, Varian Medical Systems, Salt Lake City, UT) was used in this study. The x-ray tube has nominal focal spots of 0.3 and 1.2 mm and was operated in the large focal spot mode to help smooth out the edges of the beam width modulators in the SEDR images.^{36,45,47,49}

Figure 1 shows the bench top SEDR system and experimental set up. The source-to-image distance was set to be 183 cm (72 in.). A pair of 15 cm × 15 cm square tungsten plates (2 mm thick) was used to construct the fore-collimator and to limit the x-ray output to a narrow fan-beam for slot-scan or SEDR imaging. The fore-collimator was mounted on a computer-controlled high-precision linear drive (Velmex, Inc., Bloomfield, NY) at 40.6 cm (16 in.) from the x-ray focal spot to allow the horizontally oriented fan-beam to scan vertically. This setup leads to a projection magnification factor of approximately 4.5 for the fan-beam. The fore-collimator width was adjusted to be 0.36 cm, resulting in a 1.61 cm wide fan-beam on the image plane (equivalent to 116 image lines). Although the entire scan takes 4.4 s, each image line receives only a shorter exposure during the passage of the fan-beam. The effective exposure time was estimated to be approximately 0.2 s in slot-scan imaging. Exposure can be regulated by either changing the x-ray tube current or the exposure time (by adjusting the maximum slot width).

FIG. 2. (a) X-ray beam width modulation assembly used for SEDR imaging and (b) cross-sectional (coronal) view of the beam width modulation assembly.

Instead of keeping a fixed fore-collimator gap space in slot-scan imaging, a beam width modulation assembly was mounted onto the upper fore-collimator plate to modulate the integral image signal intensity by regulating the gap space regionally for SEDR imaging. The beam width modulation assembly consists of two identical modulation modules, each with twelve 3 mm thick lead (Pb) beam blockers that attenuate more than 99.99% of the x-ray photon flux at 120 kVp. The beam blockers were 5.4 mm wide and positioned at 10 mm (center-to-center) from each other along the slot, leaving a 4.6 mm wide gap between two neighboring blockers. The two modulation modules were mounted on the opposite sides of the upper fore-collimator plate (Fig. 2) with a 5 mm offset distance to achieve full coverage of the entire 120 mm slot. Each beam blocker covered and modulated a 24.3 mm wide segment on the image plane, with 0.9 mm overlap between

the two neighboring segments. Regular slot-scan imaging can be accomplished with all beam blockers lifted away from the x-ray fan-beam.

Twenty-four microlinear stage assemblies were used to drive the beam modulation blockers. Equalization signals were digitally generated from the prescan image acquisition and converted into analog signals using a 32-channel high-speed analog output PCI card (NI PCI-6723, National Instruments Corporation, Austin, TX) and then sent to the beam blocker drivers. The digital-to-analog converters in the card provided high-speed analog output with a data rate of 45 kS/s (kilo-samples per second) per channel, ± 10 V maximum output, and 13-bit (2.44 mV) data depth. The beam width modulation assembly was calibrated using “flat-field” images acquired with various modulation voltages. The beam modulation transfer functions were computed from the flat-field images and stored as a look up table in the image acquisition/processing workstation for SEDR image acquisition.⁴⁸

An Intel Pentium VI based computer (Precision 340, Dell Inc., Austin, TX) with a CPU frequency of 2.80 GHz and 512 MB RAM was used to control image acquisition, motion of fore-collimator (for both slot-scan and SEDR imaging), and beam width modulation.

II.B. Image acquisition of an anthropomorphic chest phantom

To quantitatively evaluate the scatter rejection/redistribution and low-contrast performance of the prototype SEDR system, an anthropomorphic chest phantom (Radiology Support Devices, Long Beach, CA) was imaged in the posterior-anterior positions. The chest phantom presented the attenuation and scatter properties similar to those of a typical medium-sized patient. The x-ray tube potential was set to 120 kVp as is used in clinical chest radiography. Exposures were made at 4 mA s for conventional full-field imaging. A removable stationary antiscatter grid with a 13:1 grid ratio and 78 lines/cm (Mitaya Mfg. Co., Ltd., Tokyo, Japan) was placed in front of the detector cover for the full-field imaging. This grid attenuates about 30% primary intensities at 120 kVp.³² The grid was removed from the x-ray path for slot-scan or SEDR imaging.

For slot-scan and SEDR imaging, the effective exposure time was estimated to be approximately 0.2 s with a fan-beam width of 1.61 cm on the image plane. With 20 mA tube current for slot-scanning, it yields an effective entrance skin exposure equivalent to that for conventional full-field imaging (4 mA s). For SEDR technique, a prescan chest phantom image was acquired at 4 mA (four times lower than that for slot-scanning or full-field imaging) using the slot-scanning to determine the exposure equalization pattern for subsequent SEDR imaging. The central 2916 vertical lines of the prescan chest phantom image were evenly divided into 18 segments, each with 22.5 mm wide and covered by a beam blocker following the correction of dark current background, gain nonuniformity, and defective pixels/lines. Image data were averaged horizontally line-by-line within each segment to

Fig. 3. Exposure equalization pattern computed from a low dose prescan chest phantom image. The scale shows the required exposure ratio to the lowest exposure.

yield 18 mean signal profiles in the beam scanning (vertical) direction. These signal profiles were further convolved with a low-pass filtering kernel (kernel size 10 was used in this study as the result of previous study⁴⁸ to smooth out their shape and reduce their fluctuations. The exposure equalization pattern used to regulate the entrance exposure for individual segment can be computed from the prescan image as follows:

$$E_{eq}(i,j) = k \cdot \frac{\overline{I_{pre}}}{I_{pre}(i,j)} \quad (1)$$

where the subscripts, “eq” and “pre” refer to equalization and prescan, respectively, i and j refer to segment number and row number, respectively, k is a weighting factor (constant), and $\overline{I_{pre}}$ is the average image signal over a selected central area covering the lungs, mediastinum, and subdiaphragm in the prescan image (air regions excluded). The exposure equalization factors should be smaller than 1 for lightly attenuating regions such as lungs, indicating exposure reduction, and greater than 1 for heavily attenuating regions, such as the mediastinum and subdiaphragm, indicating exposure increase to compensate heavy x-ray attenuation and to equalize exposures to the detector. The exposure equalization pattern computed for an anthropomorphic chest phantom is shown in Fig. 3.

The exposure equalization factors in different anatomical regions were then converted to the fan-beam width regionally in order to determine the exposure required for the consequent SEDR imaging. The beam width required for modulator i at location j is therefore determined by

$$w_{eq}(i,j) = \frac{i_{pre}}{i_{eq}} \cdot W \cdot E_{eq}(i,j) = k \cdot \frac{i_{pre}}{i_{eq}} \cdot W \cdot \frac{\overline{I_{pre}}}{I_{pre}(i,j)}, \quad (2)$$

where W is the maximum beam width determined by the fore-collimator gap (was set to be 0.36 cm in this study) and i_{eq} and i_{pre} are the x-ray tube current settings for SEDR and prescan (slot-scan) imaging, respectively. The beam width w_{eq} should be equal or smaller than the fore-collimation

width W . The effective exposure for SEDR imaging can be changed by varying the weighting factor k or the tube current i_{eq} . The tube current settings for radiography are usually discrete instead of continuous in commercial x-ray systems. The selection of i_{eq} should be the lowest available tube current that is equal or greater than the product of i_{pre} and maximum E_{eq} . The weighting factor k was set to be 5 (equal to the ratio of tube current for slot-scan imaging over that for SEDR prescan) in order to achieve an effective exposure of SEDR equivalent to that of 20 mA slot-scan imaging (or 4 mA s full-field imaging). Based on the calculation of maximum E_{eq} , the tube current for SEDR imaging was determined to be 100 mA in this study. The prescan added 20% “extra” radiation dose in addition to the SEDR imaging.

In the conventional single shot full-field imaging mode, the raw image is corrected for the gain nonuniformity, dark current offset, and defective pixels/lines internally in the detector controller. However, the correction was done externally in the acquisition workstation since the detector read-out scheme was altered to the slot-scan or SEDR imaging mode. The process to correct for gain nonuniformity, dark current offset, and dead pixels/lines externally was done in a similar manner as was done internally by the detector controller.³² The “image correction” function of the detector controller was disabled so that only the uncorrected raw image data were transferred and saved in the acquisition workstation for the slot-scan or SEDR imaging mode.

II.C. Assessment of primary and scattered radiation

To assess the magnitudes of primary and scatter components, the beam aperture array method was used to separate the primary signal from the total image signal in the aperture areas. The aperture mask was constructed on a large piece of 2 mm thick lead (Pb) sheet laminated with a 4 mm thick acrylic sheet. A 2-D array of apertures (2 mm in diameter and spaced center-to-center by 2.54 cm vertically and horizontally) were opened through both lead and acrylic sheets. The combination of the lead and acrylic sheet attenuates about 99.9% of the x rays at 120 kVp, thus blocking nearly all x rays in the area outside the aperture holes.

With the aperture mask in place, the image signals inside the apertures are mostly from the primary radiation. Areas outside the apertures are blocked by the lead/acrylic sheet and measured nearly zero image intensities; however, small image signals are expected in the regions immediately surrounding the aperture areas where x rays scattered from volume irradiated by the pencil beams through the aperture holes. The image acquired without the aperture mask in place contains both the primary and scatter components. Thus, the scatter components in the aperture areas may be estimated by subtracting the two images acquired without and with the aperture mask in place as follows:

$$I_S(i, j) = I_{P+S}(i, j) - I_P(i, j) \approx I_{P+S}(i, j) - I_{Aperture}(i, j), \quad (3)$$

where $I_P(i, j)$ is the mean primary signal and $I_{P+S}(i, j)$ is the mean total (primary plus scatter) signal measured at the location where the center of aperture (i, j) is, for images ac-

FIG. 4. Synthetic digital image of an anthropomorphic chest phantom and the associated locations of the beam apertures where the measurements were made. The results were averaged and compared for four regions (highlighted): (1) lungs, (2) mediastinum, (3) retrocardium, and (4) subdiaphragm. Reprinted with permission from X. Liu, C. C. Shaw, C. Lai, M. C. Altunbas, L. Chen, T. Han, and T. Wang, “Scatter rejection and low-contrast performance of a slot-scan digital chest radiography system with electronic aft-collimation: A chest phantom study,” *Med Phys.* **35**, 2391–2402. Copyright 2008, American Association of Physicists in Medicine (AAPM).

quired without aperture mask in place. $I_{Aperture}(i, j)$ is the mean signal measured with aperture mask in place and is an approximation for $I_P(i, j)$. Because $I_{Aperture}(i, j)$ still contains a small scatter component resulted from aperture volume irradiated by the narrow pencil beam inside the aperture and the crosstalk from neighboring apertures, the approximation resulted in an overestimation of primary signals by about 3.2% in this study (2% from the in-aperture scatter and 1.2% from the crosstalk). The SPR is computed as follows:

$$SPR(i, j) = \frac{I_S(i, j)}{I_P(i, j)} \approx \frac{I_{P+S}(i, j) - I_{Aperture}(i, j)}{I_{Aperture}(i, j)}. \quad (4)$$

To quantify the scatter rejection ability, the SPR reduction ratio (SPRRR) is defined and computed as follows:

$$SPRRR(i, j) = \frac{SPR^{ff}(i, j) - SPR(i, j)}{SPR^{ff}(i, j)}, \quad (5)$$

where $SPR^{ff}(i, j)$ is the SPR in full-field, nongrid imaging.

Mean signals were computed over a 5 pixels $\times$ 5 pixels square region at the center of the aperture for images acquired without and with the aperture mask in place. The shape and size of the aperture and the measurement region were selected to reduce the error in primary signal measurement to a reasonable level due to focal spot blurring and geometric cutoff by the aperture mask.³² The aperture mask was placed at 156.2 cm (61.5 in.) from the x-ray focal spot on the tube side of the chest phantom.

Figure 4 shows a synthetic digital image to illustrate the anthropomorphic chest phantom and the locations of the beam apertures. Four different anatomical regions were selected for this study (highlighted): (1) lungs, (2) mediastinum, (3) retrocardium, and (4) subdiaphragm. SPRs and

SPRRRs were measured for apertures within each region and averaged for comparison. Aperture areas outside these regions were excluded from our analysis.

II.D. Assessment of CR and CNR

The contrast signal (C) is defined as the signal difference between a contrast object and the area surrounding it (background), while the contrast ratio (CR) is defined as the ratio of the contrast to the background signal as follows:

$$C(i,j) = I_b(i,j) - I_o(i,j), \quad (6)$$

$$CR(i,j) = \frac{I_b(i,j) - I_o(i,j)}{I_b(i,j)}, \quad (7)$$

where $I_o(i,j)$ and $I_b(i,j)$ are the mean image signals measured inside the object area and the surrounding background area, respectively. An immediate effect of scatter reduction is an increase of the image CR by reducing the scatter components in both the object area and the background area with the contrast signal C unchanged. However, the CR can be freely manipulated by varying the window setting in digital image. Thus, the CR alone by itself is insufficient for characterizing the low-contrast performance. Furthermore, the noise level, if too high, may prevent the low-contrast object from being detected or visualized. Thus, the CNR, defined as the ratio of the contrast signal to the noise level, is often used as an indicator for the low-contrast performance. It is defined and computed as follows:

$$CNR(i,j) = \frac{I_b(i,j) - I_o(i,j)}{\sigma_b(i,j)}, \quad (8)$$

where $\sigma_b(i,j)$ is the root-mean-square noise level measured in the surrounding area of the contrast object. To measure and compare the CNRs of different imaging techniques, a 2-D array of 3 mm diameter aluminum beads, spaced by 2.54 cm center-to-center in both vertical and horizontal directions, were overlaid with the chest phantom and used as the contrast objects. The aluminum beads were attached to a thin clear film and placed posterior to the chest phantom at the same location where the aperture mask was placed for primary and scatter measurement. The positions of the aluminum beads were aligned with those of the apertures so that the CNRs were measured at the same positions as for the primary signals, scatter components, and SPRs.

Two sets of chest phantom images were acquired with identical exposure settings (without the aperture mask or aluminum beads) and subtracted from each other to remove the anatomical structures for noise measurement. Normalization was performed prior to subtraction to eliminate signal variations from potential exposure fluctuation which may result in incomplete cancellation of the anatomical structures. Following the subtraction, standard deviations were computed and divided by $\sqrt{2}$ to estimate the noise levels near the surrounding areas of the low-contrast objects. As was done with the SPRs and SPRRRs, CRs, CNRs, and their improvement factors were measured and averaged for apertures within four different anatomical regions for comparison.

(a)

(b)

FIG. 5. (a) SPRs measured in the lungs, mediastinum, retrocardium, and subdiaphragm for the full-field without/with grid, slot-scan, and SEDR methods with the same beam quality and entrance exposure. (b) SPRRRs measured in the lungs, mediastinum, retrocardium, and subdiaphragm for the full-field with grid, slot-scan, and SEDR methods with the same beam quality and entrance exposure.

III. RESULTS

Figure 5(a) shows the SPRs measured for the four different imaging techniques: Full-field without grid, full-field with grid, slot-scan, and SEDR. All three scatter rejection techniques were effective in reducing the SPRs in all four anatomical regions. However, the SPR reduction was more pronounced in the heavily attenuating mediastinum, retrocardium, and subdiaphragm regions as the SPRs were originally higher in these regions. The slot-scan and SEDR imaging techniques, both based on the slot-scanning geometry, appeared to be more effective than the antiscatter grid used in this study. In fact, the scatter rejection performance of slot-scan imaging varies with the slot width used; further scatter rejection may be achieved by using a narrower slot width or a narrower maximum slot width for SEDR. Figure 5(a) also shows that both slot-scan and SEDR imaging techniques resulted in similar SPRs in heavily attenuating regions, except

that the SPRs appeared to be lower for slot-scan than for SEDR in lungs. This may be due to the increased exposures in mediastinum and retrocardium (by making the beam wider in these regions), which resulted in higher scatter contribution to the lungs where beam blockers opened relatively small to reduce exposures in the lungs. Figure 5(b) shows the SPRRRs measured for the three different imaging techniques: Full-field with grid, slot-scan, and SEDR. The SPRRRs tended to be higher in the heavily attenuating regions. However, there were no significant differences (1.7% on average) between the SEDR and the slot-scan imaging techniques. The SPRRR was about 8% lower for SEDR than the slot-scan in the lungs, indicating a moderate increasing in SPRs as a result of higher exposures in mediastinum and retrocardium. The SEDR technique resulted in an average reduction of SPRs by 73% (lungs), 82% (mediastinum), 83% (retrocardium), and 84% (subdiaphragm) versus 65% (lungs), 73% (mediastinum), 74% (retrocardium), and 73% (subdiaphragm) with the antiscatter grid method. The average (over four regions) SPRRRs were estimated to be 80%, 83%, and 71% for SEDR, slot-scan, and antiscatter grid techniques, respectively.

Figure 6(a) shows the CRs measured for the four different imaging techniques: Full-field without grid, full-field with grid, slot-scan, and SEDR. The CRs were generally higher in the lungs than in heavily attenuating regions and decreased in the order of the lungs, mediastinum, subdiaphragm, and retrocardium. Both slot-scan and SEDR demonstrated significantly higher CRs than antiscatter grid in all regions. In Fig. 6(b), the CR improvement factors, defined as the ratio of the improved CRs to those for full-field, nongrid imaging are shown and found to increase with the region in the order of the lungs, mediastinum, retrocardium, and subdiaphragm. There were no significant differences in the CR improvement factors between SEDR and slot-scan imaging, except in the subdiaphragm where the improvement factor for SEDR was 0.98 lower than that for slot-scan. The SEDR technique resulted in an average improvement of CRs by 96% (lungs), 241% (mediastinum), 349% (retrocardium), and 350% (subdiaphragm) versus 41% (lungs), 137% (mediastinum), 178% (retrocardium), and 223% (subdiaphragm) with the antiscatter grid method. Averaged over all four regions, the CRs were found to improve by 259%, 279%, and 145% for SEDR, slot-scan, and antiscatter grid techniques, respectively.

In Fig. 7(a), the CNRs are plotted and compared for the four different imaging techniques. It shows that the CNRs decreased in the order of the lungs, mediastinum, retrocardium, and subdiaphragm except that those for SEDR were about the same over the retrocardiac and subdiaphragmatic regions. All scatter rejection techniques improved the CNR, except in the lungs where the use of an antiscatter grid degraded the CNRs slightly in full-field imaging. In Fig. 7(b), the CNR improvement factors, defined as the ratios of the improved CNRs to those for the full-field, nongrid imaging technique, are plotted and listed for comparison. Figure 7(b) shows that the improvements of CNRs are more pronounced in the heavily attenuating regions than in the lungs.

(a)

(b)

FIG. 6. (a) CRs measured in the lungs, mediastinum, retrocardium, and subdiaphragm for the full-field without/with grid, slot-scan, and SEDR methods with the same beam quality and entrance exposure. (b) CR improvement factors (over nongrid full-field imaging) measured in the lungs, mediastinum, retrocardium, and subdiaphragm for the full-field with grid, slot-scan, and SEDR methods with the same beam quality and entrance exposure.

The slot-scanning geometry do improve CNRs significantly in all regions for both slot-scan and SEDR imaging methods; however, the SEDR resulted in a substantial improvement of CNRs in heavily attenuating regions due to significantly increased primary signals and SNRs from the compensation of x-ray beam intensities in these regions. The CNR improvement factor for SEDR was slightly better than that for slot-scan in the lungs. In contrast, antiscatter grid improved the CNRs only minimally in heavily attenuating regions. The SEDR technique resulted in an average improvement of CNRs by 109% (lungs), 204% (mediastinum), 198% (retrocardium), and 291% (subdiaphragm) versus -11% (lungs), 15% (mediastinum), 15% (retrocardium), and 38% (subdiaphragm) with the antiscatter grid method. The CNRs averaged over all regions were found to improve by 201%, 133%, and 14% for SEDR, slot-scan, and antiscatter grid techniques, respectively.

(a)

(b)

FIG. 7. (a) CNRs measured in the lungs, mediastinum, retrocardium, and subdiaphragm for the full-field without/with grid, slot-scan, and SEDR methods with the same beam quality and entrance exposure. (b) CNR improvement factors (over nongrid full-field imaging) measured in the lungs, mediastinum, retrocardium, and subdiaphragm for the full-field with grid, slot-scan, and SEDR methods with the same beam quality and entrance exposure.

In Figs. 8(a) and 8(b), the CNRs and CNR improvement factors (over full-field with antiscatter grid) for SEDR images acquired with various reduced tube current settings are plotted and compared. The current was set at 32, 50, and 80 mA while keeping the beam modulation pattern unchanged, resulting in an effective exposure of approximately 32%, 50%, and 80% of that 4 mA s single shot full-field imaging. The results showed that both the CNRs and CNR improvement factors increased with the effective exposure. With a tube current of 32 mA, CNRs and CNR improvement factors were higher than those for full-field imaging (4 mA s) with grid in all anatomical regions. For SEDR image acquired with a tube current of 80 mA, the CNRs and CNR improvement factors were higher in the heavily attenuating regions but slightly lower in the lungs as compared to those of slot-scan image. The CNRs averaged over all regions were found

(a)

(b)

FIG. 8. (a) CNRs and the (b) CNR improvement factors (over full-field imaging with grid) measured in the lungs, mediastinum, retrocardium, and subdiaphragm for the SEDR images with effective entrance exposures of 32%, 50%, 70%, and 100% to that of the full-field imaging technique.

to improve by 24%, 52%, and 133% over the full-field imaging with grid for SEDR images acquired with effective exposures of 32%, 50%, and 80%, respectively. The SPRs and the CRs and their improvement were not plotted here as they did not change with the mA settings when beam modulation pattern was unchanged.

A chest phantom image with low-contrast objects (Al beads) and a region-of-interest (ROI) was shown in Fig. 9(a). The zoomed-in ROI at retrocardium was illustrated with different imaging techniques: Fig. 9(b), nongrid full-field; Fig. 9(c), full-field with anti-scatter grid; Fig. 9(d), slot-scanning; and Fig. 9(e), SEDR. All scatter removal/rejection techniques demonstrated significantly improved visualization of low-contrast objects. The SEDR exhibited the lowest noise level and the highest clarity of details as a direct result of exposure compensation in the heavily attenuating region compare to antiscatter grid and slot-scanning methods.

FIG. 9. (a) Chest phantom image with low-contrast objects (Al beads); the zoomed-in ROI in retrocardium imaged with different techniques: (b) Nongrid full-field; (c) full-field with anti-scatter grid; (d) slot-scanning; and (e) SEDR.

IV. CONCLUSIONS AND DISCUSSION

The purpose of this study was to investigate and demonstrate the scatter rejection and the low-contrast performance of a SEDR system based on an a-Se FP detector and the slot-scanning geometry for chest imaging. SPRs, CRs, and CNRs of the SEDR technique were measured with an anthropomorphic chest phantom and compared to those of full-field imaging techniques used without/with an antiscatter grid and slot-scan method. It was shown that with the same effective exposure, SEDR achieved similar scatter rejection ability as slot-scan imaging in the heavily attenuating regions. However, the scatter rejection ability was slightly degraded with SEDR technique compared to the slot-scan imaging in the lungs. This was due to greater scatter contributions from the mediastinum and retrocardium resulted from increased exposure compensation in these regions. The overall scatter rejection ability of SEDR was found better than antiscatter grid method used in the conventional single shot full-field imaging.

As a part of the SEDR imaging process, a low dose prescan chest phantom image has to be acquired to determine the exposure equalization pattern prior to the SEDR image acquisition. Hence, the prescan (acquired using slot-scan technique to minimize scatter components) contributes to the total effective exposure in addition to SEDR image acquisition itself. It is equally important to consider the total effective exposure instead of effective exposure for SEDR imaging only, while conducting the low-contrast performances between different imaging techniques, even though the prescan image would not be used for assessment of low-contrast performance. In this study, the prescan chest phantom image was acquired at an effective exposure of 20% of that for the full-field imaging. By combining the exposures from both prescan and SEDR, the total effective exposures were approximately 52%, 70%, and 100% of that from 4 mA s full-field imaging for the SEDR images acquired at x-ray tube current of 32, 50, and 80 mA, respectively. Compared to the antiscatter grid method, the SEDR image acquired at tube current of 32 mA (yields total effective exposure 52% of that

with full-field imaging) achieved moderately higher CNRs and CNR improvement factors in all anatomical regions. This achievement may be attributed to the dose saving advantage due to the slot-scanning geometry used in SEDR imaging, while the antiscatter grid attenuated approximately 30% of primary x rays from reaching the detector plane. The overall CNR (average over all four regions) was found to improve by 40%, compared with 14% for anti-scatter grid method.

For the SEDR image acquired at a tube current of 80 mA (yields total effective exposure as same as that with full-field imaging), the CNRs and CNR improvement factors were higher than those of slot-scan imaging in the heavily attenuating regions but slightly lower in the lungs. The differences in CNR improvement ratio between the SEDR and slot-scan techniques were found to be -3%, 47%, 36%, and 69% in the lungs, mediastinum, retrocardium, and subdiaphragm, respectively. However, the differences in CNR improvement ratio between SEDR technique and antiscatter grid method were found to be 95%, 167%, 154%, and 207% in the lungs, mediastinum, retrocardium, and subdiaphragm, respectively. The overall average CNRs were found to improve by 40%, 74%, and 170% for SEDR imaging with total effective exposures of 52%, 70%, and 100%, respectively, while the CNR improvement ratio was found 14% only for the antiscatter grid method and 133% for slot-scan technique. The CNRs can be further improved by reducing the prescan exposure and increasing the SEDR exposure while keeping the total effective exposure unchanged if a modern commercial FP detector is available (due to improved detector sensitivity and detective quantum efficiency) as discussed previously.⁴⁸

It should be noted that the results from this study applies only to the specific antiscatter grid used for the conventional full-field radiography and the fore-collimator slot width (1.61 cm) used for the slot-scan and SEDR imaging techniques. Any change made to the fore-collimator slot width will affect the scatter rejection ability and low-contrast performance for both the shot-scan and SEDR imaging techniques. A narrower slot width results in lower SPRs, higher

CRs, and higher CNRs for both slot-scan and SEDR imaging; however, it requires higher tube current (to maintain sufficient x-ray exposure) and higher sensitivity beam modulation (increasing the complexities of beam modulation modules). Therefore, there is a trade-off in selecting imaging parameters between the slot width (or readout speed) and the tube loading. The use of a high density antiscatter grid may result in better scatter rejection ability and low-contrast performance for conventional single shot full-field radiography as well.

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