Scatter rejection and low-contrast performance of a slot-scan digital chest radiography system with electronic aft-collimation: A chest phantom study

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Anti-scatter grids have been widely used to reject scatter and increase the perceptibility of low-contrast object in chest radiography; however they also attenuate the primary x-rays, resulting in a substantial degradation of primary information. Compensation for this degradation requires the use of higher exposure technique hence higher dose to the patient. A more efficient approach to reject scatter is the slot-scan imaging technique which employs a narrow scanning x-ray fan beam in conjunction with a slit or slot shaped solid state detector or an area detector used with an aft-collimator. With this approach, scatter can be rejected effectively without the need to attenuate primary x-rays. This paper demonstrates an electronic aft-collimation method, referred to as the alternate line erasure and readout (ALER) technique, for implementing the slot-scan digital radiography with a modern flat-panel detector. With this technique, instead of first exposing the detector and then reading the image line by line, the image line on the leading edge of the scanning fan beam is reset to erase the scatter accumulated prior to the arrival of the fan beam x-rays, while the image line on the trailing edge of the scanning fan beam is read out to acquire the image signals following the fan-beam exposure. These reset and readout processes are alternated and repeated as the x-ray fan beam scans across the detector. An anthropomorphic chest phantom was imaged to evaluate the scatter rejection ability and the low-contrast performance for the ALER technique and compare them with those for the anti-scatter grid method in full-field chest imaging. With a projected beam width of 16 mm, the slot-scan/ALER technique resulted in an average reduction of the scatter-to-primary ratios by 81%, 84%, 82%, and 86% versus 65%, 73%, 74%, and 73% with the anti-scatter grid method in the lungs, mediastinum, retrocardium, and subdiaphragm, respectively. The average CNR for the slot-scan/ALER technique was found to improve by 135%, 133%, 176%, and 87% versus 15%, 15%, 38%, and -11% with the anti-scatter grid method in the mediastinum, retrocardium, subdiaphragm, and lungs, respectively. These results demonstrated that the slot-scan/ALER technique can be used to achieve equally effective scatter rejection but substantially higher low-contrast performance than the anti-scatter grid method. © 2008 American Association of Physicists in Medicine. [DOI: [10.1118/1.2921132](https://doi.org/10.1118/1.2921132)]

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I. INTRODUCTION

Over the past two decades, various digital radiography techniques have been developed to provide images in digital format for better utilization and management of the images.¹⁻¹⁷ While many improvements have been made on the detector technology, image acquisition methods, and image display system, less attention has been paid to the long recognized problem of scattered x-rays in projection imaging.

Intrinsic to the use of an area detector as the x-ray receptor is the acceptance of scattered x-rays as part of the image signals. This has long been recognized as a main cause for degraded image quality in projection imaging as it decreases both the image contrast and contrast-to-noise ratio (CNR) for low-contrast objects in the image.¹⁸⁻²¹ Various techniques have been implemented and investigated to reject scattered

radiation during image acquisition.²²⁻²⁴ Due to its simplicity, the anti-scatter grid method has by far been the most widely used technique in radiographic and fluoroscopic procedures with various area detectors, including the screen-film (S/F) combinations, image intensifier-video camera chains, storage phosphors or more recently flat-panel detectors. Although the anti-scatter grid is effective in attenuating scattered x-rays, it also attenuates the primary x-rays by as high as 30%–50%, resulting in a significant degradation of the attenuation information carried by them. To compensate for the attenuation of primary x-rays, higher exposure techniques are often used, resulting in increased patient dose.

To overcome the shortcomings of the anti-scatter grid methods, various slot-scan imaging techniques have been explored and investigated to achieve effective scatter rejection without having to attenuate the primary x-rays. With these

techniques, one or more collimated narrow fan beams were formed and used to scan across the patient.^{25–36} A dynamically collimated area detector or a moving detector array is used to record the fan-beam exposure signals only, leaving x-rays scattered from outside the fan beam undetected. The main advantage of the slot-scan imaging technique is its ability to achieve effective scatter rejection without having to attenuate primary x-rays. Early attempts in implementing the slot-scan imaging technique relied on the use of a S/F combination in conjunction with an aft-collimator. While prototypes have been built to demonstrate the feasibility of this approach, the construction and need to control and move the heavy and bulky aft-collimator, especially for chest imaging, have turned to be an obstacle in continuing to pursue this approach.^{27–29} With the advances of digital detector technology, slot-scan imaging systems using slit or slot shaped detector arrays have been developed and investigated.^{32–36} This approach has enjoyed the advantage of not having to control and move a heavy and bulky aft-collimator. However, it required specialized detector arrays to be built and moved, calling for careful alignment and synchronization with the scanning x-ray fan beam.

We have developed and demonstrated an electronic aft-collimation method, referred to as the alternate line erasure and readout (ALER) technique, to be used with a modern flat-panel detector for implementing slot-scan imaging.³⁷ With this technique, the detector is electronically configured to track the scanning x-ray fan beam and integrate the fan beam exposures. Image lines are reset immediately before the fan beam arrives to erase the scatter signals accumulated. This way, image signals read out immediately after the fan beam passes an image line do not contain scatter signals accumulated before and after the fan beam passes. In this paper, we will present the scatter rejection and low-contrast performances of the ALER technique measured with an anthropomorphic chest phantom and compare the results with those measured for the anti-scatter grid method and slot-scan imaging used in conjunction with an anti-scatter grid. The main differences between the current study and the previously published study are as follows:

- (1) We carefully assembled and optimized a prototype system that consists of many specially selected or fabricated components in this work. The x-ray system used is more modern and powerful with a smaller focal spot in order to minimize the focal spot blurring in slot-scan imaging.
- (2) A more accurate approach, aperture matrix method was used to estimate the scattered radiation and scatter-to-primary ratios (SPRs) in various anatomic regions instead of using a tungsten bar to measure the scattered radiation and SPRs for one image line.
- (3) Actual contrast signals and noise levels were measured to evaluate the low-contrast performance in this study while a theoretical model was developed to estimate how the “CNRs” would be improved by slot-scan technique based on the measurement from SPRs in the previous work.
- (4) We used the ALER technique with the fore collimator removed to obtain the full-field image to evaluate the effectiveness of scatter rejection of the ALER technique in the previous work. The method is adequate for measuring and comparing the signal levels and the SPRs but inadequate for measuring and comparing the noise levels or CNRs. In this work, we used the regular full-field acquisition technique to obtain the full-field image. This approach allows the noise properties to be measured and compared.
- (5) In the previous work, we compared the SPRs of a slot-scan image with a full-field image acquired without an anti-scatter grid only. However, images are usually acquired with a grid in clinical applications. Therefore we included full-field images acquired with a grid for comparison in this study. The comparison is thus a more relevant one.

II. MATERIALS AND METHODS

II.A. Imaging system

The slot-scan digital radiography system investigated in this study is based on an amorphous silicon/amorphous selenium (*a*-Si/*a*-Se) direct conversion flat-panel (FP) detector (DirectRay, Hologic, Inc., Newark, DE). The 35.6 cm × 42.7 cm (14 in. × 17 in.) active area of the detector is divided into a 2560 × 3072 array of image elements with a pixel size of 139 μm × 139 μm. The image readout signals are digitized into 14 bits image data via on-board analog-to-digital converters (ADCs) and transferred to a PC based detector controller. The controller, in addition to temporarily storing the image data readout, also performs a number of other functions, including monitoring and supplying power to the detector, correcting for dark current biases, nonuniform gain, and bad pixels using stored calibration data. The image readout controller was custom designed and built in our laboratory to allow the detector to be operated in either the regular full-field or slot-scan imaging modes. In the regular full-field imaging mode, the entire detector is exposed and then read out. In the slot-scan imaging mode, the detector is exposed by a scanning fan beam with the image lines read out in synchronization with the fan-beam motion. An Intel Pentium 4 based computer (Precision 340, Dell Inc., Austin, TX) with a CPU frequency of 2.80 GHz and 512 MB RAM is used to control the fan-beam motion and to acquire image data during slot-scan imaging. It is also used to store, process and display the resulting slot-scan images.

FP detectors are typically oriented to provide a field-of-view in the “portrait” mode for clinical chest radiography. However, the FP detector used in our investigation was oriented in the “landscape” instead of portrait mode because the TFT gate-line driven electronics are laid along the two short sides of the panel and the image lines are parallel to the long sides. With the landscape mode, the image lines are horizontal and they can be gated and read out in synchronization with the vertical scanning motion of the horizontally oriented fan beam during slot-scan imaging.

FIG. 1. Conventional line-by-line readout following x-ray exposure (a, b) for a commercial flat-panel digital radiography system. Alternate line erasure and readout (ALER) technique (c, d) for slot-scan digital radiography implemented with a flat-panel detector that alternatively reset the leading edge line and read out the trailing edge line in synchronization with the scanning fan-beam x-ray. (Reprinted from: X. Liu, C. C. Shaw, M. C. Altunbas, and T-P. Wang, "An Alternate Line Erasure and Readout (ALER) Method for Implementing Slot-Scan Imaging Technique with a Flat-Panel Detector—Initial Experiences," IEEE Trans. Med. Imaging **25**, 496–502. © 2006 IEEE, with permission from IEEE).

The x-ray system consists of a three phase, high frequency generator (INDICO 100, CPI Canada Inc., Ontario, Canada) coupled with a dual-focal spot x-ray tube (G1593BI, Varian Medical Systems, Salt Lake City, UT). The tube has a nominal focal spot size of 0.3 mm for the small focus and 1.2 mm for the large one. In this study, the tube was operated in the small focal spot mode to minimize focal spot blurring of the fan-beam edges.

II.B. Scatter erasure and image readout

Slot-scan imaging with an area detector could be implemented with a physical aft-collimator which tracks and allows the projected fan beam to scan across the detector while shielding the nonfan-beam area from scattered radiation. Such a collimator is bulky, heavy, and mechanically difficult to control and move at the high speed required. With a FP detector, an "electronic collimation" method can be used to eliminate the need of using a bulky, heavy aft-collimator. We have developed such a method, referred to as the ALER technique, to incorporate electronic collimation with a FP detector system for slot-scan imaging. With the ALER technique, each image line is reset to erase the scatter signals accumulated prior to the arrival of the fan beam and read out immediately after the fan beam passes. During slot-scan imaging, control signals from a specially designed controller board are used to alternately reset the image line on the leading edge of the fan-beam and to read out the image line on the trailing edge. These two operations are repeated over the entire de-

tor as the fan beam moves from top to bottom. Figure 1 shows the two different readout schemes for regular full-field imaging and slot-scan imaging with the ALER technique.

In slot-scan imaging, a scanning fan beam is used to expose the detector. The width of the fan beam on the detector is determined by the fore-slit width and the magnification factor for the slit to be projected onto the detector. As the fan beam completely passes by line 1, the exposure comes to an end and the signals from line 1 are read out. Assume that the fan beam covers n image lines, the fan beam is about to begin its exposure to line $n+1$. However, line $n+1$ has received exposure from x-rays scattered from the fan beam before the fan beam reaches it. To erase the accumulated scatter signals, line $n+1$ needs to be reset before the fan-beam exposure signals begin to accumulate. As the fan beam moves on line by line, this readout and resetting cycle is repeated over and over again for all lines in synchronization to the projected scanning fan beam until the signals from the last image line are read out.

The ALER technique was implemented by replacing the regular gate line control signals with signals from a specially designed ALER controller board. The board contains electronic switches, high frequency pulse generator, pulse sequence encoder, and decoder, etc. The ALER control signals are multiplexed with the regular gate line pulses so that the FP detector may be operated in either the regular full-field or slot-scan imaging mode. This allows us to compare the properties of slot-scan and full-field imaging using the same de-

FIG. 2. Image signal intensity plotted as a function of exposure (mAs) for images acquired with and without anti-scatter grid. Exposures made at 120 kVp with a 2 cm aluminum block placed at the x-ray tube output to simulate the x-ray spectrum and attenuation of a patient.

tector. For conventional full-field imaging, the internal gate-line control pulses are selected and used to operate the FP in the regular image readout mode. For slot-scan imaging, the ALER gate-line control pulses are selected and used to alternately erase the leading edge image line and read out the trailing edge image line as the fan beam scans across the detector. The resetting of an image line for scatter erasure can be done within a very short time (~ 1.5 ms) as compared to the regular readout time (4.4 S). Thus, the ALER erasure/readout scheme can be implemented to operate at the regular readout rate.

II.C. Image acquisition

A stationary anti-scatter grid with a grid ratio of 13:1 and a density of 78 lines/cm (Mitaya MFG. Co., Ltd., Tokyo, Japan) was inserted in the x-ray path and placed in front of the FP detector (close to the detector cover) to acquire full-field images for comparison with the slot-scan images. It was also used in conjunction with slot-scan imaging to see if it can further improve scatter rejection and image quality. The linear response of the FP detector system was first confirmed by acquiring a series of uniformly exposed images at various mAs settings while keeping the kVp fixed and then measuring the mean signals as a function of the mAs over a 100×100 region of interest (ROI) at the center of the image. A 2-cm-thick aluminum plate was added to the x-ray tube output window to simulate the attenuation and filtration by a patient. The linearity of the image signals in the exposure is demonstrated by the plotted results in Fig. 2. The plots also show that the anti-scatter grid attenuated about 30% of primary signals for the FP detector used in our study.

A computer-controlled scanning fore-collimator system was mounted at the output of the x-ray tube to generate the scanning x-ray fan beam. The system consists of a slot collimator mounted on a high-precision linear drive (Velmex,

Inc., Bloomfield, NY) to generate a horizontal fan-beam scanning in the vertical direction and to synchronize with the moving of electronic aft-collimation of the FP detector. The fore collimator was constructed from a pair of 2-mm-thick tungsten plates that attenuate nearly all x-rays at 120 kVp. To minimize the focal spot blurring effects, the tungsten edges were cut with a relief angle of 10° on each side of the collimator. The source-to-image distance (SID) was kept at 183 cm (72 in.) while the fore collimator was placed at ~ 40.6 cm (16 in.) from the focal spot resulting in a magnification factor of ~ 4.5 . The fore-collimator width was selected to be 0.358 cm, resulting in a fan-beam width of 1.61 cm in the image plane (equivalent to 116 image lines). These settings resulted in an effective exposure time of approximately 0.2 S for slot-scan imaging. The penumbra effect caused by the finite focal spot size and the edges of the fore collimator in this study was measured to be about eight image lines on both leading and trailing edges of the slot, leading to a drop of integral signal by about 1.7% only.

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. Exposures were made at 120 kVp and 4 mAs for full-field imaging while those for slot-scan imaging were 120 kVp and 20 mA (equivalent to an effective exposure of 4 mAs). These settings are comparable to typical techniques used in clinical chest radiography.

Raw image data acquired from a digital detector are subject to calibration to correct for gain nonuniformities, defective pixels, and dark current noise (offset). Image corrections were performed internally with preloaded gain and dark current noise maps for full-field imaging. Two sets of gain and dark current noise maps were generated and stored for correction in full-field imaging with and without the anti-scatter grid. For slot-scan imaging, a modified readout scheme was used and a similar set of gain and dark current correction maps were generated and used to correct the images acquired externally following the image acquisition. To generate the gain correction maps, four flat-field images were acquired and averaged to minimize the noise fluctuation for both full-field and slot-scan imaging methods. A 2-cm-thick aluminum block was placed at the output of x-ray tube to simulate patient attenuation while acquiring flat-field images. The dark current correction map was computed internally by averaging two none-exposure images with the FF method but was externally computed by averaging two none-exposure images with the SS method. The dark current corrected gain image was computed as

$$G(x,y) = G'(x,y) - D(x,y), \quad (1)$$

where $G'(x,y)$ is the average of the four uncorrected flat-field images, $D(x,y)$ is the average of two dark current images for full-field or slot-scan imaging method. The gain map for a pixel at location (x,y) was computed as

$$g(x,y) = \frac{\bar{G}}{G(x,y)} = \frac{\bar{G}}{G'(x,y) - D(x,y)}, \quad (2)$$

where $\bar{G}$ is mean value for dark current corrected gain image.

Following image acquisition, raw image data were corrected for dark current and gain nonuniformities using following equation:

$$I_{\text{corrected}}(x,y) = g(x,y) \cdot [I_{\text{raw}}(x,y) - D(x,y)]. \quad (3)$$

Artifacts due to defective (totally inactive) or marginally functional pixels may also interfere with the diagnostic use of the FP images. To eliminate these artifacts, flat-field images were first analyzed to detect and locate defective and marginally functional pixels in the flat-field image and storing their positions to locate and correct for defective pixels later. In full-field imaging, correction was made by replacing values of defective pixels with those interpolated from the surrounding pixels internally by detector controller. In slot-scan imaging, the “image correction” function of the detector controller was turned off so that only the uncorrected raw image data will be transferred to and saved in the acquisition workstation. The process to correct for defective pixels was done in a similar manner as was done internally by detector controller. After image offset/gain calibration and defective pixel correction process, the linear scaled image data can be used for the scatter and noise property studies.

II.D. Measurement of primary, scatter signals, and SPRs

Various methods have been developed to measure the scatter components in image signals, including the beam stop, slat (e.g., steel plate), edge spread, and aperture methods.^{38–43} Among them, the beam stop method has been widely used to measure the scatter component at a given point while the slat or edge spread method allows the scatter component to be measured along a straight line (for scatter distribution). In this study, we used a method referred to as the beam aperture array method to indirectly estimate the scatter component at selected locations. With this method, an aperture mask was constructed and placed at 156.2 cm (61.5 in.) from the x-ray focal spot on the tube side of the phantom for imaging. The aperture mask consists of a 2D array of aperture holes [2 mm in diameter and spaced center-to-center by 2.54 cm (1 in.) vertically and horizontally] opened in a 2-mm-thick lead sheet laminated with a 4-mm-thick acrylic sheet. The combination of the lead and acrylic sheet attenuates about 99.9% of the x-rays at 120 kV_p, thus blocking almost all x-rays outside the aperture area.

In each chest phantom image acquired with the aperture mask in place, the signals in the aperture areas are mostly primary signals. Those outside the aperture areas are nearly zero (blocked by the lead/acrylic sheet) except in regions immediately surrounding the aperture areas where x-rays scattered from volume irradiated by the pencil beams formed by the aperture mask form small image signals. The image acquired without the aperture mask in place contains both the primary signals and scatter components. Thus, signals from

the image acquired with the aperture mask may be subtracted from those acquired without the aperture mask to estimate the scatter component in the aperture areas. For each aperture, signals in an 8 pixels × 8 pixels square region at the aperture center were averaged and used to estimate the primary signal, $I_p(i,j)$, and the primary plus scatter signal, $I_{p+s}(i,j)$, at the aperture center, (i,j) . Thus, the scatter component, $I_s(i,j)$, may be estimated by subtracting the two signals as follows:

$$I_s(i,j) = I_{p+s}(i,j) - I_p(i,j) \approx I_{p+s}(i,j) - I_{\text{Aperture}}(i,j). \quad (4)$$

Notice that $I_{\text{Aperture}}(i,j)$ is only an approximation for $I_p(i,j)$ because $I_{\text{Aperture}}(i,j)$ still contains a small scatter component resulting from scatter from volumes irradiated by the narrow pencil beams. Once the primary and scatter signals are estimated, the scatter-to-primary ratios (SPRs) may be computed as follows:

$$SPR(i,j) = \frac{I_s(i,j)}{I_p(i,j)} \approx \frac{I_{p+s}(i,j) - I_{\text{Aperture}}(i,j)}{I_{\text{Aperture}}(i,j)}. \quad (5)$$

Scatter degrades image quality metrics through the SPR. Since the primary signals may decrease in the scatter rejection process, it is more relevant to measure how the SPR, rather than the scatter component itself, is reduced. Thus, to quantify the scatter rejection ability of an imaging technique in relation to the image quality, the SPR reduction ratio (SPRRR) may be defined and computed as follows:

$$SPRRR(i,j) = \frac{SPR^{ff}(i,j) - SPR(i,j)}{SPR^{ff}(i,j)}, \quad (6)$$

where $SPR^{ff}(i,j)$ is the SPR in full-field, no-grid imaging.

To evaluate the accuracy of the aperture method for scatter measurement, apertures of different sizes (1–8 mm in diameter) were used to measure the scatter component for a 10.16-cm- (4 in.)-thick acrylic block placed against the FP detector. Mean signals were computed for a 5 pixels × 5 pixels region at the aperture center and normalized to signals measured for 2-mm-diam aperture. The results were plotted as a function of the aperture diameter in Fig. 3. The mean signals were also fitted to a straight line which, when extrapolated, intersects the y axis (zero aperture size) at a relative signal of 0.981, corresponding to an overestimation of the primary signal by about 2% when a 2-mm-diam aperture is used.

The above estimation was done with a single beam aperture at the detector center. However, there are several factors that affect the selections of aperture array parameters hence the accuracy of scatter estimation; they are mainly based on the following considerations⁴³:

- The aperture should be large enough to achieve sufficient statistical precision but small enough to ensure approximate signal uniformity within the ROIs and to minimize the aperture scatter in primary signal measurement;
- The primary x-ray beam cutoff inside the aperture should be small to leave the signals in the ROI unchanged;

FIG. 3. Normalized signal intensity plotted as a function of beam aperture diameter for a number of circular apertures with a linear fit.

- (c) The interspace between apertures should be large enough to minimize the aperture cross talk and small enough to achieve sufficient statistical precision within each anatomical region.

The primary x-ray beam cutoff plays an important role in determining the effective aperture size in the measurement of image metrics. Both the focal spot blurring due to its finite size and the geometric cutoff due to the finite thickness of the lead sheet are beam angle (or location) dependent and contribute to the primary x-ray beam cutoff inside the aperture, and their effects can be estimated based on the system setup. The geometric cutoff was estimated to be zero at the center of the image and 0.456 mm (the highest) at the four corners of image plane while a centered square image is selected (only a square image with size of 35.6 cm $\times$ 35.6 cm was chosen in the study). The focal spot blurring effect was estimated to be 17.3% of effective focal spot size. The focal spot size was measured to be 0.27 mm FWHM at the center of the detector in both vertical and horizontal directions in our study using a method similar to Rong *et al.*⁴⁴ Assume three times of nominal focal spot at the end of cathode side, the focal spot blurring results in a primary x-ray beam cutoff of 0.14 mm on image plane. Therefore a maximum total primary x-ray beam cutoff due to both focal spot blurring and geometric cutoff is estimated to be 0.60 mm on image plane. For an 8 pixels $\times$ 8 pixels ROI, in order to make aperture areas free from such effects and available for estimating the primary signals, the aperture size should be at least 2 mm in diameter.

The aperture cross talk caused by scattering from neighboring apertures was estimated with current settings of 2.54 cm aperture interspace. Two images of aperture array were acquired with a 10.16-cm- (4 in.)-thick acrylic block placed in front of the FP detector. Image signals over an aperture at the center of the matrix were measured to estimate the primary signals and then measured again with the aperture blocked to estimate the magnitude of the cross scatter from the neighboring apertures. Mean signals were com-

FIG. 4. (a) Experimental setup for primary, scatter, and scatter-to-primary ratio (SPR) measurement (a stationary anti-scatter grid was inserted at the detector front when the measurements were made for both full-field and slot-scan imaging with grid); (b) Radiograph of an anthropomorphic chest phantom with associated beam apertures for separating scatter component from primary signal at various locations. Highlighted areas indicate the regions used for the assessment of measurement in the lungs (1), mediastinum (2), retrocardia (3), and subdiaphragm (4).

puted for an 8 pixels $\times$ 8 pixels region at the center of the aperture for both images. Signal value calculated from the first image contains primary x-ray, in aperture scatter, and cross scatter from neighboring apertures, while that calculated from the second image contains cross scatter only. The aperture cross talk was estimated to be 1.22% of the estimated primary x-ray intensity in our study.

The experimental setup for scatter measurement and an image of the anthropomorphic chest phantom acquired with the aperture mask are shown in Figs. 4(a) and 4(b), respectively. Also shown in Fig. 4(b) are four different anatomical regions selected for this study: (1) Lungs, (2) mediastinum, (3) retrocardium, and (4) subdiaphragm. SPRs and SPRRRs measured for apertures within these regions were averaged for presentation and comparison. Aperture areas outside these regions were excluded from our analysis.

II.E. Measurement of CR and CNR

In digital imaging, the contrast signal may be defined as the signal difference between the contrast object and the area

surrounding it while the contrast ratio (CR) may be defined as the ratio of the contrast signal to the image signal. A most immediately obvious effect of scatter reduction is an increase of the image contrast ratio which may be measured as the CR at (i, j) as follows:

$$CR(i, j) = \frac{I(i, j) - I'(i, j)}{I(i, j)}, \quad (7)$$

where $I'(i, j)$ and $I(i, j)$ are the average image signals in the low-contrast object and its surrounding area, respectively. The CR was measured and averaged for all apertures within each of the four anatomical regions studied to evaluate the effect of scatter reduction on image contrast. However, image contrast can be freely manipulated in digital imaging. Thus, the CR alone by itself is insufficient for characterizing the image quality. To accurately characterize the low-contrast performance of an imaging system or technique, both the image contrast and noise level need to be measured and compared to each other in the evaluation.

In this study, the contrast-to-noise ratios (CNRs) were measured and used as a metrics to characterize and compare the low-contrast performance of various imaging systems or techniques. The CNR is defined as the signal difference between contrast object and the surrounding area divided by the noise level, and can be computed as follows:

$$CNR(i, j) = \frac{I(i, j) - I'(i, j)}{\sigma(i, j)}, \quad (8)$$

where $\sigma(i, j)$ is the root-mean-square noise level measured in the aperture area. To measure and compare the CNRs of different imaging techniques, a 2D array of aluminum beads [3 mm in diameter and spaced by 2.54 cm (1 in.) 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 had been 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 the primary signals, scatter components and SPRs.

To compute the noise levels in the image, two sets of images were acquired with identical settings (without the aperture mask or the array of aluminum bead) and subtracted from each other to eliminate signal variations due to anatomical structures. Prior to subtraction, the second set of images were normalized to the first set to equalize the image signal levels and eliminate signal variations that may arise from potential exposure fluctuation and incomplete cancellation of the anatomical structure. Following the subtraction, standard deviations were computed and divided by $\sqrt{2}$ to estimate the noise levels at the same locations where the primary signals, scatter components, SPRs had been measured.

As was done with SPRs and SPRRRs measurement, CRs, CNRs, and their improvement factors measured for apertures within four anatomic regions were averaged for presentation and comparison.

III. RESULTS

In Fig. 5(a), the SPRs for the four different imaging techniques, full-field without grid, full-field with grid, slot-scan without grid, and slot-scan with grid, are listed and plotted for comparison. All three scatter rejection techniques were effective in reducing the SPRs in all four anatomical regions. However, the reduction was more pronounced in the heavily attenuating mediastinum, retrocardium, and subdiaphragm regions and least pronounced in the lung areas. The slot-scan imaging technique appeared to be more effective. However, since the scatter rejection performance of slot-scan imaging depends on the slot width used, this simply reflects the performance of the particular slot width used, i.e., 1.6 cm. Figure 5(a) also shows that the slot-scan imaging technique can be made even more effective when used in conjunction with the anti-scatter grid but the further reduction is more pronounced in the heavily attenuating regions and less so in the lightly attenuating regions.

In Fig. 5(b), the SPR reduction ratios (SPRRRs), defined as the differences between the full-field no grid SPRs and the reduced SPRs divided by the former, are listed and plotted for comparison. The SPRRRs tended to be higher in the heavily attenuating regions and lower in the lightly attenuating regions. The addition of an anti-scatter grid appeared to be more effective in further reducing the SPRs in the heavily attenuating regions and less effective in the lightly attenuating regions.

In Fig. 6(a), the measured CRs are listed and compared. Figure 6(a) shows that both the slot-scan imaging and anti-scatter grid methods substantially improved the CRs. As expected, slot-scan imaging used in conjunction with an anti-scatter grid further improved the CRs. CRs varied with the regions and decreased in the order of the lungs, mediastinum, subdiaphragm, and retrocardium. The pattern of variation was similar for all techniques except that for slot-scan imaging with grid, the average CR in the subdiaphragm was higher than that in the mediastinum. The degree of variation also decreased in the order of full-field no grid, full-field with grid, slot-scan imaging, and slot-scan imaging with grid.

In Fig. 6(b), the CR improvement factors are defined as the ratio of the improved CRs to those for full-field, no grid imaging. The CR improvement factor varied with the region as well as the technique. It increased with the region in the order of the lungs, mediastinum, retrocardium, and subdiaphragm. It increased with technique in the order of full-field with grid, slot-scan imaging, and slot-scan imaging with grid.

In Fig. 7(a), the CNRs are plotted and compared. In Fig. 7(b), the CNR improvement factors, defined as the ratios of the improved CNRs to those for full-field, no grid imaging, are plotted, listed and compared. Figure 7(a) shows that the

FIG. 5. (a) Scatter-to-primary ratio (SPR) and (b) SPR reduction ratio (differences between the full-field no grid SPRs and the reduced SPRs divided by the former) measured in the lungs, mediastinum, retrocardia, and subdiaphragm for comparison for anti-scatter grid and slot-scan methods with same beam quality and quantity.

CNRs varied with the region and decreased in the order of the lungs, mediastinum, retrocardium, and subdiaphragm. This observation applies to all techniques except that for slot-scan imaging with grid, where the CNR in the subdiaphragm was slightly higher than that in the retrocardium. Both Figs. 7(a) and 7(b) show that in the lungs, the addition of an anti-scatter grid slightly degraded the CNRs for both full-field and slot-scan imaging. In all other regions, the addition of an anti-scatter grid improved the CNRs only minimally for both full-field and slot-scan imaging. In contrast to the anti-scatter grid method, slot-scan imaging resulted in substantially greater improvement in the CNRs. For both the anti-scatter grid and slot-scan imaging methods, the improvement of CNRs varied with the region and increased in the order of the lungs, mediastinum, retrocardium, and subdiaphragm.

IV. DISCUSSION AND CONCLUSIONS

In this paper, we investigated and compared the scatter rejection ability and the low-contrast performance of the slot-scan/ALER imaging techniques with those of full-field imaging techniques used with and without an anti-scatter grid. It was shown that the slot-scan imaging technique used with a 16-mm-wide beam width can be used to achieve slightly better scatter rejection than the anti-scatter grid method. However, in terms of the more relevant low-contrast performance as measured by the CNRs, the slot-scan/ALER technique had a substantially higher performance. The addition of an anti-scatter grid in slot-scan/ALER imaging resulted in improved scatter rejection but similar or degraded low-contrast performance. Among the four anatomical regions, least improvement in both scatter reduction and low-

FIG. 6. (a) Contrast ratio (CR) and (b) CR improvement factor (the ratio of the improved CRs to those for full-field, no grid imaging) measured in the lungs, mediastinum, retrocardia, and subdiaphragm for the full-field (FF) and slot-scan (SS) digital radiography systems with same beam quality and quantity.

contrast performance was observed in the lungs with both the slot-scan/ALER and anti-scatter grid techniques. The improvement was higher in other regions in the order of mediastinum, retrocardium, and subdiaphragm, indicating that all scatter rejection methods became more effective in more attenuating regions.

It should be noted that the ALER technique is an electronic collimation method. With an anti-collimator, scattered x-rays are blocked and prevented from exposing detector area outside the scanning fan beam. With the ALER technique, x-ray scatter from the irradiated part of the patient is allowed to expose the detector area outside the fan beam. However, scatter signals accumulated prior to the arrival of the fan beam are erased immediately before an image line is used to accumulate the fan-beam exposure. Scatter signals accumulated after the fan-beam passes are of no consequence as the fan-beam exposure signals are already read out. Thus,

the effectiveness of the ALER technique is dictated by how well the scatter signals can be erased. Our experimental results show that the scatter erasure process has worked well in the slot-scan/ALER imaging technique. We have conducted a measurement of residual scatter to estimate the effectiveness of scatter erasure with ALER technique and found that more than 99% of scatter component accumulated prior to the fan-beam exposure was erased.

The quantitative measurement of effectiveness of the scatter erasure was performed by imaging a 5.08-cm (2 in.)-thick Lucite slab (placed in front of the detector) with ALER technique. X-ray was turned off near the end of the scanning/readout process at approximately 4 S (total scanning time is 4.4 S in our study). At the time the exposure was terminated, about 4.5% of image lines near the end contained accumulated scatter signals that had not been read out. After terminating the exposure, these lines continued to be erased and

FIG. 7. (a) CNR and (b) CNR improvement factor (the ratio of the improved CNRs to those for full-field, no grid imaging) measured in the lungs, mediastinum, retrocardia, and subdiaphragm for the full-field (FF) and slot-scan (SS) digital radiography systems with same beam quality and quantity.

then read out. Since the x-rays were turned off already, the readout signals contained only residual signals remaining after the erasure of the scatter signals. These signals would be a good indication of the efficiency of scatter erasure. They were measured to be approximately 2–3 DUs (digital units). Two additional images of the 5.08-cm (2 in.)-thick Lucite slab were acquired using the full-field acquisition mode with the same exposure level as the effective exposure level in slot-scan imaging to estimate the scatter signals accumulated prior to scatter erasure. One image was with the field fully opened, and the other with the fore-collimator to block the image lines corresponding to the 4 S of scan, leaving the opened area (corresponding to the last 0.4 S of scan in slot-scan imaging) exposed to the x-ray. The latter was subtracted from the former and the subtraction signals in the open area, measured to be 1400 DUs, give an estimate of the scatter

signals accumulated in slot scan imaging. Thus, only 0.14%–0.21% of the scatter signals remained after the erasure.

In general, a short exposure is made before the image readout in the regular full-field imaging mode while the exposure is made relatively long “during” the entire readout period in the slot-scan imaging mode. The tube loading, or mAs setting, is increased by a factor determined by the ratio of the detector height (in the scanning direction) to the scanning fan beam width determined by the spacing between the leading edge line being erased and the trailing edge line being subsequently read out on detector plane. For the configuration of our slot-scan imaging system, the tube loading (or mAs) was increased by roughly a factor of 22 to maintain the effective primary exposure at the detector input at the same level as with regular full-field imaging without grid. Narrower slot width results in better scatter rejection and low-

contrast performance; however it requires higher tube current in order to maintain sufficient x-ray exposure and hence higher tube loading.

Slot-scan imaging can also be implemented with a slot shaped detector array. With this approach, time-delay-integration (TDI) CCDs fiber-optically coupled with scintillators are often used as the detector. The advantage of this approach is the potentially lower cost of the detector components themselves. However, the need to mount the detector array on a fast translational stage to move it in synchronization with the scanning fan beam is still a serious challenge and may well elevate the cost of this approach. In contrast, the ALER technique can be implemented with a FP based digital radiography system as an added feature. Synchronization of the scatter erasure and image readout processes can be electronically fine tuned to synchronize to the motion of scanning fan beam. An additional advantage is the ability to accommodate various beam width by electronically adjusting the spacing between the image lines for scatter erasure and readout.

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