Published in final edited form as:

Phys Med Biol. 2009 November 21; 54(22): 6959–6978. doi:10.1088/0031-9155/54/22/014.

Scan equalization digital radiography (SEDR) implemented with an amorphous Selenium flat-panel detector: initial experience

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Abstract

It is well recognized in projection radiography that low-contrast detectability suffered in heavily attenuating regions due to excessively low x-ray fluence to the image receptor and higher noise levels. Exposure equalization can improve image quality by increasing the x-ray exposure to heavily attenuating regions, resulting in a more uniform distribution of exposure to the detector. Image quality is also expected to be improved by using the slot-scan geometry to reject scattered radiation effectively without degrading primary x-rays. This paper describes the design of prototype scan equalization digital radiography (SEDR) system implemented with an amorphous silicon (a-Si) thin-film transistor (TFT) array based flat-panel detector. With this system, the slot-scan geometry with the alternate line erasure and readout (ALER) technique was used to achieve scatter rejection. A seven-segment beam height modulator assembly was mounted onto the fore-collimator to regulate exposure regionally for chest radiography. The beam modulator assembly, consisting of micro linear motors, lead screw cartridge with lead beam blocks attached, position feedback sensors, and motor driver circuitry, has been tested and found to have an acceptable response for exposure equalization in chest radiography. An anthropomorphic chest phantom was imaged in the posterior-anterior (PA) view under clinical conditions. Scatter component, primary x-rays, scatter-to-primary ratios (SPRs), and primary signal-to-noise ratios (PSNRs) were measured in the SEDR images to evaluate the rejection and redistribution of scattered radiation, and compared with those for conventional full-field imaging with and without anti-scatter grid methods. SPR reduction ratios (SPRRRs, defined as the differences between the non-grid full-field SPRs and the reduced SPRs divided by the former) yielded approximately 59% for the full-field imaging with grid and 82% for SEDR technique in the lungs; and 77% for the full-field imaging with grid and 95% for SEDR technique in the subdiaphragm. The SEDR technique demonstrated a substantial improvement in PSNRs over the anti-scatter grid technique. The improvements of PSNRs are varied with the regions and are more pronounced in the heavily attenuating regions.

Keywords

exposure equalization; slot-scan; digital radiography; flat-panel detector; scattered radiation; scatter rejection; signal-to-noise ratio

1. Introduction

Projection radiography has remained as the primary tool for initial diagnosis of diseases and for screening. Despite the transition from conventional screen-film (SF) technology to flat-

panel (FP) based digital detector technology with the hope that the diagnostic performance of digital imaging will be superior to that of SF radiography and that digital imaging will provide dose efficiency that is similar to or better than that of SF radiography, the use of the area detector as image receptor is subject to the acceptance of: (1) large amount of scattered radiation and, (2) degraded low-contrast detectability in the heavily attenuating regions.

The anti-scatter grid (typically a stationary grid with high grid ratio) method was adopted from the conventional SF radiography technology to removed the scattered radiation and improve the image low contrast performance. The grid method is considered to be the most practical approach and is used in many digital radiographic systems widely (Shaw *et al.*, 1994; Bernhardt *et al.*, 2000; Chan *et al.*, 1990; Barski and Wang, 1999; Fetterly and Schueler, 2007). The main advantage of using an anti-scatter grid is its simplicity and effectiveness. However the grid attenuates a substantial fraction of the primary x-rays by as much as 30 – 50%, resulting in a significant loss of information and an increase of relative noise level in the image. Therefore to compensate for the attenuation of primary x-rays by grid, the entrance exposure must be increased to preserve the quantum noise level. An alternative technique to overcome the shortcomings of the anti-scatter grid, yet reject scattered radiation without having to attenuate the primary x-ray is the slot-scan method which uses a collimated narrow fan beam to scan the patient (Liu *et al.*, 2008; Sashin *et al.*, 1982; Plenkovich *et al.*, 1986; Holdsworth *et al.*, 1990; Barnes *et al.*, 1994; Jing *et al.*, 1998; Mainprize *et al.*, 2002; Samei *et al.*, 2005; Liu *et al.*, 2006). The slot-scan technique has the advantage of dose efficiency, as primary x-rays do not have to be attenuated. Therefore patient dose can be reduced significantly while detector exposure is unchanged. Research has shown that when a proper slot width is selected the image contrast-to-noise ratios (CNRs) from slot-scan technique are better than that from conventional full-field radiography with an anti-scatter grid.

The second issue that has been long recognized is that low x-ray photon flux in heavily attenuating regions results in poorer image signal-to-noise ratio (SNR), leading to degraded detectability and visualization of low contrast objects. The technical limitation in conventional chest radiography arises from the wide variation in patient thickness between various body parts while exposures are routinely made to optimize the contrast in the lung areas. This leads to the exposures in areas such as retrocardia, mediastinum, and subdiaphragm falling outside the useful exposure range for conventional SF radiography (Plewes, 1983). Even though with the ability to enhance image contrast digitally (windowing & leveling) for digital radiography, nodule detectability still suffers from decreased SNR from reduced photon flux over the thicker body parts. To overcome this problem, the x-ray beam incident to the patient may be spatially modulated to compensate for variation of patient anatomic structures and to achieve a more uniform exposure at the image receptor. This method, namely exposure equalization, results in an increased exposure to the heavily attenuating regions and decreased exposure to the lightly attenuating regions. Several exposure equalization techniques have been implemented and investigated in the past (Peppler *et al.*, 1982; Hasegawa *et al.*, 1986; Hasegawa *et al.*, 1987; Boone *et al.*, 1993; Boone *et al.*, 1995; Goodsitt *et al.*, 1998; Rudin and Bednarek, 1980; Rudin *et al.*, 1999; Fletcher *et al.*, 1996; Molloy *et al.*, 1999; Molloy *et al.*, 2001; Lam and Chan, 1990; Vlasbloem and Kool, 1988; Chotas *et al.*, 1990; Plewes and Wandtke, 1982; Plewes and Vogelstein, 1983; Wandtke *et al.*, 1988; Kool *et al.*, 1988; Panayiotakis *et al.*, 1998; Sabol *et al.*, 1993; Sabol *et al.*, 1996;), all of which have demonstrated improved image quality particularly in the heavily attenuating regions. Exposure equalization radiography may be divided into two categories: those based on conventional full-field image acquisition and those based on spot-/slot-scan imaging geometry. The former is called area x-ray beam equalization because it relies on the use of a two dimensional beam attenuator array to equalize the exposure onto an area image receptor. The later is called scanning equalization because it relies on the use of scanning spot or slot geometry with x-ray beam intensity modulation typically in one dimension to regulate the exposure onto a linear or area image receptor.

The area x-ray beam equalization method usually uses beam attenuator approach which contains a pre-fabricated filter mask (x-ray beam attenuating material) to absorb x-ray beam intensities in the lightly attenuating regions while keeping less (or no) absorption in the heavily attenuating regions. A low dose preliminary image is acquired in order to generate a filter mask or to select an appropriate pre-fabricated filter mask. The advantage of the area x-ray beam equalization technique is its simplicity while providing adequate exposure control, enabling short exposure time and lower tube loading in conjunction with conventional open-field radiography (SF or digital radiography), and fast framing rate in image intensifier (II) based imaging such as region of interest (ROI) fluoroscopy and digital angiography (Pepler *et al.*, 1982; Hasegawa *et al.*, 1986; Hasegawa *et al.*, 1987; Boone *et al.*, 1993; Boone *et al.*, 1995; Goodsitt *et al.*, 1998; Rudin and Bednarek, 1980; Rudin *et al.*, 1999; Fletcher *et al.*, 1996; Molloy *et al.*, 1999; Molloy *et al.*, 2001). However this approach requires a filter mask to be fabricated which is time consuming. The patient input beam quality is regulated region by region due to the relative amount of beam attenuating material used and hence introduces the complexity in calculating the attenuating thickness accurately. The area exposure equalization does not provide any scatter rejection therefore an anti-scatter grid can be used to reduce the scattered radiation, as often used in conventional SF or digital radiography.

The scanning equalization radiography (SER) (Lam and Chan, 1990; Vlasbloem and Kool, 1988; Chotas *et al.*, 1990; Plewes and Wandtke, 1982; Plewes and Vogelstein, 1983; Wandtke *et al.*, 1988; Kool *et al.*, 1988) on the other hand, employs spot-/slot-scan geometry to regulate the entrance exposure regionally while it rejects the scattered radiation effectively. Spatially and temporarily varying beam height or x-ray tube current pulse width modulation can be used to achieve exposure equalization during the scan. With the scanning equalization technique, the regional exposure is controlled through a feedback system which detects the attenuating characteristics of patient's anatomy. The feedback control can be real time where an ion chamber or the image receptor is used to detect and regulate the x-ray intensity irradiating to the image receptor. X-ray beam intensity modulation can be done either by modulating the exposure time by varying exposure aperture size while keeping x-ray tube current fixed, or by modulating the x-ray tube current while keeping the effective exposure time unchanged. An example of real-time SER commercial system is Advanced Multiple Beam Equalization Radiography (AMBER, Oldelft). With a non real-time feedback approach, a low dose preliminary image is first acquired using slot-scan mechanism, and then used to generate a beam intensity modulation function to control exposure amount regionally during the exposure equalization process. The scanning equalization technique does not alter the x-ray beam quality therefore the subject contrast remains unchanged during the exposure equalization process. The geometry used in scanning exposure equalization allows scattered radiation to be rejected effectively while keeping primary x-rays un-attenuated. Both signal-to-noise ratios (SNRs) and contrast-to-noise ratios (CNRs) can be improved significantly in heavily attenuating regions. Due to the fact that SER requires a longer total scanning time (pre-scan plus final SER scan), the SER technique can be applied to stationary imaging modality only, such as chest radiography or mammography for improved the image quality.

We are in the process of investigating a scanning equalization digital radiography (SEDR) system for chest radiography as shown in figure 1. A prototype SEDR system was developed to improve the image quality in heavily attenuating regions in chest images. The system was implemented with an amorphous Selenium (a-Se) based FP detector and the alternate line erasure and readout (ALER) technique (Liu *et al.*, 2006; Liu *et al.*, 2008) for slot-scan imaging geometry to reject scattered radiation without attenuating primary x-rays. Multiple beam height modulation was used to equalize the x-ray exposures regionally. In this study, we report on the design and construction of the SEDR system. The preliminary results of rejection and redistribution of scattered radiation and PSNRs for different anatomical regions from images of an anthropomorphic chest phantom will be discussed in this paper as well.

2. Materials and methods

2.1 Image system

The prototype SEDR developed in our lab comprises a FP detector system, a image readout mode controller, an acquisition/processing workstation, a multiple channel digital-to-analogue conversion (DAC) interface, a multiple channel linear motor driver, a seven-segment x-ray fan-beam height modulator assembly, and an x-ray unit. The digital detector used in this study is an amorphous Silicon (a-Si) thin-film-transistor (TFT) a-Se based direct conversion 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 rectangular-shaped FP detector was positioned in a “landscape” orientation (instead of “portrait” orientation in conventional chest radiography) in order to allow the electronic aft-collimation move vertically in synchronization with the motion of the scanning x-ray fan beam during SEDR imaging. The image readout signals are digitized into 14-bits image data via onboard analog-to-digital converters (ADCs) and transferred to a PC based detector controller. The detector controller not only performs image data correction (gain, dark current noise, and dead pixel) and stores calibrated data temporarily, but also supplies and monitors low and high voltages to the detector panel and peripheral electronics. The image readout mode controller was custom designed and built to alter the detector readout scheme and allow the detector to be operated either in the conventional full-field (FF) or slot-scan (SS) imaging mode. The SS mode will be used for the SEDR imaging of chest phantom as well. An Intel Pentium IV based workstation (Precision 340, Dell Inc., Austin, TX) with a CPU frequency of 2.80 GHz and 512 MB RAM is used to control the motion of the fore-collimator, beam modulator assembly, and image acquisition. Image data were transferred from the detector controller to the workstation for post-processing and display. Full-field raw image data were corrected automatically by the detector controller while SS and SEDR raw image data were corrected externally using the acquisition/processing workstation after image acquisition.

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) was used in this study. The tube has dual focal spot sizes: 0.3 mm and 1.2 mm however only large focal spot was used in this study.

A computer-controlled scanning fore-collimator system was mounted at the output of the x-ray tube to generate a narrow fan-shape beam for SS or SEDR imaging. The collimator was mounted on a high-precision linear drive (Velmex, Inc., Bloomfield, NY) to allow for the horizontal oriented fan-shape beam to be moved vertically. The fore-collimator was constructed from a pair of 15 cm × 15 cm square tungsten plates with a thickness of 2 mm that attenuates nearly all x-rays at 120 kVp. The fore-collimator was placed at 40.6 cm (16 in.) from the x-ray focal spot while the source-to-image distance (SID) was kept at 183 cm (72 in.) leading to a magnification factor of ~4.5. The fore-collimator gap was selected to be 0.358 cm, resulting in a fan beam width of 1.61 cm at the image plane (equivalent to 116 image lines). These settings resulted in an effective exposure time of approximately 0.2 s for each image line in slot-scan imaging.

The penumbra effect caused by the finite focal spot size at the edges of the fore-collimator was measured to be about 8 and 30 image lines on both leading and trailing edges of the slot, leading to a drop of integral signal by about 1.7% and 6.5% for small and large focus in slot-scan/SEDR imaging, respectively. The large focal spot mode was used in this study and the focal-spot blurring due to the large magnification factor would help minimize the edge artifacts from subsequent beam modulators during SEDR image acquisition (Peppler *et al.*, 1982; Hasegawa *et al.*, 1986; Molloy *et al.*, 1999; Molloy *et al.*, 2001).

A 32-channel high-speed analog output PCI card (NI PCI-6723, National Instruments Corporation, Austin, TX) was installed in the acquisition workstation and served as digital-to-analogue conversion interface between the acquisition workstation and the beam modulator drivers. The card provides high speed analog output of up to 32 channels with a sample speed of 45 kS/s per channel, ± 10 V output range, and 13-bit (2.44 mV) resolution. However only seven analog output channels were used to transfer the beam height modulation voltage signals from the acquisition workstation to the beam modulator drivers in the prototype SEDR system.

2.2 X-ray fan beam height modulation and calibration

The slot-scan geometry and beam height modulation mechanism were used in the prototype of SEDR system. As shown in figure 2, the x-ray was collimated by a pair of tungsten plates to form a narrow fan beam with fixed gap space. A seven-segment beam height modulation assembly was custom developed and mounted on the top plate of the fore-collimator. Each beam height modulator has a lead beam blocker attached to the lower end of the modulation assembly. X-ray beam height (defined as the vertical space between the lower fore-collimator plate and beam modulator) can be regulated independently by moving each x-ray beam blocker vertically. The x-ray intensity irradiated to the patient is therefore modulated regionally during the SEDR imaging process. A slot-scan image can be acquired when all the beam height modulators are lifted up away from the x-ray fan beam path or the SEDR imaging can be performed when beam height modulators are moved down with local integral x-ray beam intensity corresponding to the detector exposure required. The beam height modulator array is positioned at approximately 2 mm away from the surface of the upper fore-collimator plate. The trapezoid shaped beam modulators were constructed with 3 mm thick lead which attenuate more than 99.9% of the x-ray photon flux at 120 kVp. The physical arrangement allows each modulator to move up and down freely with proper coverage at the interface of two subsequent beam modulators. Each beam modulator covers about 10.6 mm wide horizontally, with 0.3 mm overlap on each side of neighboring beam blockers, corresponding to an approximate 4.77 cm wide strip in the image plane.

A micro linear stage assembly was developed to drive the beam modulator blockers. The driver system consists of micro linear motors, lead screws to translate the rotational movement to the linear movement, linear position feedback sensors, and driver circuitry. The complete beam modulation driver system is illustrated in figure 3. The reference signal can be used to manually set the initial displacement for each individual beam modulator. The beam modulator moves up or down and produces a displacement proportional to the change of the input modulation signal (from multiple channel DAC card). The position feedback circuitry senses the movement and generates a signal negative to the input modulation signal when the beam height modulator reaches required displacement. The maximum displacement of the beam modulator assembly is 20 mm with a voltage change of 9.2 V, resulting in an average displacement sensitivity of 2.17 mm/V.

The beam height modulation array has to be calibrated before the acquisition of SEDR image. All beam height modulators were pre-set (roughly) to just block the x-ray beam from transmitting through the fore-collimator path at the beginning. A series of “flat-field” slot-scan images (the images were striped rather than totally flat due to the beam height variation among beam modulators) were acquired with the input voltages vary from 0.1 V to 2.5 V with an increment of 0.2 V. The system was designed to have the x-ray beam height proportional to the modulation signal. However the relationship between image signals and modulation signals may not be perfectly linear due to the angular geometric variation during the scanning. Four images were acquired and averaged to minimize the noise fluctuation for each input voltage setting. Corrections for defective pixel, dark current noise, and gain non-uniformity were performed prior to image averaging. The central part of the processed image was then divided

into seven segments horizontally with each segment strip contains 324 vertical lines. Image data were averaged horizontally line-by-line within each segment strip to yield seven signal profiles in the scanning direction for each modulation voltage setting. A total of thirteen segmented “flat-field” image profiles (size 7×2560) were generated. Image signal intensity versus modulation voltage for each individual segment at each scanning location can be determined from these segmented “flat-field” image profiles. A sample plot of image signal intensity versus modulation voltage is shown in figure 4. The normalized image signal intensity versus modulation voltage, is therefore defined as beam modulation transfer function (BMTF). The initial resolution of BMTF is relatively low and cannot be used directly for accurate beam height modulation. Therefore the interpolation process is implemented to increase the resolution of the BMTF in order to cover at least one image line with a minimum, non zero, change of modulation voltage. The BMTF was interpolated with 20-points between two initial sample points following the seven-order polynomial curve fitting, resulting in an improvement of modulation resolution from 0.2 V to 0.01 V. The 0.01 V voltage change corresponds to an average minimal modulation displacement of 0.0217 mm (0.01 V times 2.17 mm/V), yielding a change of 0.098 mm in image plane. It is desirable to have an average minimal modulation displacement smaller than the detector pixel size. The final BMTF was stored as a look up table (BMTF-LUT) in the image acquisition/processing workstation for use in final SEDR imaging.

The modulation response is an important factor to the beam height modulation array. Factors that limit the modulator’s response include the friction caused by the mechanical movement of the motor and lead screw assembly, position feedback sensor, and the load of the beam modulation blocker (the sum of the lead modulation blocker, aluminum arm and assembly screws). To evaluate the actual beam height modulation response, a square waveform voltage was applied to the beam modulation assembly and the output of the linear motor driver (at two terminals of the linear motor) was measured. This output signal appeared as a positive (or negative) pulse at the rising (or falling) edge of the square waveform. The measurement was recorded using a four channel digital oscilloscope (Model TDS 3054B, Tektronix Inc., Beaverton, OR) with one probe connected to the input square waveform, and the other connected to the terminals of the linear motor.

2.3 Pre-scan image acquisition

A pre-scan image was acquired and to compute the exposure equalization pattern (or beam height modulation factor, the percentage of modulated beam height to the maximum beam height, i.e. the physical dimension of the fore-collimator). The radiation exposure from the prescan image counts toward the total patient exposure in SEDR imaging. Therefore the exposure level for the pre-scan image has to be as low as possible to minimize overall patient dose, but high enough to preserve proper signal level and accuracy especially in heavily attenuating regions for the calculation of the exposure equalization pattern. An exposure of 20% of the conventional radiography setting was used for the pre-scan image acquisition in this study. The pre-scan image was acquired at 4 mA (equivalent to 0.8 mAs) using the slot-scan imaging mode when all x-ray beam modulators are moved away from the narrow x-ray fan-beam path to allow the fan-beam to expose the chest phantom directly.

2.4 Image segmentation and exposure pattern calculation

Following the acquisition of the pre-scan image, uncompensated raw image data was transferred to the image acquisition workstation and corrected for dark current background, gain non-uniformities, and defective pixels/lines using a method previously reported on (Liu *et al.*, 2008). The corrected image was then divided into seven segment strips horizontally in the central area of the chest phantom image. Each segment contains 324 vertical image lines, corresponding to a 4.5 cm narrow strip on image plane. Image data were averaged horizontally line-by-line within the segment strip to yield seven image profiles in the scanning direction.

This process was the same as what was done for the beam modulation calibration. Each segmented pre-scan image profile line was convolved with a linear filter kernel to smooth out the signal profile and reduce the noise fluctuation. Various kernel sizes have been studied to determine the proper filtering effect on the pre-scan image profile while maintaining the anatomic information less impacted.

The objective of exposure equalization is to regionally modulate the entrance exposure level to achieve uniform exposure levels (average over a large area) over the main part of the chest image, while keeping the overall average entrance skin exposure (ESE) level at the same level as in conventional full-field chest radiography. The exposure equalization factor that will be used to regulate the entrance exposure for individual segment can be computed from the pre-scan image as follows:

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

where the subscripts, “eq” and “pre”, refer to equalization and pre-scan, respectively; $\overline{I_{pre}}$ is average image signal over a central area which covers the lungs, mediastinal, and subdiaphragmatic regions in the pre-scan image (signals of air at the boundary of the chest were excluded). The exposure equalization factors should be smaller than 1 for lightly attenuating regions such as lungs, indicating reduced entrance exposure should be expected; while the exposure equalization factors are greater than 1 for heavily attenuating regions such as the mediastinum and subdiaphragm, indicating increased entrance exposure is expected in order to elevate detector exposures in these areas in SEDR imaging. Therefore the x-ray tube current need to be increased to compensate for the increased entrance exposure in heavily attenuating regions during SEDR imaging. The tube current setting for SEDR imaging is determined by the follows:

$$i_{eq} \geq i_{pre} \cdot E_{max} \quad (2)$$

$$E_{max} = \max(E_{eq}(i, j)) \quad (3)$$

where i_{eq} and i_{pre} are the x-ray tube current settings for SEDR and pre-scan (slot-scan) imaging, respectively. The tube current settings in commercial x-ray generators are usually discrete instead of continuous. The selection of x-ray tube current for SEDR imaging i_{eq} is therefore determined by a lowest available setting no less than i_{pre} times E_{max} . The beam height modulation factor, defined as the ratio of x-ray beam height required for SEDR imaging to the maximum beam height (the fixed fore-collimator height) as follows:

$$m_{eq}(i, j) = \frac{i_{pre}}{i_{eq}} \cdot E_{eq}(i, j) = \frac{i_{pre}}{i_{eq}} \cdot \frac{\overline{I_{pre}}}{I_{pre}(i, j)} \quad (4)$$

The beam height required for modulator i at location j for SEDR imaging, is therefore determined by:

$$h_{eq}(i, j) = m_{eq}(i, j) \cdot H = \frac{i_{pre}}{i_{eq}} \cdot \frac{\overline{I_{pre}}}{I_{pre}(i, j)} \cdot H \quad (5)$$

where H is the maximum beam height determined by the fore-collimator gap, was set to be 0.358 cm in this study.

The beam height modulation factor, $m_{eq}(i, j)$ calculated from Eq. 4 is compared with BMTF-LUT(i, j) to determine the modulation voltage $V_{eq}(i, j)$ to be applied. The modulation voltage $V_{eq}(i, j)$ was stored in the acquisition workstation before SEDR imaging and was sent to beam modulator assembly synchronized to the moving of fan beam x-rays with an external trigger signal provided by detector system. A flow chart for SEDR imaging is illustrated in figure 5.

2.5 Chest phantom image acquisition

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 kV_P and 4 mAs for full-field imaging, the techniques comparable to those used for clinical chest radiography procedures. A stationary but removable anti-scatter grid with a 13:1 grid ratio and 78 lines/cm (Mitaya MFG. Co., Ltd., Tokyo, Japan) was used for full-field imaging technique and to compare the scatter rejection and noise properties with the SEDR imaging technique. It has been measured that the use of an anti-scatter grid attenuates about 30% of primary x-rays at 120kV_P in this study (Liu *et al.*, 2006; Liu *et al.*, 2008). The grid was removed from the x-ray path when the SS (for pre-scan image acquisition) and SEDR imaging techniques were evaluated.

An edge device technique (Cooper *et al.*, 2000) was used to quantify the scattered radiation and SPR across chest phantom images. The technique requires two exposures and gives a direct measurement of the spatial distribution of scatter along the edge in any direction (particularly in the horizontal or vertical directions), differing from the beam blocker technique (which provides scatter estimation at certain locations only). A large piece of rectangular tungsten plate (2 mm thick) with sharp edge was used as the edge device in this study. The edge device was carefully positioned in front of the chest phantom at a distance of 155 cm (61 in.) from the x-ray source for measuring the position dependent scattered radiation horizontally across the chest phantom image. The focal spot blurring caused by the tungsten edge was measured to be 0.055 mm and 0.218 mm at the image plane for small and large focus, respectively, are negligible in the scatter measurement. Two images were acquired to estimate the scattered radiation along the edge. The first image was acquired with upper portion of the chest phantom blocked by the tungsten plate, and the second image was acquired with the edge re-oriented 180 degrees from its original position with lower portion of the chest phantom blocked by the tungsten plate. The scatter distribution, $I_S(x)$, was estimated by adding up the scatter components from the two measurements along shadow of the edge. A third chest phantom image was acquired to measure the total (primary plus scatter) image signal, $I_{P+S}(x)$, without presence of the edge device at the position where scatter was measured. The primary signal, $I_P(x)$, was computed as the difference between the total and scatter signals, $I_{P+S}(x) - I_S(x)$.

The edge device technique was developed to estimate the scatter spatial distribution using edge spread function (in direction perpendicular to the edge) and requires a homogeneous scattering media to be imaged in full-field digital mammography initially (Cooper *et al.*, 2000). The work performed in this study does not require the edge spread functions to be used hence the technique is not limited to a homogeneous scattering media. To validate this technique for chest radiography, we have also performed scatter estimation using lead disks of various diameters

ranging from 16.4 mm down to 2.55 mm. A uniform Lucite block with a thickness of 2 in. was placed at the detector surface and was imaged at 120 kVp with exposure of 4 mAs. Scatter fractions (signal measured behind the disk over signal measured without disk at the disk center) versus lead disk diameter were plotted for three different imaging techniques: full-field no grid, full-field with grid, and slot-scan. By extrapolating the plot to zero-diameter lead disk, the true scatter fractions are found to be 0.540, 0.266, and 0.185 for full-field no grid, full-field with grid, and slot-scan imaging, respectively. The scatter fractions estimated from edge spread technique are 0.541, 0.269, and 0.182, resulting in errors of 0.19%, 1.13%, and 1.62% for full-field no grid, full-field with grid, and slot-scan imaging, respectively. These results indicated that the edge spread technique can be used to estimate the scatter radiation with very small errors for both full-field and slot-scan (or SEDR) chest radiography.

Following the measurement of $I_S(x)$ and $I_{P+S}(x)$, the position dependent SPR and SPR reduction ratio (SPRRR, defined as the differences between the full-field non-grid SPRs and the reduced SPRs divided by the former), $SPR(x)$ and $SPRRR(x)$, were computed as follows:

$$SPR(x) = \frac{I_S(x)}{I_P(x)} = \frac{I_S(x)}{I_{P+S}(x) - I_S(x)} \quad (6)$$

$$SPRRR(x) = \frac{SPR^{ff}(x) - SPR(x)}{SPR^{ff}(x)} \quad (7)$$

where the superscripts, “ff”, refer to conventional full-field imaging without anti-scatter grid. The SPRRR measures the percentage of SPR reduced with scatter removal/rejection methods from the conventional non-grid full-field image signal.

In digital imaging, the contrast signal is defined as the signal difference between the contrast object and the area surrounding it and can be approximated as follows:

$$I - I' = (I_P + I_S) - (I'_P + I'_S) = I_P - I'_P = I_P \cdot (1 - e^{-\Delta\mu \cdot \Delta t}) \approx I_P \cdot (\Delta\mu \cdot \Delta t) \quad (8)$$

where $\Delta\mu$ is the difference in the linear attenuation coefficient between the low-contrast object and background; Δt is the thickness of the low-contrast object. Since $\Delta\mu \cdot \Delta t$ does not vary with the scatter rejection techniques investigated in this study, the CNR is approximately proportional to the ratio of the primary signal to noise level:

$$CNR = \frac{I - I'}{\sigma} \approx \frac{I_P}{\sigma} \cdot \Delta\mu \Delta t \propto \frac{I_P}{\sigma} \quad (9)$$

Thus, we can use the primary signal-to-noise ratios (PSNRs) to compare the low-contrast performance of various imaging systems or techniques as long as the object contrast, $\Delta\mu \cdot \Delta t$, is not altered (generally the case if the kVp and filtration remain the same). The advantage of using the PSNRs for low-contrast performance comparison is that the presence and use of a contrast object are not required. The position dependent primary signal-to-noise ratio (PSNR), $PSNR(x)$, was computed as follows:

$$PSNR(x) = \frac{I_p(x)}{\sigma_{p+s}(x)} = \frac{I_{p+s}(x) - I_s(x)}{\sigma_{p+s}(x)} \quad (10)$$

To compute the noise level in the image, two images acquired with identical technical settings (without steel ruler in place) were subtracted from each other to eliminate the signal variations due to anatomical structures. Proper normalization was performed during the subtraction of two images to minimize the signal variations that may arise from potential exposure fluctuation hence the incomplete cancellation of anatomical structures. 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. Five horizontal image lines, including the line where scatter was measured and two lines from each neighboring side were divided into 384 subregions horizontally and each sub-region contained 8×5 pixels. Noise measurement was performed over each sub-region to compute position dependent PSNRs.

2.5 SEDR chest image restoration

Figures 6(a) and (b) show a pre-scan (low dose slot-scan) chest phantom image and SEDR chest phantom image, respectively. It's observed that the signal intensities were enhanced in the heavily attenuating regions of SEDR chest phantom image. However the SEDR image contains artifacts at the joints of two neighboring beam modulators due to the limited spatial resolution of the beam modulators. The chest phantom image needs to be restored to have the appearance of a conventional radiograph. Following the SEDR imaging, a low dose SEDR exposure map was acquired at 120 kVp and 10 mA without the chest phantom in x-ray path. The relationship of image intensities between SEDR chest image and SEDR exposure map can be expressed by following equation:

$$I_{SEDR}(i, j) = k \cdot I_{map}(i, j) \cdot e^{-\mu_{eff}(i, j) \cdot t(i, j)} \quad (11)$$

where I_{SEDR} and I_{map} are the image intensities for SEDR chest image and SEDR exposure map, respectively; μ_{eff} is the effective linear attenuation coefficient of the object (bone and soft tissue for chest imaging); k is a scaling factor. The restored SEDR image is therefore computed as follows:

$$I_{restore} = \mu_{eff}(i, j) \cdot t(i, j) = \log(k \cdot I_{map}(i, j)) - \log(I_{SEDR}(i, j)) \quad (12)$$

The restored SEDR image is logarithm processed and has an appearance of conventional radiography. The SEDR exposure map and restored SEDR chest phantom images are shown in figures 6(c) and (d).

3. Results

Figure 7 shows captured input square waveform applied to the beam height modulation driver and the output measured at the two terminals of linear motor. The output pulse width was measured to be 0.124 s at full width half maximum (FWHM) when a 2 V_{p-p} square waveform (~ 1 Hz) input is applied, resulting in an average modulation response of 16.13 V/s for the beam height modulation assembly.

The maximum signal response calculated from the pre-scan chest image was plotted as a function of filtering kernel size for the seven modulation segments and was illustrated in figure 8. During the SEDR imaging, the beam height modulators are usually operated at a low relative

speed (typically below 4 V/s) while a higher speed is only required at the interface of heavily and lightly attenuating regions. Filtering kernel size of 5 or larger is required to ensure the proper response of beam height modulation assembly. The filtering kernel size of 10 was used in this study.

In figure 9(a) and (b), the position dependent scatter components of the chest phantom image for three different imaging techniques, are plotted for comparison in the lung and subdiaphragm, respectively. Both anti-scatter grid and SEDR techniques reduced the scattered radiation effectively; however SEDR technique demonstrated better scatter reduction than the anti-scatter grid. The scatter signals at the central area are about the same for both anti-scatter grid and SEDR techniques in figure 9(b). This is due to a significant increase of x-ray exposure (about four times higher) in that region for SEDR imaging.

The position dependent SPRs of the chest phantom image for three different imaging techniques, are plotted for comparison in figures 10(a) for the lung and 10(b) for the subdiaphragm. By using slot-scan geometry the SEDR technique resulted in significantly lower SPRs in both lung and subdiaphragm regions compared to the full-field with grid method.

To compare the SPR reduction for two different scatter reduction techniques, the SPR reduction ratios (SPRRRs), defined as the differences between the non-grid full-field SPRs and the reduced SPRs divided by the former, are plotted for comparison in figures 11(a) for the lung and 11(b) for the subdiaphragm. The SPRRRs tended to be higher in the heavily attenuating regions and lower in the lightly attenuating regions. In the lung regions, the SPRs were reduced by approximately 59% for the full-field imaging with anti-scatter grid and 82% for SEDR technique. In the subdiaphragm, the SPRs were reduced by an average of 77% for the full-field imaging with anti-scatter grid and 95% for SEDR technique. The scatter reduction ability of SEDR technique is more pronounced in the heavily attenuating regions than in the lightly attenuating regions. This is because the primary x-ray intensities were largely elevated in the heavily attenuating regions than in the lightly attenuating regions. The variation of SPRRRs, calculated from the differences between maximum SPRRRs and minimum SPRRRs divided by the average SPRRRs in the central five modulation segments are 42.9% and 20.6% in the lung, and 22.7% and 7.5% in the subdiaphragm for full-field with grid and SEDR respectively, indicating a more uniformly distributed SPRRRs with SEDR imaging technique.

The position dependent PSNRs of the chest phantom image for three different imaging techniques, are plotted for comparison in figures 12(a) for the lung and 12(b) for the subdiaphragm. The full-field with grid method produced slightly higher PSNRs over the non-grid full-field method in the heavily attenuating regions. However the change of the PSNRs was reversed in the lung regions. The decreased primary signal and x-ray quantum noise due to the use of an anti-scatter grid have a combined effect on resultant PSNRs. In this study, the magnitude of quantum noise reduction is slightly low than the magnitude of primary reduction in the lung regions. The similar result was observed in a separate study of CNR (Liu *et al.*, 2006; Liu *et al.*, 2008). The SEDR technique demonstrated a substantial improvement in PSNRs over the anti-scatter grid technique. The PSNR Improvement factors (PSNRIFs), defined as the differences between the improved PSNRs and non-grid full-field PSNRs divided by the later, are plotted for comparison in figures 13(a) for the lung and 13(b) for the subdiaphragm. The improvements of PSNRs vary with the regions and are more pronounced with the SEDR technique in the heavily attenuating regions than the lightly attenuating regions. This is due to the combined effect of increased primary signals and reduced x-ray quantum noise from the compensation of x-ray beam intensities in the heavily attenuating regions.

4. Discussion and Conclusions

There are several concerns related to SEDR imaging technique. First, the SEDR technique combines the slot-scan geometry with the x-ray beam intensity modulation to compensate for exposure variations due to the change of the anatomical structure. Any drawbacks involved with slot-scan imaging technique would also apply to the SEDR imaging technique. Research has indicated that the potential pulmonary or cardiac motion artifacts in slot-scan images would be minimal because the effective exposure time in slot-scan is comparable to that of conventional radiography (Samei *et al.*, 2005). However, in SEDR imaging, the situation is different from the slot-scan imaging because two images (pre-scan and SEDR) were acquired at separate times, and hence potential phase differences could exist between the pre-scan and SEDR images. Such phase differences may impact accurate delivery of exposure locally during SEDR imaging but it should not introduce additional artifacts in the final restored SEDR image because the same exposure equalization pattern (calculated from pre-scan image) was used to produce both SEDR chest image and SEDR exposure map.

Secondly, the x-ray tube loading may impact the potential clinical applications of SEDR imaging. We have discussed the x-ray tube loading issue in previously published papers on slot-scan imaging (Liu *et al.*, 2006; Liu *et al.*, 2008). Our conclusion is that it is achievable to image a large size patient (both PA and lateral positions) using a slot-scan imaging technique with current x-ray tube technology. However a trade-off has to be considered in selecting imaging parameters between the readout speed (or slot width) and the tube loading. An increase of tube loading (or mAs) by approximately a factor of 22 is required to maintain the effective primary exposure at the detector input at the same level as with conventional full-field imaging technique without a grid. However for SEDR imaging, an additional increase of the x-ray tube current by 5 to 10 times higher than that of slot-scan imaging is necessary in order to compensate for the anatomical variations. This yields a tube current requirement of 110 – 220 mA to perform SEDR imaging. The x-ray system used in this study has a maximum output of 200 mA at 120 kV_P and 5 S. This should be adequate for imaging a mid size patient. An x-ray system with larger output capacity is required for imaging a large size patient in SEDR imaging. The x-ray dose efficiency for SEDR imaging would be even lower (1/10 to 1/5) than that for slot-scan imaging.

Thirdly, the pre-scan image acquisition contributes to total patient dose in SEDR imaging. In this study, the image quality was compared for chest phantom images for both full-field and SEDR imaging techniques with same average exposure level; however the dose contribution from the pre-scan imaging was not counted into the total patient dose. If a clinical application of SEDR imaging is implemented, exposure from the pre-scan imaging should count into the total patient dose. In general, slot-scan technique provides a dose saving of 30% - 50% while maintaining the image quality comparable to or better than that of anti-scatter grid technique for chest radiography. The same dose saving advantage therefore applies to the SEDR imaging technique as well because the equalization pattern (based on the average exposure to the x-ray detector) was computed from a pre-scan image acquisition using same slot-scan geometry. Hence it is possible to keep the total average patient dose unchanged while maintaining image quality comparable to or better than those in full-field radiography by slightly reducing exposure level in final SEDR imaging. The exposure setting for the pre-scan image acquisition should be high enough to minimize errors of equalization pattern particularly in heavily attenuating regions. In this study, the exposure level for pre-scan imaging is roughly 20% to that of final SEDR imaging. At this exposure level the minimal image signal was measured to be about 40 DU (digital unit). The pre-scan exposure can be further reduced down to 5% of total exposure for modern commercial detectors due to improved sensitivity and detective quantum efficiency. As an example, a pre-scan image and SEDR image acquired at exposures

equivalent to 5% and 95% of that for full-field imaging would maintain patient dose unchanged while producing better detectability of low contrast objects in heavily attenuating regions.

We have demonstrated the scatter rejection and image quality improvement of SEDR imaging technique by measuring the SPRs and PSNR improvement factors and comparing them with those of a conventional full-field imaging technique. In addition to reduction of the SPRs, it is equally important to achieve sufficiently high photon flux at the detector input in heavily attenuating regions of chest images. Along with slot-scan geometry, scanning exposure equalization technique has resulted in improved image quality for detection and visualization of low contrast objects through scatter rejection and increased photon flux in the heavily attenuating regions, despite the spatial variation of attenuating properties in the patient body. This should provide radiologists with chest radiographs with more uniform image SNRs thus lessening the limitations posed by image noises. Given the wide use of chest imaging as an initial diagnostic step for many health problems or concerns, this improvement may help bring about more accurate diagnosis. However, this study was based on a bench-top experimental imaging system using only a 7-channel modulation mechanism. Future improvement will include the development and investigation of a system employing a beam height modulation assembly with 24-channel modulation driver. Such a system would provide finer regional exposure control for SEDR chest imaging. Although SEDR has the potential to improve the SNRs in heavily attenuating regions, the effect on disease detection and diagnosis may not be the same in different anatomical regions, e.g. lungs, mediastinum, retrocardium, and abdomen. Therefore, clinical research is required to determine the effect of improved image SNR, brought about by SEDR, on disease detection or diagnosis in various anatomical regions.

Acknowledgments

This work was supported in part by research grants CA104759 and CA124585 from NIH-NCI, a research grant EB00117 from NIH-NIBIB, and a subcontract from NIST-ATP.

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Figure 1. Schematic drawing of the scan equalization digital radiography (SEDR) system. The system is designed based on an a-Se flat-panel detector and slot-scan geometry.

Figure 2.
X-ray beam height modulation assembly to regulate beam intensity regionally by moving lead beam blockers up and down independently.

Figure 3.
Schematic drawing of the x-ray beam height modulation assembly.

Figure 4. Image signal intensity plotted as a function of beam modulation voltage. The data was curve fitted and interpolated to generate a beam modulation transfer function look up table (BMTF-LUT) for SEDR imaging.

Figure 5.
Flow chart to illustrate the SEDR imaging process.

Figure 6.
(a) Pre-scan (slot-scan) chest phantom image; (b) SEDR chest phantom image; (c) SEDR exposure map; (d) restored SEDR chest phantom image.

Figure 7.
Captured input (channel 2) and output (channel 4) signals from beam height modulation array to estimate the beam height modulation response.

Figure 8.
Beam modulation response plotted as a function of filtering kernel size for different modulator channels.

Figure 9.

The position dependent scatter components of the chest phantom image measured at (a) lung and (b) subdiaphragm for full-field, full field with anti-scatter grid, and SEDR imaging.

Figure 10.

The position dependent SPR of the chest phantom image measured at (a) lung and (b) subdiaphragm for full-field, full field with anti-scatter grid, and SEDR imaging.

Figure 11.

The position dependent SPR reduction ratio (defined differences between the full-field no grid SPRs and the reduced SPRs divided by the former) of the chest phantom image measured at (a) lung and (b) subdiaphragm for full field with anti-scatter grid, and SEDR imaging.

Figure 12.

The position dependent PSNR of the chest phantom image measured at (a) lung and (b) subdiaphragm for full-field, full field with anti-scatter grid, and SEDR imaging.

Figure 13.

The position dependent PSNR improvement factor of the chest phantom image measured at (a) lung and (b) subdiaphragm for full-field, full field with anti-scatter grid, and SEDR imaging.