A real-time regional adaptive exposure method for saving dose-area product in x-ray fluoroscopy

Steve Burion

Triple Ring Technologies, Inc., 39655 Eureka Drive, Newark, California 94560

Michael A. Speidel

Department of Medical Physics, University of Wisconsin-Madison, Madison, Wisconsin 53705

Tobias Funk^{a)}

Triple Ring Technologies, Inc., 39655 Eureka Drive, Newark, California 94560

(Received 3 December 2012; revised 27 March 2013; accepted for publication 31 March 2013; published 24 April 2013)

Purpose: Reduction of radiation dose in x-ray imaging has been recognized as a high priority in the medical community. Here the authors show that a regional adaptive exposure method can reduce dose-area product (DAP) in x-ray fluoroscopy. The authors' method is particularly geared toward providing dose savings for the pediatric population.

Methods: The scanning beam digital x-ray system uses a large-area x-ray source with 8000 focal spots in combination with a small photon-counting detector. An imaging frame is obtained by acquiring and reconstructing up to 8000 detector images, each viewing only a small portion of the patient. Regional adaptive exposure was implemented by varying the exposure of the detector images depending on the local opacity of the object. A family of phantoms ranging in size from infant to obese adult was imaged in anteroposterior view with and without adaptive exposure. The DAP delivered to each phantom was measured in each case, and noise performance was compared by generating noise arrays to represent regional noise in the images. These noise arrays were generated by dividing the image into regions of about 6 mm², calculating the relative noise in each region, and placing the relative noise value of each region in a one-dimensional array (noise array) sorted from highest to lowest. Dose-area product savings were calculated as the difference between the ratio of DAP with adaptive exposure to DAP without adaptive exposure. The authors modified this value by a correction factor that matches the noise arrays where relative noise is the highest to report a final dose-area product savings.

Results: The average dose-area product saving across the phantom family was $(42 \pm 8)\%$ with the highest dose-area product saving in the child-sized phantom (50%) and the lowest in the phantom mimicking an obese adult (23%).

Conclusions: Phantom measurements indicate that a regional adaptive exposure method can produce large DAP savings without compromising the noise performance in the image regions with highest noise. © 2013 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4801908>]

Key words: dose-area product, fluoroscopy, radiation dose reduction, inverse geometry, adaptive exposure

I. BACKGROUND

There has been growing awareness in recent years of the importance of reducing radiation dose in medical imaging with x-rays.^{1–5} While all age groups can benefit from dose savings, we assert that reduction of radiation dose in pediatric cardiac interventions is particularly important for the long-term health of pediatric patients. Children in the age group of 0–15 years account for 7% of all cardiac angiography procedures.⁶ While adults are often diagnosed with and treated for coronary artery disease, children are primarily treated for congenital heart disease. Congenital heart disease is very complex in nature and leads on average to four- to eightfold longer interventional procedure times relative to the coronary artery disease procedures in adults.⁷ To reduce iodine exposure in children and to also improve the ability to discern the coronary vasculature, biplanar systems are preferred in pediatric

interventions,⁸ which can further increase the radiation burden. Further still, children can undergo up to 10 fluoroscopic procedures by the time they reach the age of 10.⁸ Radiation received in the first decade of life leads to a threefold higher risk of developing fatal cancer than the average population, or tenfold higher than older adults.^{8,9} It is estimated that a typical pediatric fluoroscopic procedure leads to eight fatal cancers in 10,000 patients.¹⁰

Dose reduction in interventional pediatric procedures is recognized as a high priority, and all manufacturers of fluoroscopes implement special pediatric protocols in their imaging systems.¹¹ However, the design and geometry of standard fluoroscopic equipment significantly limits the implementation of some dose saving strategies.

In particular, one dose saving strategy that is not easily implementable on standard equipment is regionally adaptive exposure control. Rather than exposing every part

of the image with the same radiation dose, an adaptive system would vary the exposure depending on the opacity of the region exposed. In such a system, exposure can be reduced in the relatively transmissive lung field without altering exposure (and image signal-to-noise ratio) in more opaque regions.

The general idea of adaptive exposure has been proposed before for other applications, such as overcoming limitations imposed by the dynamic range of x-ray film. For example, since the median exposure to the detector behind lungs is about three times larger than behind nonlung regions, Boone *et al.* proposed variable lung templates to be introduced into the x-ray beam to avoid underexposure of nonlung tissue.¹² Other equalization techniques have been reported in the literature. Most notably the advanced multiple beam equalization radiography (AMBER) system used a scanning slit with feedback loop to introduce modulators into the beam depending on the exposure.¹³ This technique was effective but prone to artifacts due to cardiac or patient motion as the mechanical nature of the modulators made the system response relatively slow. Another approach was pioneered by Molloy *et al.*¹⁴ in which an equalization mask was molded based on an acquired preliminary image and inserted into the radiation field. The process took about 7 s.¹⁴ More recently, Bartolac *et al.* showed in a simulation study that dose savings between 30% and 50% can be achieved in CT by using an adaptive exposure method coined fluence field optimization.¹⁵ However, current hardware does not support the implementation of such an approach.

The purpose of medical imaging is to provide the physician with images that have sufficient contrast-to-noise ratio (CNR) to perform the clinical task at hand. In an object with highly varying opacity, the CNR will vary throughout an image if it is acquired with uniform exposure across the field of view. In more opaque areas fewer photons will be detected, thereby increasing relative noise as Poisson noise is given by the square root of the number of detected photons. Thus, CNR in the resulting image can either be sufficient in the dark regions and higher than diagnostically necessary in the bright regions or lower than diagnostically necessary in the dark regions and sufficient in the bright regions. Both situations are undesirable: If dark regions are underexposed, information from these regions cannot be used for diagnosis. If bright regions are overexposed, the patient will receive higher radiation dose than clinically necessary.

Regional adaptive exposure aims to expose the object in a way that opaque and transparent regions yield equal CNR in the final image, thereby producing images using the minimum amount of exposure needed to achieve diagnostically useful CNR. Importantly, by design adaptive exposure lowers the CNR in the very bright regions to a level that is diagnostically sufficient.

While the efficacy of adaptive exposure has been recognized, current imaging systems are not capable of modulating the regional exposure quickly enough to be used in the clinic.^{14,16} However, in this study we show that our scanning beam digital x-ray (SBDX) C-arm fluoroscopy system can be used to implement a real-time adaptive exposure method.

FIG. 1. In a conventional system the image is formed by the projection of a single focal spot x-ray tube onto a large-area detector (left). In the SBDX system an image is formed by combining the projections of the multiple focal spots on a large-area x-ray tube onto a small detector (right).

The SBDX technology deviates significantly from the design of a conventional fluoroscope. An extended discussion on the method and the imaging performance has been published,^{17–19} and here we highlight only some of the unique features of the SBDX technology. Refer to Fig. 1 for a schematic representation.

In contrast to a conventional fluoroscope, in which the x-ray tube has only a single focal spot, the SBDX x-ray tube has an extended emissive surface across which a scanning electron beam dwells sequentially on a 100×100 grid with up to 8000 active focal spot positions. The x-ray tube technology is very similar to a CRT tube, but rather than using a phosphorus target generating visible light from low energy electrons, the SBDX tube uses a tungsten target that generates x-ray photons from high energy electrons. At each focal spot position x-ray photons emitted by the tungsten target are directed toward the detector by use of a focusing collimator, projecting a small view or patch of the imaging volume onto the detector. Each focal spot is circular with trapezoidal profile and measures approximately 0.6 mm full-width-half-maximum.¹⁸

Additionally, while a conventional fluoroscope uses a large integrating detector positioned relatively close to the patient, the SBDX system uses a small photon-counting detector positioned relatively far from the patient. The SBDX detector is a custom 160×320 pixel CdTe detector having a 0.33 mm pixel pitch and a maximal count rate of 5×10^6 counts s^{-1} mm^{-2} . For this study the detector was operated in a 2×2 binning mode with an effective pixel pitch of 0.66 mm. In the SBDX system detector and x-ray source are 150 cm apart with the isocenter 105 cm from the detector.

In a conventional fluoroscope one image is formed on the large area detector. The SBDX system acquires sequential images from up to 8000 projections (detector images), i.e., from up to 8000 focal spots, on the small-area detector. Every projection gives only a very small view of the imaging volume;

the full image is reconstructed by combining these projections using shift-and-add operations.¹⁸

Therefore, while images from a conventional fluoroscope are planar and without 3D information, the SBDX data are inherently tomosynthetic. Every projection samples the object from a slightly different angle and the shift-and-add algorithm reconstructs a unique focal plane of the imaging volume depending on the shifting distance. A complete image is formed by reconstructing a set of focal planes throughout the imaging volume. The stack of planes can be presented to the physician similarly to a tomographic image. To provide a convenient 2D display format, the imaging planes can be collapsed into a 2D image by selecting the objects that are in focus in various regions of the different focal planes.

The unique imaging geometry of the SBDX system allows for a straightforward implementation of an adaptive exposure method as we will discuss in detail in Sec. II. We also present phantom studies that show dose-area-product savings on the order of 40% at equal noise levels in the darkest areas relative to those obtained without adaptive exposure.

II. METHODS

II.A. SBDX standard image acquisition

The SBDX system accommodates a variety of imaging modes with different frame rates, tube voltages, and image sizes. As mentioned earlier the image acquisition in the SBDX system is very different from a conventional x-ray system. For a single frame a large number of detector images are acquired by raster scanning an electron beam across the x-ray target of the SBDX x-ray tube. The beam is parked behind collimator holes that are equally spaced with a pitch of 0.23 cm. In a standard imaging mode, i.e., 7-in. image diameter and 15 Hz repetition rate, the electron beam scans a 71×71 subset of the available collimator holes.

At each collimator hole position a detector image is acquired. These detector images can be pipelined to real-time reconstruction hardware or alternatively saved to a data file (detector data file) for offline reconstruction.

The electron beam needs to be controlled with high precision. Scan parameters such as tube voltage, electron beam deflection parameters, etc. are derived in an alignment procedure. The scan parameters are saved in a look-up table (LUT file) and are accessed whenever an image is acquired. Another feature of the LUT-file is that it specifies at which collimator holes the electron beam is turned on by saving for every collimator hole a beam on/beam off bit. For example, the beam-on/beam-off bit can be used to implement electronic collimation; rather than using physical shutters the electron beam can be turned off at unwanted positions.²⁰

Another important detail is that the electron beam delivers a highly localized thermal load to the x-ray target, leading to instantaneous local heating. To operate at high tube current without melting the target the electron beam has to be moved to the next focal spot position after $1.04 \mu\text{s}$. However, this time is not sufficient to collect enough photons for reasonable image quality in a typical fluoroscopic setting. This problem is solved by revisiting the focal spot at a later time to repeat

the acquisition. Multiple acquisitions lead to improved photon statistics. In a standard 71×71 15 Hz scan each collimator hole is visited eight times. In other words the SBDX system acquires eight low count acquisitions (noise realizations) of the same image that are summed to obtain the final detector image. Each noise realization is referred to as one rescan; hence, eight rescans are performed to obtain the final detector image.

Our adaptive exposure method utilizes the rescan process to regionally control the exposure in the image by turning the beam off in the less attenuating regions during subsequent rescans. During acquisition of the first rescan, the counts in every projection, i.e., from each focal spot position, are summed and input into a 71×71 detector-count matrix. Counts in the detector-count matrix scale linearly with number of subsequent rescans, e.g., after two rescans a given matrix element has approximately twice as many counts as after one rescan. Therefore, the detector count matrix can be either updated after each rescan with measured data or calculated *a priori* using the number of rescans. A detector-count matrix estimated from a single rescan reflects the expected distribution of brightness in the final image or, in other words, the expected noise in the different image regions. Regional adaptive exposure aims to expose areas with low noise (i.e., high brightness) less than those with high noise (i.e., low brightness). The goal is to equalize noise across an image by varying the exposure delivered to regions of an image based on their expected levels of brightness.

After each rescan the detector counts at each element of the detector-count matrix are compared against a threshold to determine if a sufficient number of counts have been acquired at the corresponding focal spot positions. If the number of detector counts is greater than the threshold at any given matrix location, the electron beam will be turned off at the corresponding focal spot position during subsequent rescans.

The threshold setting determines the target image brightness and noise level. As discussed further below many threshold selection approaches are possible and the current approach was delineated experimentally. To define the threshold, first we obtain the intensity distribution in the central region of the detector-count matrix acquired by the first rescan [Fig. 2(b)]. Since the count matrix only reflects intensity after one rescan, we then predict what the count levels will be after all rescans by multiplying the initial count levels by the total number of rescans that will be completed to form the image. Next, we calculate a “raw threshold” from this rescaled matrix, defined as the 30th percentile of the intensity distribution [Fig. 2(c)]. This raw threshold value corresponds to the brightness level in the darker regions of the image. Last, rescan-dependent thresholds are defined. Rather than comparing the detector counts after each rescan against a fixed threshold, the threshold is changed as a function of completed rescans. Thus, to determine if a focal spot position is turned off during the second rescan the detector counts measured in the first rescan are compared against 83% of the raw threshold. Then for subsequent rescans the threshold is increased linearly back up to 100% of the raw threshold at the last rescan [Fig. 2(d)].

FIG. 2. (a) Reconstructed image of the child-sized phantom acquired without adaptive exposure. (b) The detector-count matrix is calculated by integrating the detector counts for each collimator hole. The white outline in the count matrix image indicates the central region that was used to calculate the threshold. (c) The plot shows the integrated detector counts as function of the percentile (only taking collimator holes within the central region into account). The raw threshold is the value at the 30th percentile. (d) The plot shows how the rescan-dependent thresholds are related to the raw threshold. Lower and upper thresholds that are used in the hysteresis algorithm are also shown.

The use of scaled, rescan-dependent thresholds allows focal spots producing relatively bright detector images in the first few rescans to be turned off earlier than if a single, rescan-independent threshold were used as illustrated in the following example. Consider a given detector matrix element producing 90% of the raw threshold in the first rescan. If a single, rescan-independent threshold were used, the beam would not be turned off until after the second rescan, at which point it would have produced 180% of the desired number of counts. Therefore, without rescan dependent thresholds a particular collimator hole can greatly exceed the desired number of counts. However, with the rescan dependent threshold the detector matrix element exceeds the threshold for the second rescan (83% of the raw threshold) and the beam would be turned off after the first rescan at that position.

The approach to setting the threshold, e.g., use of the 30th percentile of the detector count number distribution for the raw threshold and beginning the first rescan dependent threshold at 83% of the raw threshold, is not unique. Our current experience is that while choosing the threshold is important, there is a broad range of possible approaches for choosing the threshold that can give similar dose reductions. We also found that a single approach cannot necessarily yield the greatest DAP savings for all the different phantom sizes. We selected our approach based on two primary criteria: (1) that the approach is simple; and (2) that the approach produces large dose reductions for all phantoms, even if not the greatest possible for a given phantom. These criteria ensure that the approach can be implemented in real-time application of regional adaptive exposure. We believe that our current approach is a good starting point but that over time and with more data further refinements will be possible.

One complication in the implementation of regional adaptive exposure is that the current beam scan hardware only allows for a limited number of switches between beam-on and beam-off states during a given acquisition. To accommodate this limitation we are using a lower and an upper threshold to introduce hysteresis in the switching process. The lower and upper thresholds are simply predetermined percentages of the rescan-dependent thresholds. For example, if a rescan-dependent threshold is x then the lower threshold could be

0.9 x and the upper threshold could be 1.1 x [Fig. 2(d)]. Essentially our method takes into account whether the beam was on or off prior to a switch of beam state. If the beam was on at the previous collimator hole then the detector count is compared against the higher threshold and is therefore less likely to be turned off. In contrast if the beam was off in the previous collimator hole then the detector count is compared against the lower threshold and is therefore more likely to stay off. Setting of the lower and upper threshold has to be optimized. Having a large span between lower and upper threshold will reduce the number of beam on/beam off switches considerably but comes at the price of being less responsive to variations in the detector counts and consequently less effective to adapt the exposure with brightness. Too small of a span between lower and upper threshold, though, does not accomplish the goal of reducing the number of switches. In the current implementation an exposure map and the number of beam-state switches is calculated using a lower and upper threshold at $\pm 10\%$ of the main threshold. If the number of beam-state switches exceeds the allowable maximum number of switches the calculation is repeated using a larger range. In our experiments we found that most experiments were possible at a range of $\pm 10\%$ of the main threshold. However, two experiments had to be performed at a higher range (Table II). We recognize that a variable range for lower and upper threshold is not desirable and we plan to upgrade our hardware to enable more switches and potentially perform all scans with a single threshold.

II.B. Implementation of the regional adaptive exposure technique

In summary, our implementation of adaptive exposure will consist of acquiring a single rescan image; determining a raw threshold; calculating the rescan-dependent thresholds; splitting the rescan-dependent thresholds further into lower and upper thresholds; comparing the detector counts from each focal spot to the appropriate threshold; and turning the beam on or off accordingly. In a real-time implementation these steps have to be performed while the imaging frame is being acquired. However, conceptually these steps can be divided

into an algorithmic stage and an execution stage. All computation is performed in the algorithmic stage using data of the first rescan. The result is an exposure map that details if a focal spot is turned on or off during image acquisition. The exposure map is a matrix of integer values between 1 and 8. A value of 1 indicates that at the given collimator hole the electron beam is only turned on during the first rescan. A value of 2 specifies that the beam is on during the first two rescans and so on. The beam-on/beam-off bit in the LUT file is modified for each rescan according to the exposure map. In the execution stage the modified LUT file is used to acquire adaptive exposure images.

This distinction is important as it facilitates the development of adaptive exposure. The algorithmic stage can be performed offline using an image acquired prior to the adaptive exposure acquisition and the resulting LUT file can be uploaded to the system for the execution stage. The advantage of this is that the algorithm can be developed using the convenience of high-level programming languages. At the same time the resulting exposure maps can be tested on the actual scanning hardware with all the limitations of an actual system. Therefore, it is possible to measure realistic image noise and dose savings.

II.C. Experimental setup

We performed experiments on an anthropomorphic phantom family mimicking a 1-year-old, a 10-year-old, and a 20-year-old male (CIRS, Norfolk, VA). In addition we performed experiments on a phantom of a 20-year-old male with an iodinated pig heart (Fig. 3). We added an anthropomorphic fat layer and 5-cm thick Lucite layer to that phantom to mimic an obese male.

All phantoms were imaged in anteroposterior (AP) position with the center of the heart centered in the field of view and positioned at the isocenter of the SBDX system. We have chosen exposure levels to produce contrast to noise levels achieved in clinical fluoroscopic exposures. Aluminum filtration was added in the beam path to reproduce beam hardness used in clinical settings. In the case of the pediatric phantoms the aluminum filtration is denoted filter A. In addition we used an erbium filter of 150 μ m thickness in the two pediatric

FIG. 3. Phantom family used in this study. The black phantom has an iodinated pig heart and is here shown with the additional fat layer (obese adult).

phantoms to vary the beam hardness with the goal of changing the dynamic range of the fluoroscopic image. The erbium filter is denoted filter B. Key characteristics of the phantoms, exposure parameters, and beam hardness are summarized in Table I.

In each experiment, a phantom was positioned and a single frame was first acquired using the standard, i.e., nonadaptive, exposure technique. The detector file and the associated LUT-file were then downloaded. The previously discussed algorithm was implemented in Matlab (MathWorks, Natick, MA). The detector-count matrix and thresholds were calculated from the detector file and used to derive an exposure map. As discussed earlier, the exposure map was used to generate modified LUT-files, which were then used to acquire data.

We then acquired 100 frames with adaptive exposure using the modified LUT-file and 100 frames without adaptive exposure. Immediately after acquisition we positioned a dose-area product meter (PDC, Radcal Corporation, Monrovia, CA) in front of the collimator to measure the dose-area product with and without adaptive exposure. The active area of the DAP meter was 30 $\times$ 30 cm and covered the exposure area of

TABLE I. Summary of phantom and exposure parameters.

Phantom	Simulated age group	AP-thickness (cm)	Voltage (kVp)	Tube current (mAp)	Power (kWp)	Exposure (R/min)	Entrance HVL (mm Al HVL)
Infant (filter A)	1	12	80	75	6	1.1	5.3 ^a
Infant (filter B)	1	12	80	75	6	0.4	5.9 ^b
Child (filter A)	10	17.5	80	75	6	1.1	5.3 ^a
Child (filter B)	10	17.5	80	150	12	0.8	5.9 ^b
Adult	20	23	80	209	16.7	6.2	3.6 ^a
Adult with iodine	20	23	80	209	16.7	6.2	3.6 ^a
Adult with iodine (perp)	20	23	80	209	16.7	6.2	3.6 ^a
Obese adult with iodine	20	29.5	120	200	24	8.5	5.4 ^a

^aHVL measured.

^bHVL derived from simulated spectrum.

FIG. 4. Steps to calculate the noise array. The figure shows images for the child-sized phantom with standard acquisition (top row) and regional adaptive exposure (bottom row). (a) Reconstruction of one frame of the 100 frames taken for each mode. (b) The subtraction image is derived by subtracting two subsequent frames. For clarity we show here the normalized subtraction image that was derived by dividing by the sum of the two frames. (c) The regional noise matrix is derived by dividing the subtraction image into a grid of 15×15 pixel regions. For each region the mean is calculated and placed as an element in the noise matrix. (d) The noise array is derived from the noise matrix. The array consists of elements of the noise matrix that are sorted in descending order.

the SBDX system, which is approximately 16×16 cm for a 7 in. scan at the collimator surface and approximately 12×12 cm at the isocenter of the system. For the DAP measurements we acquired data for 30 s, which is sufficient to measure the dose-area product with better than 0.5% accuracy.

Detector data files were downloaded and reconstructed on a frame by frame basis using a reconstruction algorithm outlined by Speidel *et al.*¹⁸ Imaging planes were reconstructed at 45 cm (the system isocenter), 40 cm, and 50 cm from the source (z -height).

II.D. Analysis

CNR is a measure of detectability and is commonly used to evaluate image quality. Typically CNR is measured at one particular position of the image, e.g., the center of the image. CNR is given by

$$\text{CNR} = \frac{C}{\delta\sigma},$$

with C being the contrast and $\delta\sigma$ the relative noise in the image.²¹ Contrast can be measured by placing a contrast disk, e.g., of iodine, onto a uniform background and calculating the signal with (N_c) and without (N) the disk:

$$C = \frac{N - N_c}{N}.$$

Contrast depends on a variety of factors such as beam quality and detector response.

The relative noise $\delta\sigma$ is determined by the standard deviation of the detected counts divided by the mean of the detected counts:

$$\delta\sigma = \frac{\sigma}{N}.$$

This approach works well with uniform images. For nonuniform images pairwise subtraction of subsequent images can be performed leading to noise images with the actual signal removed. We discuss this method in more detail later.

For the following we assume that adaptive exposure will affect the noise ($\delta\sigma$) in the image but not the contrast (C). This assumption is justified as adaptive exposure is achieved by adding more or less rescans at any given collimator hole position. As discussed earlier each rescan constitutes a low count acquisition of the same image and therefore the composite image does not alter contrast as it is a relative quantity, normalized by the average number of counts. In the following we have omitted the contrast entirely from the discussion and only compare relative noise.

Since adaptive exposure changes the exposure delivered to an imaging volume on a regional basis, we are interested in comparing the regional variations of the relative noise $\delta\sigma$ in images acquired with and without adaptive exposure.

Figure 4 illustrates the approach that we have chosen to determine and compare regional noise variation. As a first step we reconstructed the 100-frame sets that were taken for each experiment. We then calculated 50 independent subtraction images by pairwise subtraction of successive image frames. Every subtraction image was divided into J nonoverlapping

regions of 15×15 pixels in size, each region approximately covering an area of 5.8 mm^2 . In each region of the subtraction images we calculated the standard deviation $\sigma_{i,j}$. The index i runs over the number of subtraction images and the index j over the number of regions. The reconstructed frames were also divided into the same 15×15 pixel regions, and the mean signal ($N_{i,j}$) in each was calculated. Finally we calculated the mean relative noise $\delta\sigma_j$ for each region j :

$$\delta\sigma_j = \frac{\text{mean}_i(\sigma_{i,j})}{\text{mean}(\text{mean}_i(N_{2i-1,j}), \text{mean}_i(N_{2i,j})) \cdot \sqrt{2}},$$

where mean_i indicates that the mean was taken across imaging frames. The relative noise $\delta\sigma_j$ for each region in a given frame was stored in a noise matrix. The noise matrix has higher values in regions with high noise and lower values in regions with lower noise. Regions with high noise correspond to opaque regions of the phantom and regions with low noise correspond to transparent regions of the phantom. For that reason the image of the noise matrix [Fig. 4(c)] appears to be a blurred version of the actual image.

Rather than comparing standard and adaptively exposed frames region by region, we generated 1D noise arrays for comparison. The arrays were populated with the elements of the noise matrix, sorted in descending order. We then plotted these arrays, representing regional noise with and without adaptive exposure. In the plots [Fig. 4(d)] dark regions, i.e., regions which exhibit large relative noise, have a small abscissa and bright regions, i.e., regions with small relative noise, have a large abscissa. As discussed earlier, the premise of adaptive exposure is that maintaining the noise performance in the dark regions and lowering the noise performance in the bright regions can optimize the balance between exposure and CNR. In terms of our plot that means the curves should track at small abscissas and diverge at large abscissas; the curve for standard imaging should decrease faster than the curve with adaptive exposure.

To evaluate dose savings it is important to compare images that have equal image quality in the low performing regions of the image. Therefore, if the plots of the noise arrays for the two acquisition methods were not matched in the low performing regions we fitted a correction factor D_x to the data. We included a percentage x of the opaque regions to perform a least square fit using

$$\sum_{j=0}^{xN} \left(\delta\sigma_S(j) - \frac{1}{D_x} \cdot \delta\sigma_A(j) \right)^2,$$

with N being the total number of regions, S denoting regions without adaptive exposure, and A regions with adaptive exposure. For accurate dose-area product savings comparison we assert that 20% of the low performing regions in the image with adaptive exposure should be comparable to the noise performance of 20% of the low performing regions in the image without adaptive exposure. This percentage value is essentially reflecting the minimum CNR at which the image is diagnostically valuable. Choosing this value has a certain subjectivity, and further observer studies will be necessary to establish equivalence of the diagnostic value of images with

and without regional adaptive exposure. To account for this fact we performed fits for $x = 10\%$, 20% , and 30% .

The correction factor D_x can be rationalized as a virtual increase in tube current. The overall noise performance of the image with adaptive exposure could be improved by acquiring the image with higher tube current. To improve the noise performance by D_x we would have to increase the tube current by $(D_x)^2$ as our noise is governed by Poisson noise. Consequently the dose to the patient would go up by $(D_x)^2$.

Having implemented an appropriate correction factor, we calculated a corrected dose-area product saving δD as

$$\delta D = 1 - \frac{\text{DAP}_A \cdot D_x^2}{\text{DAP}_S},$$

with DAP being the measured dose-area product.

Sorted noise arrays were calculated for three different z -heights (40, 45, and 50 cm). This is an important test as the overlap of the projections in the image changes with distance from the source. In the most extreme case of an imaging plane directly at the detector all projections overlap completely and the reconstructed image is just the sum of all projections. Adaptive exposure would not be possible in that case. However, with realistic source distances projections overlap only partially. Nevertheless it is expected that the noise performance will depend on the z -height.

III. RESULTS

Figures 5 and 6 show the eight cases that we have studied ranging from an infant to an obese adult. Pediatric cases are shown in Fig. 5 and adult cases in Fig. 6. The reconstructed images are windowed to the individual minima and maxima but are otherwise presented without any further post-processing. We observe that adaptive exposure is also an effective tool for dynamic range compression for the reconstructed images. In the reconstructed images the regions with highest relative noise are marked by white circles. Without adaptive exposure the highest relative-noise regions coincide with the darkest areas of the images as expected by count statistics. With adaptive exposures the highest relative-noise regions are more distributed across the image and some of these regions coincide with areas where exposure map indicates that fewer than eight rescans were completed. The plots of sorted noise arrays for each case confirm these results; these plots show that with adaptive exposure, the relative noise in the image is flattened, or equalized, across a greater percentage of image pixels compared to images without adaptive exposure. This effect is particularly pronounced in the pediatric phantoms and for the nonobese, noniodinated adult phantom.

The exposure maps for all phantoms show a concentration of eight rescans occurring around the spine and the heart as could be expected from the anatomy. Heart and spine are highly attenuating while the surrounding lung tissue is much less opaque. Interestingly, for the adult phantoms, particularly the case simulating the obese patient, a relatively large area of the overall exposure map is exposed with the maximum

FIG. 5. Summary of the pediatric cases studied. For every case four plots are shown. Reconstructed image at isocenter without adaptive exposure (upper left) and with adaptive exposure (upper right). The white circles show the location of lowest performing regions (below the 5th percentile). Plot of the sorted noise array for the experiments with and without adaptive exposure (lower left). In addition the corrected (including regions below the 20th percentile) curve is shown. Exposure map used in the adaptive exposure experiments (lower right). White indicates eight rescans and black indicates 0 rescans; increasing gray levels indicate decreasing numbers of rescans.

or close to the maximum of rescans. This leads to a more uniform exposure map, and hence to less DAP savings. We attribute these high numbers of rescans to the fact that the additional fat layer decreasing the relative difference between lung and more opaque areas.

Figures 7 and 8 summarize results for the dose savings obtained in this study. Figure 7 focuses on the dose savings obtained at the isocenter after applying correction factors fitted to the regions below the 10th, 20th, and 30th percentile. We find that there is a relatively small spread between the

three different correction factors, indicating a relatively small dependence on the particular choice of the percentile level. Figure 8 focuses on differences in dose savings for different reconstruction planes, i.e., 45, 40, and 50 cm from the source, all using a 20th percentile-fitted correction factor. The overall change with reconstruction plane is also relatively small. Overall, we observe a trend that dose savings get smaller with increasing distance from the x-ray source. This behavior may be expected as the overlap of the detector images increases with increasing distance from the source.

FIG. 6. Summary of the adult cases studied. For every case four plots are shown. Reconstructed image at isocenter without adaptive exposure (upper left) and with adaptive exposure (upper right). The white circles show the location of lowest performing regions (below the 5th percentile). Plot of the sorted noise array for the experiments with and without adaptive exposure (lower left). In addition the corrected (including regions below the 20th percentile) curve is shown. Exposure map used in the adaptive exposure experiments (lower right). White indicates eight rescans and black indicates 0 rescans; increasing gray levels indicate decreasing numbers of rescans.

In Table II the results are summarized. The table shows that the number of switches between beam states is kept near 1200. In some cases it was even necessary to go lower than that boundary to avoid hardware failures. As discussed earlier this boundary is necessary due to limitations in the imaging hardware. However, the range between lower and upper threshold for the hysteresis algorithm is larger for the adult phantom with the iodinated heart. This phantom has a larger variation in contrast than the other phantoms, leading the algorithm to faster switching of beam states. This is counter-

acted by a larger range in the hysteresis algorithm. The difference between hysteresis ranges could explain the difference in dose saving between adult and adult with iodinated heart.

Table II also states the measured and the corrected DAP saving at the isocenter including regions below the 20th percentile. The average corrected DAP savings is $(42 \pm 8)\%$. The table also includes the smallest corrected DAP savings obtained within all three reconstruction planes and all three percentiles. This value is a lower boundary on the DAP saving that is possible with our technique.

FIG. 7. Comparison of dose savings for the eight cases studied. Black indicates the measured dose saving and the white and gray bars show the dose saving after the data were corrected to include regions below the 10th, 20th, and 30th percentiles.

IV. DISCUSSION

In this study we find that using regional adaptive exposure with chest imaging the dose-area product is on average reduced by 42% while a clinically relevant CNR is maintained. We also determined that even assuming less favorable comparison metrics, e.g., reconstruction closer to the detector or

matching regions up to the 30th percentile, the results still show sizable dose-area product savings ranging from 21% to 42%.

Our data indicate that the highest dose savings are possible on relatively small patients. Thus, this technique is particularly well-suited to provide dose savings in pediatric patients—the most vulnerable patient population.

FIG. 8. Dose savings corrected at the 20th percentile level for different reconstruction heights (at isocenter and ± 5 cm from isocenter).

TABLE II. Summary of results obtained in the eight cases studied.

Phantom	Number of on/off switches	Range for hysteresis (%)	Measured DAP saving (%)	Corrected DAP saving (%)	Smallest DAP saving (%)
Infant (filter A)	1260	10	51	45	38
Infant (filter B)	1102	10	54	49	40
Child (filter A)	1180	10	60	49	41
Child (filter B)	1172	10	62	50	42
Adult	1194	10	51	45	40
Adult with iodine	1138	20	40	28	26
Adult with iodine (perp)	930	20	45	30	26
Obese adult with iodine	1126	10	40	23	21

For larger patients DAP savings seem to be somewhat smaller. However, as pointed out earlier this trend is not quite conclusive as the phantom used in the study of larger patients had an iodinated heart resulting in greater variability in contrast distribution. This in turn required us to operate with more hysteresis, potentially reducing the possible DAP saving. Nevertheless, even in these cases DAP savings of at least 20% are possible.

Dose-area product is a good predictor of effective dose to the patient,¹⁰ and it is clear that a reduction in dose-area product translates into dose savings. However, it should be pointed out that in our study the reduction in dose-area product is a result of a highly nonuniform exposure pattern and further investigation is necessary to establish how much the effective dose is reduced with adaptive exposure.

Regional adaptive exposure has been completely implemented using a varying number of rescans with each rescan using the same beam settings, such as beam current and tube voltage. However, it is conceivable that similar results could be obtained by varying the beam settings in different rescans. For example, if beam current could be modulated then the dynamic range of the signal seen by the detector would be reduced, thereby enabling the use of a detector with less stringent requirements for dynamic range.

We do not expect use of regional adaptive exposure to affect the spatial resolution of the SBDX system. Spatial resolution of the SBDX system is mainly determined by focal spot size and detector resolution, parameters which are unaffected by the number of rescans utilized to acquire an image or region of an image. An image consisting of just one rescan should exhibit the same spatial resolution as a composite image of eight rescans.

Our current experiments are performed on a limited number of angulations of the C-arm. We found that it was impossible to mimic other clinically relevant angles without introducing unrealistic shadowing in the images due to missing extremities in the phantoms. We therefore limited this study to the AP position. We tried to span the study over a wide variety of patient sizes with a particular emphasis on pediatric patients who undoubtedly would benefit the most from significant dose savings in fluoroscopic procedures. However, we are aware that this study can only serve as an initial proof of

concept and further validation of the efficacy of our approach is needed in a human trial.

The present study demonstrates that it is possible to do real-time adaptive exposure. As discussed earlier the problem can be separated into an algorithmic and an execution stage. We have a real-time implementation of the execution stage showing that the x-ray fluence can be varied electronically and the variations are fast enough to achieve real-time imaging frames with non-uniform exposure.

The next step in our research effort is to port the algorithmic stage into hardware to enable real-time computation of exposure maps. Fundamentally our algorithm does not present barriers to this porting, and we do not expect any further hardware limitations for the full implementation of the adaptive exposure technique.

A limitation that we found with our current hardware is the limited number of switches between beam-on and beam-off states. Introduction of hysteresis makes it possible to perform adaptive exposure within the limits of our current hardware. However, for future implementations it would be advantageous to increase the number of possible switches. This would improve the flexibility of the system and potentially also allow for even higher dose savings.

The successful real-time implementation of adaptive exposure will offer a new avenue to save dose in x-ray imaging. We envision adaptive exposure becoming a natural part of the fluoroscopic imaging chain and requiring minimal user input.

In this study we have discussed the potential of real-time adaptive exposure for fluoroscopic applications, but we see also potential for CT. As mentioned earlier, Bartolac *et al.*¹⁵ found in a simulation study that large dose savings are possible in CT by varying the regional fluence in the projections—adaptive exposure can achieve exactly that. Recently, the use of scanning beam x-ray sources has been proposed to be used in computed tomography,^{22–24} and we believe that in such CT systems adaptive exposure can be readily applied.

In conclusion, we have presented a method for adaptive exposure and demonstrated the use of hardware to achieve adaptive exposure. We have used a family of anthropomorphic phantoms to acquire and analyze real-time fluoroscopic images and have shown that DAP savings on the order of 40% can be achieved with adaptive exposure.

ACKNOWLEDGMENTS

The authors like to thank Sabrina David, Dr. Phillip Moore, Dr. Brian Wilfley, and Dr. Michael Van Lysel for insightful discussions and advice during this project. This work was primarily funded through a National Institutes of Health (NIH) Challenge Grant (Grant No. RC1 HL100436). Additional support for this work was provided by NIH Grant No. R01 HL084022.

^{a)} Author to whom correspondence should be addressed. Electronic mail: tobiasfunksf@gmail.com

- ¹P. P. Dendy, "Radiation risks in interventional radiology," *Br. J. Radiol.* **81**(961), 1–7 (2008).
- ²M. V. Marx, "The radiation dose in interventional radiology study: Knowledge brings responsibility," *J. Vasc. Interv. Radiol.* **14**(8), 947–951 (2003).
- ³M. S. Stecker *et al.*, "Guidelines for patient radiation dose management," *J. Vasc. Interv. Radiol.* **20**(7 Suppl.), S263–S273 (2009).
- ⁴D. L. Miller *et al.*, "Radiation doses in interventional radiology procedures: The RAD-IR study Part II: Skin dose," *J. Vasc. Interv. Radiol.* **14**(8), 977–990 (2003).
- ⁵D. L. Miller *et al.*, "Radiation doses in interventional radiology procedures: The RAD-IR study Part I: Overall measures of dose," *J. Vasc. Interv. Radiol.* **14**(6), 711–727 (2003).
- ⁶UNSCEAR 2000, "The united nations scientific committee on the effects of atomic radiation," *Health Phys.* **79**(3), 314–329 (2000).
- ⁷L. C. Martinez *et al.*, "Patient doses from fluoroscopically guided cardiac procedures in pediatrics," *Phys. Med. Biol.* **52**(16), 4749–4759 (2007).
- ⁸K. J. Strauss, "Pediatric interventional radiography equipment: Safety considerations," *Pediatr. Radiol.* **36**(Suppl. 14), 126–135 (2006).
- ⁹D. L. Preston *et al.*, "Solid cancer incidence in atomic bomb survivors exposed in utero or as young children," *J. Natl. Cancer Inst.* **100**(6), 428–436 (2008).
- ¹⁰K. Bacher *et al.*, "Patient-specific dose and radiation risk estimation in pediatric cardiac catheterization," *Circulation* **111**(1), 83–89 (2005).

- ¹¹D. Stueve, "Management of pediatric radiation dose using Philips fluoroscopy systems DoseWise: Perfect image, perfect sense," *Pediatr. Radiol.* **36**(Suppl. 14), 216–220 (2006).
- ¹²J. M. Boone, J. Duryea, and E. H. Moore, "Filter wheel equalization in chest radiography: Demonstration with a prototype system," *Radiology* **196**(3), 845–850 (1995).
- ¹³H. Vlasbloem and L. J. Kool, "AMBER: A scanning multiple-beam equalization system for chest radiography," *Radiology* **169**(1), 29–34 (1988).
- ¹⁴S. Molloy, A. Van Drie, and F. Wang, "X-ray beam equalization: Feasibility and performance of an automated prototype system in a phantom and swine," *Radiology* **221**(3), 668–675 (2001).
- ¹⁵S. Bartolac *et al.*, "Fluence field optimization for noise and dose objectives in CT," *Med. Phys.* **38**(Suppl. 1), S1–S17 (2011).
- ¹⁶T. Xu *et al.*, "Area beam equalization: Optimization and performance of an automated prototype system for chest radiography," *Acad. Radiol.* **11**(4), 377–389 (2004).
- ¹⁷M. A. Speidel *et al.*, "Comparison of entrance exposure and signal-to-noise ratio between an SBDX prototype and a wide-beam cardiac angiographic system," *Med. Phys.* **33**(8), 2728–2743 (2006).
- ¹⁸M. A. Speidel *et al.*, "Scanning-beam digital x-ray (SBDX) technology for interventional and diagnostic cardiac angiography," *Med. Phys.* **33**(8), 2714–2727 (2006).
- ¹⁹S. Burion and T. Funk, "X-ray dose reduction through adaptive exposure in fluoroscopic imaging," *J. Vis. Exp.* **55**, e3236 (2011).
- ²⁰S. Burion *et al.*, "X-ray dose reduction by adaptive source equalization and electronic region-of-interest control," *Proc. SPIE* **7961**, 79612B (2011).
- ²¹A. Panse *et al.*, "The micro-angiographic fluoroscope (MAF) in high definition (HD) mode for improved contrast-to-noise ratio and resolution in fluoroscopy and roadmapping," *IEEE Nucl. Sci. Symp. Conf. Rec.* **1997**, 3217–3220 (2010).
- ²²S. David *et al.*, "Experimental validation of an OSEM-type iterative reconstruction algorithm for inverse geometry computed tomography," *SPIE Proc.* **8313**, 83133M (2012).
- ²³S. S. Hsieh *et al.*, "An inverse geometry CT system with stationary source arrays," *SPIE Proc.* **7691**, 76911W (2011).
- ²⁴T. G. Schmidt *et al.*, "An inverse-geometry volumetric CT system with a large-area scanned source: A feasibility study," *Med. Phys.* **31**, 2623–2627 (2004).