Clinical apparatus for the reduction of dose area product for patients undergoing x-ray catheterization

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Purpose: The authors describe a design for prepatient region of interest attenuators (ROIAs) to reduce dose area product (DAP) for clinical use. The authors describe a model to predict DAP values from x-ray technique parameters recorded during a clinical procedure for image sequences obtained in the presence or absence of ROIAs. The model was developed primarily to determine what the DAP to a patient undergoing cardiac catheterization with a ROIA would have been if no ROIA had been used allowing a determination of DAP reduction.

Methods: Copper ROIAs with thicknesses that vary gradually so as not to cause significant image artifacts were constructed. X-ray image sequences were acquired on a clinical catheterization system with and without ROIAs with varying x-ray technique parameters. DAP values were measured for all said exposures using an ionization chamber and compared to a model the authors developed.

Results: The model can predict DAP values within 3.5% on average with or without ROIAs when compared to ionization chamber measurements.

Conclusions: The proposed experimental design is adequate for measuring DAP reductions on the order of 1.5–3.5 that are expected when introducing a ROIA during patient catheterization imaging.

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Key words: dose area product reduction, x-ray attenuator, region of interest x-ray attenuator, interventional imaging, fluoroscopy, cardiac catheterization, percutaneous intervention

1. INTRODUCTION

Interventional x-ray catheterization procedures result in some of the highest radiation doses of any commonly performed x-ray examination.¹ Furthermore, the development of new percutaneous devices and treatment procedures has resulted in a large increase in the use of x-ray guidance to perform minimally invasive therapies. Doses and dose area products (DAPs) vary widely among different interventional procedures and also within equivalent procedures.^{2–5} The peak skin dose can reach a value of 2 Gys within 1 h of fluoroscopy causing radiation injuries⁶ and varies greatly because of its strong dependence on patient table position and imaging angulations that change between image acquisitions.⁴ Exposure to medical personnel is also of concern due to the large number of procedures performed. For interventionists, the eye is a sensitive target for radiation requiring special protection.⁷ Electrophysiologists can on average perform 155 procedures/yr before attaining their yearly radiation exposure limits.⁷

Modern x-ray systems are optimized to maximize the ratio of image contrast for features of interest relative to image noise subject to radiation exposure constraints. This is achieved by adjusting the x-ray tube current to control radiation fluence,

adjusting the x-ray energy spectrum via changes in kilovoltage and the amount of copper filtration. If x-ray noise is the dominant source of image noise, further reduction in x-ray exposure can only be achieved with an associated reduction in contrast to noise ratio. However, it is possible to reduce x-ray exposure while maintaining diagnostic image quality if certain parts of the images are less important or generally exhibit features from which critical information can still be observed using less radiation. Rudin⁸ as well as Labbe *et al.*⁹ have proposed and developed prepatient region of interest x-ray attenuators that attenuate radiation outside a circular region located in the center of the image and their results suggested that a reduction in DAP of 70% is achievable. Region of interest attenuators (ROIAs) cause darkening of the image in the periphery, which was corrected with an x-ray attenuation map. The construction of the map requires knowledge of the attenuator thickness at every irradiated point. Xu *et al.*¹⁰ proposed a method to generate arbitrarily shaped ROIAs and demonstrated its applicability with an anthropomorphic chest phantom. Robert *et al.*¹¹ proposed a ROIA design with a gradual transition in attenuation between the center of the image and the periphery allowing corrections for darkening of the image in the periphery to be done via digital filtering.

ROIA prototypes have undergone limited clinical testing and thus many questions regarding their clinical use have not been resolved. Difficulties in obtaining direct measurements of reductions in DAP and dose as well as modeling of the effect of ROIs may explain why these questions have not been addressed. The purpose of this paper is to describe ROIs and validate DAP measurements and modeling methods for use in a patient clinical trial that will determine the clinical utility/efficacy of ROIs in x-ray catheterization procedures.

2. BACKGROUND

Our goal is to measure reductions in DAP to patients and dose to medical personnel during clinical catheterization procedures as a result of the use of ROIs. A pilot study, approved by the Institutional Review Board (IRB) of our institution, was performed with ten patients to identify the challenges in performing radiation measurements in a clinical environment. The study as well as simple attenuation calculations of the effect of ROIs suggested a two to threefold reduction in DAP for sequences whereby the ROI is used.

Figure 1 is a histogram of DAP values for 1030 catheterization procedures performed in our catheterization laboratory over 1 yr. The data show large variability in DAP values. Given our resources limiting the number of patients to approximately 70, we concluded that a study in which patients are randomized preprocedure between a conventional and ROI procedure with the goal of seeing differences in DAP between the two groups may not achieve significance.

Thus, we propose a model that will allow the estimation of DAP reductions as a result of the use of ROIs using fewer patients than would be required using a conventional comparison of DAP between a control and ROI group. The model is meant to be used with data from clinical cases as its input and not to circumvent the need for clinical data.

FIG. 1. A histogram showing the distribution of DAP values per examination for 1030 patients undergoing cardiac catheterization. The mean and standard deviation of the DAP values are almost equal highlighting the large variation in DAP between examinations.

For each clinical imaging sequence, the proposed model can provide estimates of the patient DAP based on x-ray technique parameters obtained via catheterization system logs with and without a ROI irrespective of whether or not a ROI was actually used. The model is being developed primarily to answer this question for patient image sequences for which a physical ROI was used: “What would the DAP have been for a given ROI image sequence in our study if no ROI had been used?” The answer to the said question which allows the determination of DAP reduction for an image sequence or patient study may not be as powerful as the answer one would obtain from a traditional “treated” versus “control” trial, but it will determine, using fewer patients, whether ROIs have merit for use in interventional imaging. This approach will also allow us to look at DAP reductions on a finer basis, that is per imaging sequence.

The proposed model does not directly address the problem of estimating personnel dose measurements. The addition of a ROI reduces DAP in a predictable manner and it has been shown in a clinical environment that DAP values added over many imaging procedures correlate to dose measurement made at a fixed location in the vicinity of the x-ray equipment¹² as expected from radiation physics. The DAP-dose dependency only manifests itself in a clinical environment by averaging over many exposures due to several uncontrolled variables, including, but not limited to operator position and gantry orientation. However, if one could repeat a clinical procedure identically with the exception of using half the DAP by reducing the tube current, the dose to the operator would be reduced by a factor of 2. In our study, the ROIs favor the transmission of higher energy photons, which have a higher likelihood of being scattered and exiting the patient. Thus, for example, a twofold DAP reduction due to a ROI would result in a dose reduction of less than two.

3. METHODS

3.A. Region of interest attenuators

ROIs of varying physical dimensions were machined from copper sheets on a computer-controlled lathe. The design can be seen in Fig. 2 where the attenuator is secured to a Lucite

FIG. 2. A removable mount which holds the DAP chamber, ROI attenuator, as well as OSL dosimeters.

FIG. 3. Plots of ROIA thickness versus radial distance for the thinnest and thickest ROIs used in our study and those with the smallest and largest apertures. A copper ROI attenuator is shown in the inset. Prescribed ROIA aperture diameters decrease as the numeric component increases, so the ROIA has greater attenuation potential. An increasing alphabetic component indicates a greater prescribed ROIA thickness and again greater attenuation potential. ROIA thicknesses and apertures can deviate from prescribed values due to machining challenges.

cassette, which is partially inserted into a mount that locks the said cassette into place. This mount was developed to test our concept with minimal impact on clinical equipment allowing its removal. The prototype is designed to ease the introduction and removal of a ROIA because the operator may choose not to use it for the entire procedure. It may also be desirable to change the ROIA during the procedure to increase or decrease the DAP reduction.

The ROIs exhibit a central region with no copper and hence no attenuation. Figure 3 is a plot of ROIA thickness as a function of radial distance for four models having different characteristics. Thicknesses were measured on a milling machine by placing a dial test indicator into the collet and translating the milling bed. The copper discs were pressed flat using machinist parallels clamped to the table. We verified

that the discs were in contact with the table by pressing them manually onto the table and probing visually for deflections. The ROIs exhibit gradual changes in thickness so as to limit the severity of image artifacts caused by their introduction. As we move radially outward starting at the center of the fovea, a very thin (~ 0.25 mm thickness) copper region is encountered which is as thin as we could reliably make on a lathe. Machining of the thinnest parts is the most difficult task because the cutting tool occasionally tears the copper, resulting in discs that were discarded. Further outward, the thickness increases gradually until the maximum thickness is encountered and this thickness remains approximately constant as we move outward beyond the x-ray field of view (FOV) boundaries. We constructed ROIs having different DAP reducing characteristics because the apparent size of the attenuating region seen in the image varies depending on the FOV and SID used.

The gradual change in ROI thickness causes image intensity changes that vary gradually. As a result, a simple unsharp mask filter could be used to equalize the large area contrast increasing image intensity in the periphery without detailed knowledge of the ROI characteristics.¹³ The ROI aperture edge exists because we did not want to attenuate and harden the x-ray beam in the clinically important central part of the image. The edge causes a subtle image artifact that is not visible in all image sequences and *interventionists* were not concerned that it would affect guidance. The edge affects a small area of the image and is fainter than artifacts caused by other anatomical structures such as ribs and vertebral bodies.

3.B. X-ray imaging

All x-ray experiments were conducted using a General Electric Health Care (GEHC, Buc, France) Innova 2121-IQ biplane catheterization system. Although it is a biplane unit, studied procedures were conducted using only the frontal plane as is done in our institution. The clinical images shown to the operator undergo temporal and spatial processing by the system to reduce perceived noise and reduce large area contrast

FIG. 4. (Left) ROI attenuated image showing exposure control zones, which must not be attenuated to avoid interfering with catheterization system exposure feedback mechanism. The size exposure control region for pediatric protocol is a smaller fraction of the field of view than is the case for the adult protocol. (Right) A sampling of ROI attenuators in Lucite cassettes designed to allow rapid insertion and removal.

TABLE I. Summary of the physical characteristics of the ROIAs to be used in the proposed patient study. The transition region refers to the distance from the thinnest part of the copper to the point where the thickness is constant. A value of “No” for the minimum SID means that the ROIA cannot be used without interfering with the x-ray system exposure feedback mechanism when using an adult imaging protocol. A value of “Unrestricted” means that the ROI can be used over the full range of x-ray system SIDs. The ROIAs listed above can be used at all SIDs if the x-ray system is operated with the hybrid pediatric imaging protocol.

ROIA characteristics				X-ray system adult protocol minimum SID (cm)			
ROI name	Maximum thickness (mm)	Aperture diameter (mm)	Transition zone (mm)	20 cm FOV	17 cm FOV	15 cm FOV	12 cm FOV
1C	1.18–1.21	34.6	8.8	98.5	84.7	Unrestricted	Unrestricted
2D	1.09–1.12	31.1	12.18	104.7	90.03	Unrestricted	Unrestricted
3A	0.99–1.01	31.4	12.05	104.7	90.0	Unrestricted	Unrestricted
3C	1.19–1.23	30.2	9	111.6	96.0	Unrestricted	Unrestricted
4A	1.02–1.05	27.4	9.65	No	102.9	87.9	Unrestricted
4C	1.19–1.23	27.87	8.4	No	102.9	87.9	Unrestricted
5A	1.00–1.02	22.75	7.52	No	No	107.0	88.1

so that more gray levels are available to display structures of interest such as opacified arteries. This processing reduces peripheral darkening of the image due to ROIAs although it was not specifically designed for this purpose.

The x-ray catheterization system exposure parameters including tube current, filtration, kVp, and pulse width are determined automatically through a feedback control loop. One of the key measures used by this loop is the average intensity measured within a circular region in the central part of the image. If a ROIA was to attenuate a part of the x-ray beam within the said region, the system would react by changing the x-ray exposure parameters, possibly increasing dose in parts of the central region of the image and decreasing x-ray contrast by making the x-ray beam more penetrating. Figure 4 shows a clinical image onto which the circular regions used for exposure feedback are drawn. If an adult x-ray protocol is used, the exposure feedback is assessed in a circular region having a diameter equal to 1/2 of the width/height of the square FOV. If a pediatric x-ray protocol is used, the circular region is reduced to a diameter equaling 1/3 of the FOV allowing ROIAs with smaller apertures to be used so that greater DAP reductions can be achieved.

We elected to acquire many sequences in our study using a hybrid protocol derived from the pediatric protocol using the standard x-ray customization capabilities offered by the equipment. The pediatric protocol also differs from the adult version by favoring shorter temporal x-ray pulse width and a lowered dose target. The first difference is not effective because the radiation pulse increases automatically with patient thickness in the pediatric protocol so that it will equal the values associated with the adult protocol in adults. Using the customization capabilities of the equipment, we eliminated the said second difference by setting identical dose targets between these protocols. Native pediatric protocols were left unchanged for corresponding pediatric procedures performed independently of our trial at the institution.

The apparent size of the nonattenuating region in the x-ray image for a given attenuator is dependent on the FOV and the source image distance. A mechanism to compensate for changes in SID thereby maintaining a constant apparent ROIA size in the image would require tighter integration of the ROIA with the imaging system affecting its certification

for routine clinical use, and would be premature in a study to determine how much interventionists will use the device. Table I provides the minimum SID that can be used for a given FOV and ROIA when using the adult imaging protocol. There are no SID restrictions in effect when using any of the ROIAs with a hybrid protocol.

3.C. Data acquisition workstation

A DAP model described in Sec. 3.F below requires the determination of the dimensions of the irradiated FOV determined by the position of the x-ray collimator blades. This information unlike many other system parameters is not available via system logs and requires its estimation from the x-ray images. Catheterization systems do not by default store image frames from fluoroscopic sequences which constitute the majority of the acquired images. The operator has a limited time to request storage of a fluoroscopic sequence after which images are lost. Thus, it was determined that a mechanism was needed to save all of the images by default for the purpose of the study. We also wanted to automatically record DAP data from our ionization DAP chamber for all patient examinations. A multiprocessor server/workstation depicted in Fig. 5 was used to execute a program we developed to collect raw x-ray images (1024 × 1024 pixels 12 bits/pixel) directly from

FIG. 5. Diagram showing the prepatient ROI attenuator and M4 DAP-meter chamber. The DAP-meter is connected to the research workstation. Raw image data from the flat panel detector were also transferred to the research station via a fiber optic link and a custom data acquisition card.

FIG. 6. Diagram showing various DAP estimates. The M4 DAP values are obtained using an ionization chamber while the IDAP and SDAP are estimated using x-ray physics models.

the flat panel detector (FPD) and record DAP-meter readings continuously during catheterization.

Raw images were obtained using a custom acquisition card in the workstation connected to the FPD via a fiber optic cable. This solution developed by GEHC for research sites does not interfere with system operation. These images were not made visible to the interventionist during the procedure. We expect the contrast to noise ratio of the ROIA images in the periphery will be reduced. The raw images that are being collected will allow the investigation of image processing algorithms that are optimized to reduce darkening of the image caused by the ROIAs. These images can also be used to evaluate methods to reduce x-ray noise in peripheral regions of the images.

3.D. M4 DAP data collection

The research workstation is connected to a Diamentor M4 (PTW, Freiburg, Germany) DAP-meter via a serial link over USB and all functions of the M4 are controlled via this interface including the selection of the units of measurement and zeroing of the cumulative DAP value. Figure 2 shows that the ionization chamber is mounted above the ROIA such that it records DAP values after attenuation by the copper discs. The workstation sends a request for the cumulative DAP which is saved to a log for later analysis. The cumulative DAP sampling frequency was 2 Hz for the first ten patients in the clinical study but it was soon realized that a higher sampling frequency would be advantageous when syncing this information to logs from the catheterization system so it was subsequently increased to 5 Hz for the rest of the study. The DAP values were corrected for each imaging sequence as stipulated by PTW using kVp and copper filtration values obtained from the x-ray system logs (Sec. 3.E). Note that DAP measurements are insensitive to changes in the distance between the ionization chamber and the x-ray source provided that the chamber dimensions exceed the extent of the x-ray field of view. These DAP estimates are henceforth referred to as M4 DAP.

3.E. Innova system logs

The catheterization system generates logs showing the start and end times of all x-ray exposures as well as the x-ray techniques factors, gantry and table geometry, and many additional parameters related to system operation. One set of the said values are produced per fluoroscopic or record sequence and are those in effect at the end of the x-ray exposure (last images). These x-ray technique factors are used by our DAP model. The logs also provide DAP henceforth referred to as IDAP and dose

TABLE II. Ranges of x-ray technique parameter use in 70 exposures to test our DAP model. For each experimental condition, an exposure was obtained with no ROIA, ROIA 3A, and ROIA 4A. ROIA characteristics are described in Table I. Every DAP measurement is repeated twice ($\times 2$). For experiments 1–12, the table shows that there were two exposures for each of 0.1, 0.3, 0.5, 0.7, 0.9, and 1.0 mA s. Experiments 13–24 involved changes in copper filtration. Experiments 25–38 involved changes in kVp. Experiments 39–70 involved changes in field of size dimensions due to collimation. In that case, there were two exposures taken at 600, 1000, 1400, and 1800 mm (u) field sizes and the process was repeated eight times for a total of 32 experiments. The (v) field size was also varied with the first eight exposures done with a value of 600 mm, the next eight exposures were done with a value of 1000, etc.

Experiment index	Field size (u) mm	Field size (v) mm	SID (mm)	mA s	kVp	Copper filtration (mm)
1–12	1200	1200	900	0.1 $\times$ 2, 0.3 $\times$ 2, 0.5 $\times$ 2, 0.7 $\times$ 2, 0.9 $\times$ 2, 1.0 $\times$ 2	80	0
13–24	1200	1200	900	0.5	120	0.0 $\times$ 2, 0.1 $\times$ 2, 0.2 $\times$ 2, 0.2 $\times$ 2, 0.3 $\times$ 2, 0.6 $\times$ 2, 0.9 $\times$ 2
25–38	1200	1200	900	0.5	60 $\times$ 2, 70 $\times$ 2, 80 $\times$ 2, 90 $\times$ 2, 100 $\times$ 2, 110 $\times$ 2, 120 $\times$ 2	0
39–70	(600 $\times$ 2, 1000 $\times$ 2, 1400 $\times$ 2, 1800 $\times$ 2) $\times$ 4	600 $\times$ 8, 1000 $\times$ 8, 1400 $\times$ 8, 1800 $\times$ 8	900	0.5	80	0

FIG. 7. A plot of relative differences between DAP values from our model (SDAP) and values obtained with an ionization chamber (M4 DAP) for experimental conditions summarized in Table II and with no ROIA. The labels within the graph show which variable is varied for different ranges of experiment indices.

estimates calculated using a proprietary model that takes into account the x-ray technique parameters.

3.F. DAP model

We have developed a mathematical model based on work by Tucker *et al.*¹⁴ to predict the dose and DAP values for exposures with and without a ROIA. Values from our model are henceforth referred to as SDAP. The x-ray tube characteristics needed from this model including a description of all materials attenuating the x-ray beam, as well as the anode angle were obtained from GEHC.

FIG. 8. A plot of relative differences between DAP values from our model (SDAP) and values obtained with an ionization chamber (M4 DAP) for experimental conditions summarized in Table II and with ROIA 3A. The labels within the graph show which variable is varied for different ranges of experiment indices.

FIG. 9. A plot of relative differences between DAP values from our model (SDAP) and values obtained with an ionization chamber (M4 DAP) for experimental conditions summarized in Table II and with ROIA 4A. The labels within the graph show which variable is varied for different ranges of experiment indices.

In our model, the rectangular irradiated field with dimensions $(x, y) \in (-X/2, X/2) \times (-Y/2, Y/2)$ at the fixed ROIA position, d , 33.0 cm from the x-ray source, is divided into small rectangles and at each point the x-ray photon fluence spectrum $\Phi(x, y, E, kVp, d)$, is calculated using the Tucker *et al.*¹⁴ model where E is the photon energy and kVp is the x-ray tube kilovoltage. The beam is attenuated by various x-ray tube components each having a thickness t_n and attenuation $\mu(E)_n$. One of these components is made of copper having a thickness that depends on the exposure conditions. The ROIA is characterized by a spatially varying thickness $t_{ROIA}(x, y)$, and attenuation coefficient $\mu_{ROIA}(E)$. The air kerma is calculated as described in Johns¹⁵ resulting in

$$SDAP_{\text{primary}} = \int_{-Y/2}^{Y/2} \int_{-X/2}^{X/2} \int_0^{kVp} (\Phi(x, y, E, kVp, d) e^{-\sum \mu(E)_n t_n} \times e^{-\mu_{ROIA}(E) t_{ROIA}(x, y)}) (\mu(E)/\rho)_{\text{abs}}^{\text{air}} E dE dx dy, \quad (1)$$

where $(\mu(E)/\rho)_{\text{abs}}^{\text{air}}$ is the mass energy absorption coefficient of air. The thickness of each material is corrected for obliqueness of the x-ray at each location. The anode angle is similarly corrected for obliqueness. Due to the proximity of the ROIA to the DAP chamber, the contribution of scatter to DAP is included in our model and is written as

$$SDAP_{\text{scatter}} = \int_{-Y/2}^{Y/2} \int_{-X/2}^{X/2} \int_0^{kVp} (\Phi(x, y, E, kVp, d) e^{-\sum \mu(E)_n t_n} \times e^{-\mu_{ROIA}(E) t_{ROIA}(x, y)} (\mu_{ROIA}(E) - \mu_{ROIAab}(E)) \times t(x, y)) (\mu(E)/\rho)_{\text{abs}}^{\text{air}} E dE dx dy. \quad (2)$$

The scatter DAP is obtained by calculating the fraction of x-ray that is scattered per unit thickness as $(\mu_{ROIA}(E) - \mu_{ROIAab}(E))$, where $\mu_{ROIAab}(E)$ is the energy dependent x-ray absorption coefficient. The total DAP is given by SDAP

$= \text{DAP}_{\text{primary}} + \text{SDAP}_{\text{scatter}}$. Our DAP model, unlike the M4 DAP (measured with a chamber) and IDAP (GEHC proprietary model), can be calculated to include or exclude the presence of a ROIA for a given clinical x-ray image sequence. If irradiation is performed with no ROI, $t(x, y) = 0$ for all x, y values and thus the exponent with $t(x, y)$ disappears in Eqs. (1) and (2) equals zero.

The only practical disadvantage in SDAP estimates is the dependency of the x-ray acquisition parameters from the x-ray system logs, which are only available for the last frame of each imaging sequence. In clinical image sequences, these x-ray technique factors can vary as a function of time having higher or lower values than those recorded at the end. However, this is only relevant for the occasional sequences whereby one more of the gantry angles is changed or the table moved during acquisition, or the FOV or collimation is changed.

3.G. Data time alignment

The cumulative M4 DAP (2 or 5 samples/s) estimates each have a timestamp based on the research workstation clock. The radiation events described in the x-ray system logs have timestamps that are based on a different clock associated with the x-ray computer system. In order to use data from both systems, the time difference between the two clocks, which can be as large as 10 min must be determined. The cumulative M4 DAP data are analyzed to identify radiation events. An event begins if the difference between successive cumulative M4 DAP values in time, $\text{cM4 DAP}_{i+1} - \text{cM4 DAP}_i > 0.1 \text{ cGy cm}^2/\text{s}$ and lasts until this difference is maintained. The x-ray system logs record the use of radiation as events providing the times for the start and end of irradiation. These events are time shifted until the total overlap time for all events on both systems is maximized using a correlation technique. We also compute standard deviation of the residual time differences between the start of irradiation times for the M4 and the logs for the best time shift and this is repeated for the stop times. These statistics provide confirmation that the data have been time aligned correctly. The M4 radiation event and log events (start/stop) times are plotted as pulses as a function of time with height equal to the DAP rate as are the optimally time shifted IDAP pulses. These plots are computed for each patient allowing visual inspection of the correctness of the time shifting procedure.

The identification of radiation events is subject to occasional signal spikes of unknown origin leading to spurious

radiation events. Thus, once the time difference has been determined, we recalculate the M4 DAP values for each sequence using the corrected x-ray system start/stop radiation times. Thus, we identify the cumulative M4 DAP value immediately after the end of the irradiation according to the corrected x-ray system log time and subtract this value with the cumulative M4 DAP value obtained immediately before the start of irradiation again according to the corrected x-ray system time.

3.H. SDAP value *in vitro* validation

In our clinical study, we wish to determine amongst other parameters, the DAP reduction factor to the patient which we define as $R = \text{SDAP}_{\text{ROIA}}/\text{SDAP}_{\text{noROIA}}$ for a given x-ray acquisition (fluoroscopic or record) sequence. The model allows the calculation of the DAP reduction in a manner that is more granular than would have been possible by only comparing DAP cumulated over examination for patients randomized into a ROIA and no ROIA study arm. Notice that in this calculation, we use the ratio of two model derived DAP values making our estimate less sensitive to systematic errors in our model as well as variation in technique parameters during acquisition.

For the purpose of validating our model, we wish to compare it with ionization chamber measurements. Figure 6 illustrates the various DAP estimates available in a clinical use scenario and shows that we can compare M4 DAP values with SDAP values when no ROIA is present for validation. In the presence of a ROIA, we can compare SDAP post ROIA values to M4 values as well. In this paper, we focus on validation obtained under conditions that are more rigidly controlled where the same exposure can be repeated exactly with and without ROIA.

The SDAP model in presence of ROIA was verified *in vitro* on the x-ray system in service mode allowing all x-ray technique factors to be specified manually and fixed for the length of each x-ray exposure. A total 20 cm of Lucite was added to the x-ray beam and exposures lasted 3–4 s. DAP values from the Diamentor were normalized by the number of frames. SDAP values were calculated per image frame as well. Exposures were obtained for 70 distinct combinations of x-ray technique factors. For each technique factor combination summarized in Table II, three x-ray exposures were taken, one with ROI 3A, one with ROIA 4A, and one without ROIA. Thus, the Diamentor M4 DAP chamber yielded estimates of $\text{M4 DAP}_{\text{postROIA3A}}$, $\text{M4 DAP}_{\text{postROIA4A}}$, and $\text{M4 DAP}_{\text{noROIA}}$

TABLE III. Mean (systematic) differences, maximum differences, and standard deviation for SDAP values versus M4 DAP values obtained without a ROIA and with ROIA 3A and 4A. The values are calculated for the model without corrections and with corrections. The corrections were made to minimize the mean differences in DAP values.

Experimental conditions	No ROIA DAP			ROIA 3A DAP			ROIA 4A DAP		
	Mean diff (%)	Max–min (%)	Std dev (%)	Mean diff (%)	Max–min (%)	Std dev (%)	Mean diff (%)	Max–min (%)	Std dev (%)
No model correction	–8.1	13.3	2.9	–17.4	16.8	3.4	–16.8	15.5	3.1
Model corrected	–0.4	9.6	2.2	–3.5	10.3	2.5	–0.8	10.3	2.5

that can, respectively, be compared with the model estimates $SDAP_{postROIA3A}$, $SDAP_{postROIA4A}$, and $SDAP_{noROIA}$.

3.1. Irradiated field of view

Validation of the model is done with the x-ray unit in “service mode” whereby we can specify all of the imaging parameters including collimation. However, the model is designed to be used with clinical data in which case we must rely on system logs. These logs do not provide information regarding collimation and it must be, therefore, determined from the raw images using a method we developed. The method integrates the pixel values along a line on the left side of the image and again on the right side of the image. Both lines are, respectively, at the same distance from the left and right sides of the images, taking advantage of the symmetry in the position of the collimator blades. The lines move in unison toward the center of the image and the left and right integrated line values are added. When these combined integrated values reach a predetermined threshold, the distance between the left and right lines is an estimate of the width of the field of view. The process is also performed in the horizontal direction to provide the height of the irradiated field. The threshold value has been experimentally determined.

4. RESULTS AND DISCUSSION

4.A. SDAP value accuracy

Figures 7–9 show the percentage error for the SDAP estimates when compared to M4 DAP values obtained using an ionization chamber for exposure parameters specified in service mode. The values are plotted as a function of an experiment index, where each value corresponds to a unique combination of exposure conditions summarized in Table II. In order to achieve good agreement between the model and

Fig. 11. A graph showing DAP values from our model (SDAP) versus values obtained with an ionization chamber (M4 DAP) in the presence of ROI 3A described in Table I for experimental conditions summarized in Table II.

our ionization chamber measurements, we determined that corrective factors were necessary for some of the parameters of the model: the permanent copper filtration thickness value was increased by a factor of 1.2, while the component that is varied by the imaging system was increased by a factor of 1.06 and the tube current was increased by 1.16. This was done iteratively to reduce the variation in percentage difference error shown in Figs. 7–9. Table III shows how systematic and random errors changed as a result of these adjustments. The thickness of all ROIs was multiplied by a factor of 0.80 to reduce systematic errors in SDAP values obtained with ROIs.

The x-ray generation model by Tucker *et al.* is semiempirical and was initially tested by the authors on a small

Fig. 10. A graph showing DAP values from our model (SDAP) versus values obtained with an ionization chamber (M4 DAP) for experimental conditions summarized in Table II.

Fig. 12. A graph showing DAP values from our model (SDAP) versus values obtained with an ionization chamber (M4 DAP) in the presence of ROI 4A described in Table I for experimental conditions summarized in Table II.

FIG. 13. Clinical raw (directly from the flat panel detector) images obtained without a ROIA in (A) and with ROIA in (B)–(D) showing the irradiated field boundaries as dotted lines determined using a program we developed. The algorithm takes advantage of symmetry in the positioning of the collimator jaws and thus yields correct results even when one side of the image is strongly attenuated by the mediastinum, spine or the diaphragm, or any combinations thereof.

number of x-ray tubes.¹⁴ Figure 7 shows that systematic errors remain following the application of corrective factors. This is likely due to differences in x-ray tube design relative to the tubes that were used for model development and the prescription of empirical model factors which affect the x-ray photon spectrum. These differences may explain the deviations observed with variations in filtration in kVp. It may be possible to improve these results by adjusting empirical factor of the model. The deviations observed as a function of changes in field of view are comparatively small suggesting that our treatment of these variables is correct. There exists a newer model by Poludniowski *et al.*¹⁶ claiming improved accuracy but computer code for this model was not available for integration into our software.

Figures 10–12 show DAP values from the model (SDAP) corrected as described above plotted versus M4 DAP values for experimental conditions summarized in Table II. Figure 10 shows that the aforementioned x-ray tube current correction was also chosen to make the slope equal to one. In the case of Figs. 11 and 12 which were obtained with ROIA 3A and 4A, respectively, a multiplicative ROIA thickness correction was chosen to have a value of 0.804 such that the SDAP for ROIA 3A underestimates DAP by 2.2% while SDAP for ROIA 4A overestimates DAP by 2.2%. The differences in systematic errors between ROIA 3A and 4A maybe due to inconsistencies in copper thicknesses such that thicknesses along different radial lines are not the same. The thickness measurements shown in Fig. 3 were each obtained along a single radial line and the model assumes that ROIA possess circular symmetry. ROIA thickness measurement errors are biased toward overestimation due to difficulties in maintaining the entire ROIA surface flat against the milling machine table. This problem can generally be detected by applying pressure to the disc causing noticeable deflection where contact is poor and repositioning of the machinist parallels and clamping them. We estimate that the thickness systematic errors could be as large as 50 μm .

It is important to note that the correction applied to tube current and copper filtration via thickness to reduce systematic errors between SDAP and M4 DAP values can also be performed with clinical data. The principal difference is that these corrections for SDAP without a ROIA and S4DAP with a ROIA will be necessarily computed from different exposures having different known technique parameters unlike phantom experiments whereby the very same exposure could

be repeated three times. The number of clinical exposures available for correcting SDAP values will be much greater than was practical to use in phantom experiments. However, the x-ray parameters vary during clinical image acquisition and system logs containing information used in the DAP model only provide parameter values for the last acquired images. Thus, we expect a lower correlation between measured and model based DAP values during clinical use.

4.B. Irradiated beam boundary determination

The SDAP calculations depend on a determination of the irradiated beam dimensions. Testing of the program to determine the said dimensions with simple phantoms would not predict success when applied to clinical images containing highly attenuating structures like the mediastinum or ROIA. Therefore, testing was done with clinical data from the trial as the data collection phase had just been completed. Field boundaries from our program represented by dotted lines were superimposed onto x-ray images as shown in Fig. 13 allowing visual inspection of the results. We inspected the overlaid boundaries for every 100th image out of 708 724 images (64 patients, 4145 fluoroscopy or record sequences). We found no large errors in boundary determination due to a failure in identifying the boundary but smaller errors were observed. An image viewing program providing cursor coordinates was used on a smaller subset of images to quantify these discrepancies: The difference in pixel coordinates of the computer determined field boundary and the boundary determined visually was obtained for the 95th image from every sequence for all patients in our study. The maximum error in boundary determination was 28 pixels (5.6 mm) except for the last three patients for which it was 36 pixels (7.2 mm). The errors we measured were due to discrepancy in the left and right collimator jaw position due to mechanical play which changed over the course of the study. Robustness of our approach depends on left/right and up/down symmetry of the collimator jaws. These errors are small when compared to the 15–20 cm field of view used during cardiac catheterization.

5. CONCLUSIONS

DAP values vary greatly between examinations making a comparison of the said values between patients in which a

ROIA is used in some image sequences versus controls for which no ROIA is used difficult. A model predicting DAP value with and without ROIA can provide an estimate of DAP reduction using fewer patients. Our results demonstrate that on average we can predict DAP values with and without ROIA within 2.6% on average after corrections for systematic errors. The model is to be used to quantify reductions in DAP due to ROIA on the order for approximately 1.5–3.5. Therefore, we are confident that the model is sufficiently robust to provide good estimates of DAP reductions for our clinical trial for each x-ray acquisition sequence for which a ROIA was used.

The model allows DAP reduction predictions to be made for ROIA incorporating additional features not found in the actual prototype. In particular, it is trivial to vary the model ROIA to x-ray tube distance such that the attenuated region of the image is as large as possible without interfering with x-ray exposure feedback and determine what the DAP reduction would have been as a result of this enhancement in our clinical study.

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