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Understanding automated dose control in dynamic X-ray imaging systems

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Background

Modern interventional X-ray systems capture and display dynamic images in real time. X-ray output is controlled via automated closed loop control mechanisms. Interventional procedures may be lengthy, and radiation dose therefore must be kept to a minimum to avoid radiation induced injury [1]. The image quality must be adequate to complete the procedure whilst keeping the radiation dose As Low As Reasonably Achievable (ALARA). Automatic dose rate control (ADRC) systems are designed to accomplish the following four goals:

- 1) To automate the selection of radiographic factors required, making the system fast and easy for clinicians and radiographers to use. Manual control of radiographic parameters would be impractical in a busy clinic, and operator error could lead to inappropriate radiation dose rates being used.
- 2) To consistently provide adequate image quality. The ADRC should keep image noise at an acceptable level and achieve sufficient contrast. Image quality requirements will vary depending on the clinical task being undertaken.
- 3) To ensure that the patient dose adheres to local standards or regulations and diagnostic reference levels. The ALARA principle (keep dose As Low As Reasonably Achievable) must be followed.
- 4) To ensure that the hospital is provided with an acceptable X-ray tube lifetime.

Interest in understanding ADRC has been growing in the medical physics community in the last few years [2], and details of specific X-ray system's ADRC operation have been described [3][4], although such reports are uncommon in the literature.

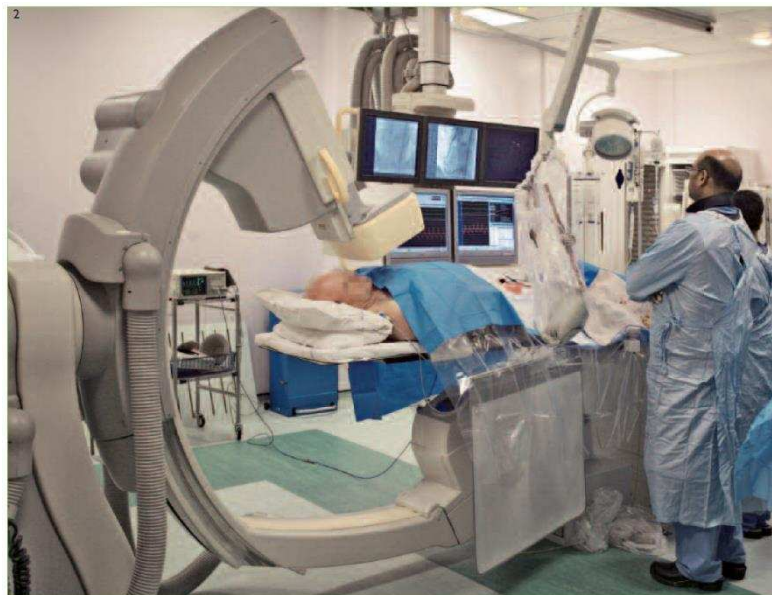


Figure 1: Cardiac catheterisation laboratory.

Learning Objectives

The aim of this poster is to explain the ADRC operation of interventional X-ray systems, specifically the Philips Allura X-ray system. With this knowledge, clinicians may more optimally utilise the X-ray system – to maximize image quality with the lowest possible dose not only to the patient but also to the staff.

Providing medical physicists with information on ADRC operation will allow the better understanding of routine quality assurance and acceptance testing investigations, which can be complicated by the closed loop operating control of ADRC.

Procedures Details

Types of X-ray imaging used in interventional imaging

In cardiology, the term “cine” refers to cardiac X-ray image acquisition by photographic film, as was the case on early image intensifier based cardiac X-ray imaging systems of the past. However, all solid state, flat panel detector based cardiac X-ray systems utilise a single imaging chain for both fluoroscopy and digital “cine” acquisition; a single digital image detector is responsible for image capture in both modes. The main difference between the modes is the higher image quality (due to the often up to 10 times higher X-ray dose) and storage of digital “cine” image acquisition runs, unlike with fluoroscopy. For modern cardiac systems, use of the term “acquisition” is therefore more suitable than the term “cine,” and it will be used throughout this poster.

Digital subtraction angiography (DSA) is a type of image acquisition which utilises digital subtraction to remove unwanted background anatomy from the images; this is used for neuro and vascular interventions. A “mask” image is captured prior to the vessels being injected with contrast medium. Once the contrast medium has entered the vessels of interest, the mask image is subtracted from the images containing contrast medium, leaving only the contrast medium to be viewed. The vessels are the only clear objects remaining in the image, as long as the patient has not moved.

Fluoroscopy is the most utilised imaging mode, allowing clinicians to view and manipulate interventional devices inside a patient. It utilises a lower patient dose rate than acquisition, and hence a lower level of image quality.

ADRC Design

On the left hand side of Table 1 are the parameters which can be controlled by ADRC. The range of these parameters is defined by the X-ray system limitations. In the centre of the table are variables which impact the control parameters, and are defined by limits of clinical system geometry; these variables also interact with each other. On the right hand side are the outcomes; they must remain within the limits of regulations from governing bodies such as the US Food and Drug Association (FDA) [5] and standards from organisations such as the International Electrotechnical Commission (IEC) [6].

<u>Control parameters</u>	<u>Interaction variables</u>	<u>Clinical/technical outcomes</u>
Peak tube voltage (kVp)	Patient thickness	Image quality

Beam filtration	Contrast detail Z	Patient dose
Tube current (mA)	Detail thickness	
X-ray pulse duration (ms)	Gantry angle	
Detector dose	Grid / air gap	
Frame rate	Scatter/primary	
	X-ray field size	Under limits of
		IEC standards
Under limits of	Under limits of	FDA regulations
X-ray tube / generator	System geometry	Local ALARA philosophy

Table 1: Parameters impacted by or impacting ADRC design

ADRC operation

The operation of one system's ADRC (Allura FD10, Philips Healthcare, Best, The Netherlands) will be described in detail. Systems of another manufacture or model will vary in operation, but there will be many similarities.

When imaging is initiated by the user, the equivalent water thickness of the patient (hereafter "patient thickness") is estimated. This is initially set by the system's programmed application selected by the user. Alternatively, if there was a prior fluoroscopy run, as is often the case, the effective patient thickness is calculated from data derived from the fluoroscopy sequence instead. This calculation takes place within the system and is based on the radiographic factors, beam filtration, source to image distance (SID) and pixel intensity values from the previous fluoroscopy run. Once estimated, the patient thickness is the starting point used to ascertain the initial radiographic factors to use (see Figure 2). There may also be copper spectral filtration in the X-ray beam, depending on the application selected by the user. The peak tube voltage is selected according to the patient thickness and according to the thickness (if any) of copper spectral filtration; the tube current and pulse duration are selected according to the peak tube voltage selected (see Figure 2).

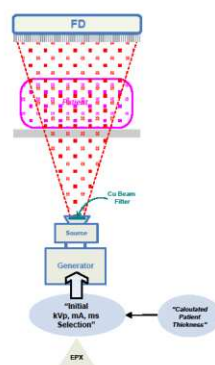


Figure 2: Initial radiographic factors are selected

Once the acquisition or fluoroscopy run has begun using these initial radiographic factors, detector output is measured after each image frame, using digital image pixel values. Each user-selectable application has a requested detector output which determines the average pixel intensity

of the image. A dose rate control mechanism calculates the difference between the measured and requested detector output, after each image frame. The peak tube voltage, tube current, and pulse duration are then updated in the next X-ray pulse. This process is repeated after each frame throughout the run in order to achieve the requested detector output (see Figure 3).

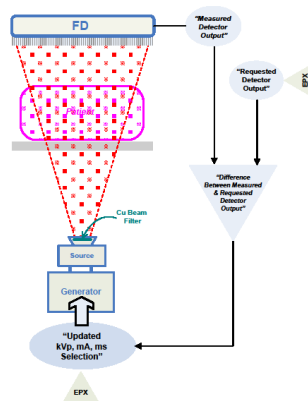


Figure 3: Updating radiographic factors

After each fluoroscopy run, the patient thickness is updated via calculation within the system. This calculation is performed at the end of each fluoroscopy run to account for changes in effective patient thickness, and begins the described process again. For example, if the image projection (gantry angle) changes, the effective patient thickness changes, and radiographic factors must be adjusted accordingly. The distance between the X-ray source and patient is not taken into account for the patient thickness calculation. Therefore, patient table height does not have an impact on operation of the ADRC. The patient thickness calculation is used to adjust radiographic factors when a fluoroscopy run is captured after a change in effective patient thickness. If only the SID or image field of view (FOV) is changed, radiographic factors are updated without the need for a fluoroscopy run.

In DSA the ADRC operation differs slightly. An image run begins with a “test shot” X-ray pulse, using the initial radiographic factors. Detector output from the test shot image frame is measured, and the ADRC mechanism mentioned above calculates the difference between the measured and requested detector output. The radiographic factors are updated in the next X-ray pulse – where the subtraction mask begins. This is not repeated after each image frame as it is for acquisition and fluoroscopy. In DSA, these factors are now fixed for the entire run; i.e. in Figure 3 the radiographic factors are only updated once per DSA run.

Limitations

The characteristics and limitations of the imaging system's X-ray tube and generator play an important role in designing ADRC. Specific constraints limit which radiographic factors may be used in combination. The X-ray tube and generator will have a range of tube voltage and current values that are available (e.g. 40 – 125 kVp and 50 – 1000 mA respectively). However certain combinations

of factors cannot be used together, for instance high kVp-high mA combinations, and low kVp-high mA combinations. Figure 4 shows an example of available kVp-mA values, with grey shading indicating combinations that are not permissible. More details on the reasons for the limitations are provided in the following sections.

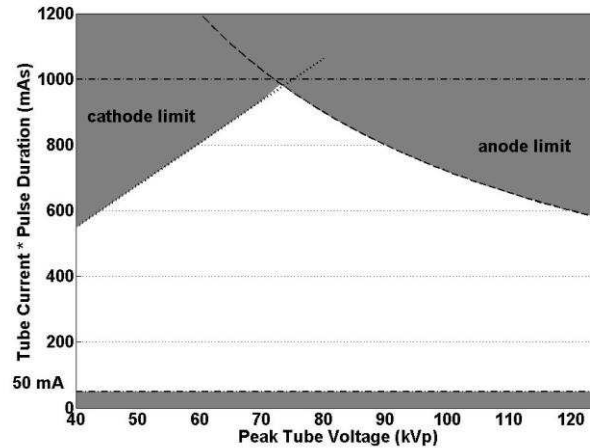


Figure 4: Typical anode, cathode, and tube current limits form ADRC “canvas” for cardiac image acquisition (Philips Allura interventional X-ray system)

Technical Limits

The X-ray tube’s anode introduces two main limits related to heating of the focal track and the heat capacity and rate of cooling of the anode assembly. Focal track heating is mainly an issue in higher X-ray output operating modes, i.e. cardiac acquisition and DSA. This is a thermal limit set to avoid thermal damage of the focal track. The actual numerical values of this limit depend on the X-ray tube type and design (which includes anode angle used), focal spot selection, length of image run, and duty factor (product of X-ray pulse duration and image frame rate). The duty factor is the ratio of the X-ray pulse duration to the time duration between X-ray pulses; that is, the fraction of the pulse cycle where irradiation occurs. For example, the duty factor may double when switching from cardiac image acquisition (15 f/s * 7 ms = 0.1) to neuro DSA mode (2 f/s & 125 ms = 0.2). This doubling in duty factor would necessitate a lower anode limit. In addition to the focal track, the anode assembly must not be overheated. Heat capacity of the anode, and its rate of cooling, affect this limit. Heating within the X-ray tube assembly and X-ray generator must also be controlled. This is mostly related to lengthy fluoroscopy procedures, and is controlled by imposing a fixed power load (“isowatt”), in order to allow adequate cooling of the X-ray tube housing and generator. The curve looks similar to the anode limit shown on the canvas; the values of this limit depend on the X-ray tube and generator used.

Operation of the X-ray tube filament also has a limitation, and this is represented on the grey “canvas” by the cathode limit (Figure 4). Electron emission from the cathode is restricted at low X-ray tube voltage. Actual values of the cathode limit depend on the X-ray tube type and design, and focal spot selected. A clinician may wish to use the smaller focal spot to increase spatial resolution, but this will lower on the anode and cathode limits, thus altering which radiographic factors may be selected. The canvas shown here would be much smaller.

X-ray systems have a minimum and maximum tube current; actual limit values depend on the X-ray tube and generator used. In fluoroscopy, some systems use a “grid switch” to remove ramp and tail edges of X-ray pulses [7] to reduce motion blur and low energy radiation which does not contribute to image quality. The emitter structure needs to be extremely stable under grid switch conditions; in order to ensure X-ray focus stability the maximum X-ray tube current is limited.

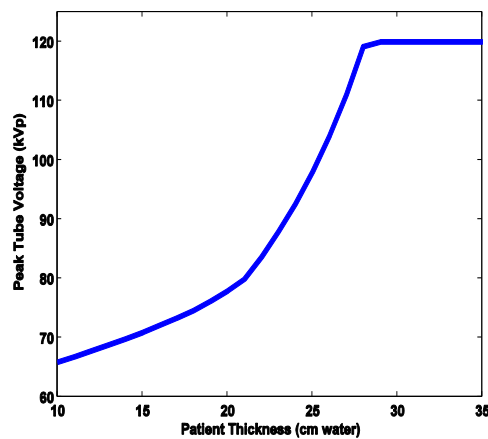
Legal Limits

The FDA has set a maximum limit on fluoroscopy patient exposure rates - 10 R/min (8.8 cGy/min) - for all fluoroscopy systems. This patient entrance dose rate limit, if plotted on the canvas shown in Figure 4, would appear as a curve similar to the anode limit shown. Between legal (patient dose rate) limits and technical (thermal) limits on such a canvas, the innermost curve acts as the canvas frame. IEC standards state that a manufacturer have two of its differing default fluoroscopy modes within one application named "normal" and "low" (patient dose rate) modes. These modes must be designed such that when a user switches from low dose to normal mode, the patient dose rate is at most doubled, for reference air kerma measurements [6]. By patient dose, both the FDA and IEC are referring to the entrance air kerma (without backscatter).

The FDA and IEC each define its own measurement point for patient dose rate. The IEC point is defined at 15 cm in front of the isocentre of the the C-arm as the "patient entrance reference point" for air kerma. The FDA measurement point is defined at 30 cm in front of the detector entrance surface, and therefore dependent on the SID. When the detector is moved, the patient dose rate measurement point changes, and the radiographic factors required to maintain the same limiting patient dose rate also change, due to the inverse square law.

ADRC Curves

Figure 5 shows example dose control curves – those used for the vascular, medium–dose, 15 frames per second fluoroscopy mode – with radiographic factors shown as a function of patient thickness.



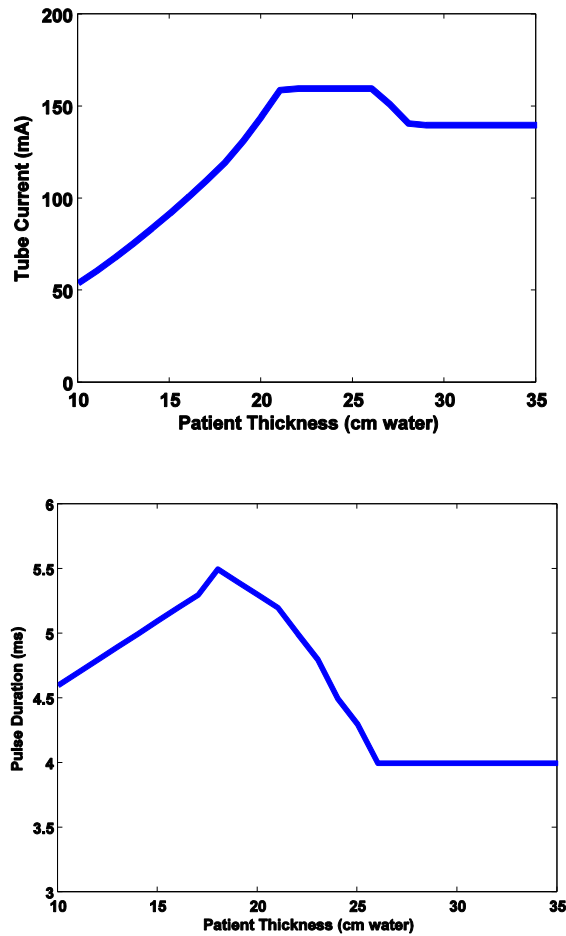


Figure 5:ADRC curves for vascular 15 frames per second medium dose fluoroscopy mode

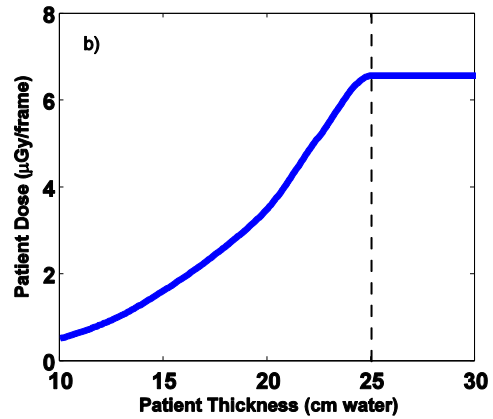
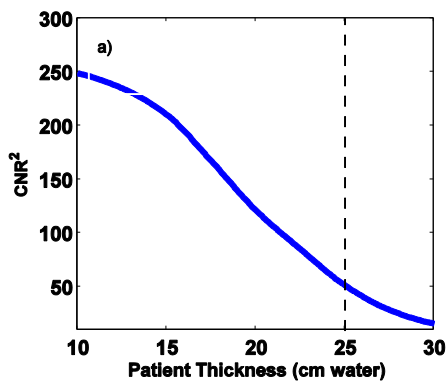
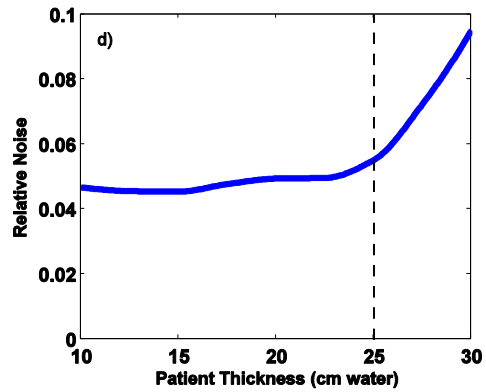
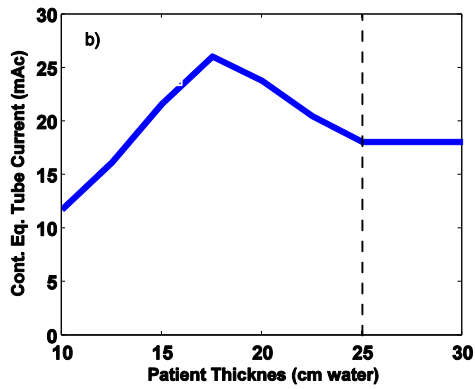
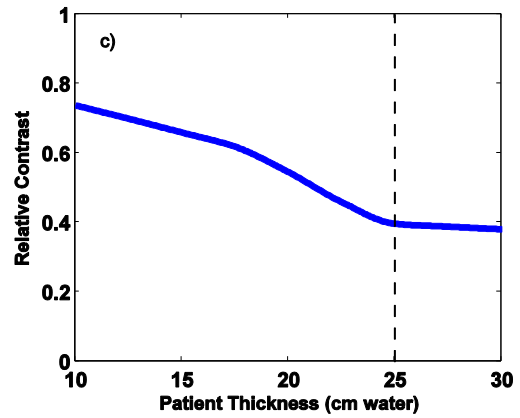
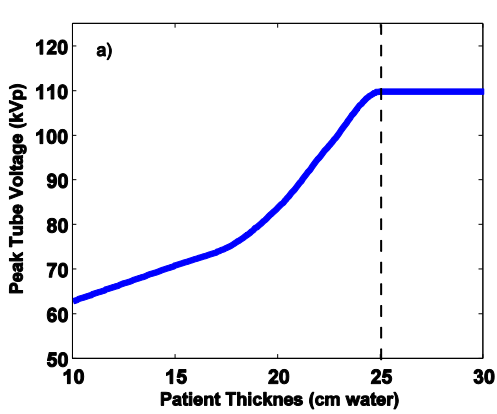
Effect on Image Quality & Dose

The ADRC dictates the image quality and patient dose delivered. This is illustrated using measurements from a dated cardiac X-ray system (H5000, Philips Healthcare, The Netherlands) and image quality predictions produced by a software based X-ray system model to calculate contrast, noise and patient dose (entrance air kerma). Iodine (0.2 mm thickness) was modelled to simulate contrast medium.

This particular ADRC operates by attempting to maintain a constant detector dose rate. In this case, the peak tube voltage is increased in response to increasing X-ray attenuation of the patient. Therefore, peak tube voltage is determined by the patient size. This X-ray system reaches its peak tube voltage limit at 25 cm patient thickness, as demonstrated by the dotted lines in Figure 6. Beam energy increases with rising peak tube voltage and with X-ray beam attenuation (i.e. hardening) from increasing patient thickness. Contrast is reduced by the increasing beam energy and then levels off when the peak tube voltage remains fixed. At this point the contrast still decreases very slightly due to beam attenuation.

Like the peak tube voltage, the tube current remains at a fixed value for patient thicknesses of 25 cm and greater. At 25 cm patient thickness, the system limits are reached; here the relative image noise (normalised to the background) begins to increase rapidly. There is a rise in noise when

the number of X-ray photons reaching the detector drops. This is due to the Poisson distribution of X-ray photons; noise is equal to the square root of the number of X-ray photons per pixel.



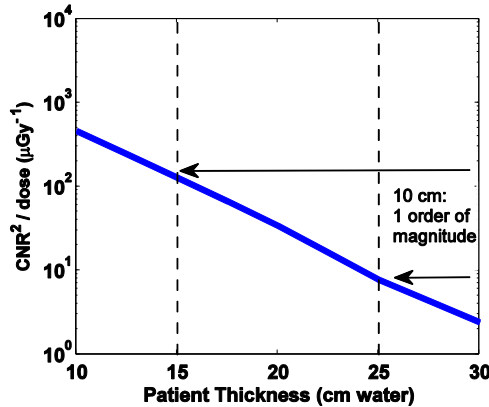


Figure 6: Effect of radiographic factors on image quality and patient dose

In examining the image contrast and noise together, overall image quality measurement CNR^2 for this particular X-ray system decreases with increasing patient thickness. Input air kerma to the patient would increase as patient thickness increases, until the system limits are reached. Scatter would cause a slight increase in patient dose, which is not shown here. Image quality to dose optimisation metric CNR^2 /patient dose would decrease as patient thickness increases, indicating that it would be more difficult to optimise the balance of image quality and patient dose for thicker (larger) patients. It is clear that with 10 cm of added patient thickness, there is an order of magnitude of difference in the optimisation metric.

Significance of Patient Thickness

As a first approximation increasing patient thickness causes a reduction in CNR^2 /patient dose of 1.25^t , where t is the increase in patient thickness in cm. This means that for a 3 cm increase in patient thickness, in order to maintain the same level of image quality the patient dose needs to increase by a factor of two. For a 6 cm increase, a four times increase in patient dose is required to maintain the same level of image quality.

Within one patient procedure, these changes in patient thickness may be observed simply from changing imaging projections i.e. gantry angle. For example, in cardiac imaging a posterior-anterior (PA) view of the heart typically shows internal system-calculated patient thicknesses of 22-24 cm, whereas for the "spider view" thicknesses can be as large as 40 cm. The spider view is right anterior oblique (RAO) 30-40 degrees, caudal 30-40 degrees whereas PA angles are near 0 and 0 degrees. For a change in angle of 45 degrees, an increase of approximately four times in patient dose may be observed; this is due to the increase in patient thickness of approximately 6 cm water equivalence. Similarly, for neuro X-ray imaging there may be differences of approximately 6 cm difference between lateral and frontal anatomic projections. Figure 7 shows predicted projection thicknesses and corresponding entrance dose rates when imaging an elliptical phantom of dimensions 30 x 18 cm from various angles.

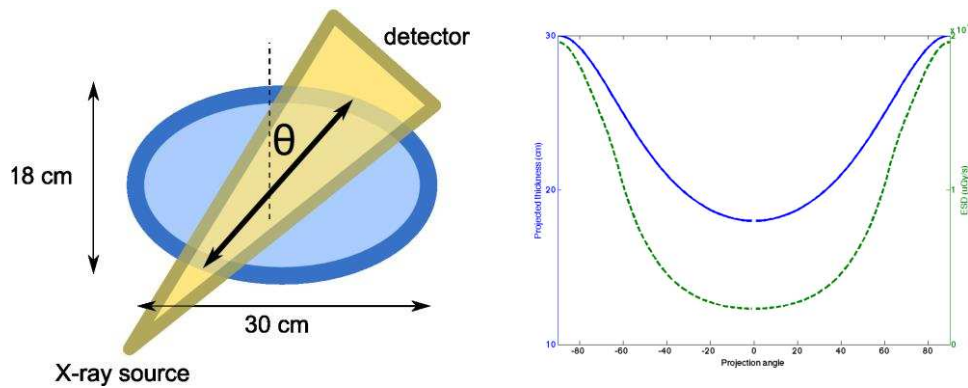


Figure 7: Projected thicknesses and entrance surface dose rate for a simulated elliptical phantom as a function of projection angle.

Conclusions

An understanding of ADRC operation is necessary for physicists performing measurements on interventional X-ray systems. During measurements when any change is made which might cause the ADRC to alter its radiographic factors (such as a change in phantom thickness), a fluoroscopy run should be performed after the change and should be ignored in terms of measurements. A sufficient number of acquisition or fluoroscopy images at the beginning of a sequence should not be included in image quality measurements to allow the ADRC to settle on a working set of radiographic factors.

Clinicians and radiographers should strongly consider the significance of projection (gantry) angles when performing interventional procedures. When using a steep projection angle it is considered by the ADRC as being a very thick patient, and this is reflected in the relatively high dose X-ray settings used to capture images of adequate quality.

It is beneficial for interventional X-ray system users – clinicians and medical physicists – to understand ADRC mechanisms which control these systems. In describing the Philips Allura interventional X-ray system ADRC, it is hoped that this poster will help educate users as well as instigate further discussions and broaden the knowledge of this topic in the field of medical X-ray imaging.

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