Effective Dose Estimation in Fast-kV Switch Dual Energy Computed Tomography

by

Justin Mark Raudabaugh

Medical Physics Graduate Program
Duke University

Date:_______________________

Approved:

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Terry T. Yoshizumi, Supervisor

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Rathnayaka M. Gunasingha

___________________________
James G. Colsher

Thesis submitted in partial fulfillment of the requirements for the degree of Master of Science in the Graduate Program of Medical Physics in the Graduate School of Duke University

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ABSTRACT

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Abstract

*Project 1: Effective Dose Estimation in Fast-kV Switch Dual Energy CT*

**Purpose**

The objective of our study was to test a new approach to approximating organ dose by using the effective energy of the combined 80kV/140kV beam used in fast kV switch dual-energy (DE) computed tomography (CT). The two primary focuses of the study were to first validate experimentally the dose equivalency between MOSFET and ion chamber (as a gold standard) in a fast kV switch DE environment, and secondly to estimate effective dose (ED) of DECT scans using MOSFET detectors and an anthropomorphic phantom.

**Materials and Methods**

A GE Discovery 750 CT scanner was employed using a fast-kV switch abdomen/pelvis protocol alternating between 80 kV and 140 kV. The specific aims of our study were to (1) Characterize the effective energy of the dual energy environment; (2) Estimate the f-factor for soft tissue; (3) Calibrate the MOSFET detectors using a beam with effective energy equal to the combined DE environment; (4) Validate our calibration by using MOSFET detectors and ion chamber to measure dose at the center of a CTDI body phantom; (5) Measure ED for an abdomen/pelvis scan using an anthropomorphic phantom and applying ICRP 103 tissue weighting factors; and (6)
Estimate ED using AAPM Dose Length Product (DLP) method. The effective energy of the combined beam was calculated by measuring dose with an ion chamber under varying thicknesses of aluminum to determine half-value layer (HVL).

**Results**

The effective energy of the combined dual-energy beams was found to be 42.8 kV. After calibration, tissue dose in the center of the CTDI body phantom was measured at 1.71 ± 0.01 cGy using an ion chamber, and 1.73±0.04 and 1.69±0.09 using two separate MOSFET detectors. This result showed a -0.93% and 1.40 % difference, respectively, between ion chamber and MOSFET. ED from the dual-energy scan was calculated as 17.42 ± 0.05 mSv and 12.62 ± 0.04 mSv by the MOSFET method for female and male respectively, and 14.62 mSv by the DLP method.

**Project 2: Evaluation of Phillips DoseAware Personal Dose Meter System**

**Purpose**

The Phillips DoseAware system is a novel approach to providing radiation workers with real-time information on individual dose levels through the use of personal dose meters wirelessly connected to a wall display. The intended benefit of this system is the encouragement of safer work practices by allowing radiation workers to see the dose they are receiving at any point in a medical procedure. The goal of this
project was to access the accuracy of the real-time dose information being relayed to the wall display. Our approach to this was two-fold: (1) to test the manufacturer listed performance specifications of the system, and (2) to compare the performance of the system with other commercially accepted detectors in radiology environment.

**Materials and Methods**

The initial performance testing was done using a Precision X-Ray X-RAD 320 biological irradiator to irradiate the Phillips Personal Dose Meter (PDM) and a 6cc ion chamber as gold standard. The performance specifications tested were dose rate dependence, energy dependence, and angular dependence. In order to test dose rate and energy dependence, five sets of data were obtained from the detectors corresponding to different X-ray tube potentials. Angular dependence was investigated using a constant set of beam parameters, and foam wedges cut at various angles to provide the displacement.

The second phase of testing was conducted in an interventional radiology room outfitted with a Phillips Allura XPERA FD 20/20 fluoroscopy system. An adult anthropomorphic phantom was placed on the patient table to provide a realistic scattering medium. Three locations in the room were selected to represent positions where radiology personnel would likely be stationed during a typical procedure. At each of these locations a PDM and a RadEye G detector were positioned facing towards
the X-ray source. The exposure reading from each detector was recorded for three low-dose fluoroscopy scans, and three cine-angiography scans. The detectors were then replaced with ion chambers and the scans were repeated.

Results

The initial performance testing verified that the Phillips DoseAware system functioned well within the manufacturer listed specifications. The interventional radiology experiment showed adequate agreement between the PDM and the ion chamber as gold standard.

Project 3: Estimating Effective Dose from Novel Digital Radiography Renal Stone Protocols

Purpose

The goal of this project was to estimate effective dose from novel digital radiography protocols proposed by Duke Urology to image renal stones. These four protocols have been developed to serve as an alternative to Computed Tomography in hopes to lower dose to patient during diagnosis.

Materials and Methods

The protocols were performed using a GE Discovery XR656. Skin entrance dose to an adult anthropomorphic phantom was recorded using a 0.18 cc ion chamber for three scans of each protocol. Each protocol was modeled as a series of individual X-ray
projections at integer angles across the total tube sweep angle for the procedure. The
organ doses for these projections were then simulated using PCXMC software and then
totaled for each protocol. ICRP 103 tissue weighting factors were used to estimate
effective dose from this data.

Results

The effective dose for the protocols was estimated to be 0.98 ± 0.002 mSv, 2.18 ±
0.006 mSv, 2.02 ± 0.004 mSv, and 2.02 ± 0.005 mSv respectively.
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1. Effective Dose Measurements in Fast-kV Switching Dual Energy Computed Tomography

1.1 Introduction

Dual Energy Computed Tomography (DECT) is a method of Computed Tomography (CT) imaging in which two CT scans are performed on the patient using different X-ray tube potentials which allow for energy-selective reconstruction. Reconstruction in CT relies on measuring the line integral of the linear attenuation coefficient along each projection. The linear attenuation coefficient depends primarily on atomic number and electron density and the energy of the X-ray photons. This allows the attenuation coefficient of a material to be modeled as a function of energy using a set of basis functions. These basis functions provide information on how attenuation for a material will change between two energy spectra. In DECT a second measurement of attenuation at a different energy provides an additional data set. This additional data set provides a measurement of the change in attenuation between the two energy projections. This information can then be compared to known changes of attenuation provided by the basis functions for various materials in order to differentiate material composition. [1][2]

Acquiring datasets at two separate energies allows for a number of advantages over conventional Computed Tomography. The two data sets provide attenuation information corresponding to each energy that can then be used for enhanced material
discrimination, improved image registration, and a reduction of beam hardening effects.

[3] [4]

The utilization of separate energy scans introduces a level of difficulty when attempting to physically measure organ dose due to the energy dependence of internal dosimeters. There are currently very few publications that attempt to estimate effective dose in a dual energy environment using physical measurements for organ dose, and these studies specifically pertain to dual source computed tomography [5][6][7]. The goal of this paper is to propose a new method of measuring organ dose that circumnavigates this difficulty in dual-energy environments.

1.1.1 Methods of Dual Energy Computed Tomography

DECT has been implemented using several designs including rotate-rotate software, photon-counting detectors, Dual Source CT, and fast-kV switching. [8] Rotate-rotate software creates the DE data by having the gantry complete one rotation acquiring data at the first kV setting, then repeating the rotation at a second tube potential. The advantage of this method is that the software can be used when the CT scanner does not have the ability to simultaneously acquire data at two energies. The repeated rotations result in slower scan time and make the system susceptible to motion artifacts due to the delay in separate data acquisitions.

Another method of DECT makes use of rapid photon-counting detectors. These detectors re-bin the x-ray spectrum into low and high kV data sets in order to allow for
material discrimination. K-edge X-ray filters are used to selectively remove photons with energy between the high and low kV bins. [9] This system allows for material discrimination using a single image acquisition at one tube potential.

There are currently two primary methods used to create a dual-energy environment available as commercial products. The first of these methods is referred to as Dual Source Computed Tomography (DSCT), in which two separate X-ray tubes are operated at different energies simultaneously. The two sources are operated at differing tube currents and each have a different level of filtration. The second method of DECT makes use of a single X-ray tube that rapidly switches between low kV and high kV between consecutive projections, alternating at a rate of up to 4.8 kHz. Due to the fast rate of kV switching the single X-ray tube is operated at a constant current, with longer dwell time for the low kV projections to compensate for lower attenuation. For this study we used a fast-kV switching system to create our dual-energy environment, with tube potential alternating between 80 kV and 140 kV. A visualization of this system can be seen in Figure 1. The key advantage of fast-kV switch DECT compared to DSCT is a reduction in motion artifacts due to the rapid switching between energy projections; however it offers limited control of tube current for the different projections.
Figure 1: Illustration of fast-kV switching DECT. The left image illustrates the alternating of energy between consecutive projections. The right image illustrates the rise and fall of the tube potential. [10]

1.1.2 Challenges

Physical measurements of individual organ dose due to an imaging procedure are generally obtained using small internal dosimeters such as thermoluminescent dosimeters (TLD) and metal oxide semiconductor field-effect transistor (MOSFET) detectors. These dosimeters have an energy dependence which gives rise to difficulty when calibrating the detectors in a dual-energy environment. The difference in MOSFET calibration factors between 80 kV and 140 kV can be seen in Table 1. In order to correctly calibrate our detectors it is necessary to characterize the dual-energy environment.

<table>
<thead>
<tr>
<th></th>
<th>80 kV</th>
<th>140 kV</th>
</tr>
</thead>
<tbody>
<tr>
<td>Calibration Factor (mV/cGy)</td>
<td>33.8</td>
<td>27.8</td>
</tr>
</tbody>
</table>
1.1.3 Goals

The objective of this study was to test a new method of estimating organ dose by using the effective energy of the combined dual-energy environment to calibrate our internal dosimeters for dose measurements. This process can be viewed as the combination of two primary goals: (1) Validating dose equivalence in a dual energy environment between our MOSFET detectors and an ion chamber serving as a gold standard; (2) Measuring organ dose using MOSFET detectors in an anthropomorphic phantom, and estimating effective dose in our dual-energy environment by applying ICRP 103 tissue weighting factors. [11]

1.2 Experimental Methods

This study was conducted using a GE Discovery CT 750HD implementing Gemstone Spectral Imaging (GSI) abdomen/pelvis protocol. This GSI protocol is a fast-kV switching dual-energy scan, rapidly alternating between 80 kV and 140 kV. Our experimental method includes six primary components: (1) Characterize the effective energy of the dual-energy environment; (2) Estimating corresponding f-factor for soft tissue; (3) Calibrating our MOSFET detectors using a single energy beam with effective energy equal to that of the dual-energy environment; (4) Validating our calibration by comparing readings from MOSFET and an ion chamber using a CTDI body phantom in

| % difference | 19.5 |
the dual energy environment; (5) Measuring ED for an abdomen/pelvis DECT scan using an anthropomorphic phantom and applying ICRP 103 tissue weighting factors; and (6) Estimating ED using AAPM Dose Length Product (DLP) method [12] for comparison.

1.2.1 Determining Effective Energy in Aluminum for Combined Dual-Energy Spectra

The effective energy of an X-ray beam is defined as the monoenergetic photon energy that would produce the same first half-value layer as the beam itself in a given material. By directly measuring the HVL, in mm AL, of the combined dual energy beams we can calculate a linear attenuation coefficient using the simple relation:

$$\mu = \frac{\ln 2}{HVL} \quad (1)$$

Once this value is known we can use attenuation data such as NIST X-ray attenuation databases [13] to find our effective energy.

The GSI protocol was performed in diagnostic mode which allows the X-ray tube to remain stationary above the patient couch. The combined dual-energy beam was then measured using a 6cc ion chamber (10x5-6, Radcal, Monrovia, CA). This measurement was then repeated four times adding different thickness of thin aluminum sheets between the X-ray tube and ion chamber. The ion chamber measurements were plotted vs. thickness of added aluminum filtration in order to perform exponential regression
analysis, which was used to calculate a HVL of 5.02 mm. Using the method described above this value was found to correspond to an effective energy of 42.8 kV [13].

![Figure 2: Ion chamber reading vs. aluminum thickness for combined dual-energy beams](image)

1.2.2 Estimating f-factor for soft tissue

It is necessary to specify in our calibration that we seek to find dose in soft tissue. This is accomplished by using an f-factor which correlates the dose we measure in air to that of a medium of interest, in this case soft tissue. The equation for f-factor can be seen below:

$$ f - \text{factor} = 0.873 \left( \frac{\left( \frac{\rho_{\text{en}}}{\rho} \right)_{\text{ST}}}{\left( \frac{\rho_{\text{en}}}{\rho} \right)_{\text{AIR}}} \right) $$  \hspace{1cm} (2)
Where \( \left( \frac{\mu_{en}}{\rho} \right)_{ST} \) represents the ratio of mass energy absorption coefficients for soft tissue and air. The f-factor corresponding to our effective energy was found to be 0.923. This value was later used to formulate calibration factors for each individual MOSFET detector.

It is also worth noting that the f-factor changes very little within the effective energy range of the separate low and high kV beam, which is roughly 40-55 kV. This relative stability in f-factor can be seen in Figure: 3

![F-Factor vs Energy](image)

**Figure 3: f-factor vs. energy in the range of our projected effective energy**

### 1.2.3 Calibration of MOSFET Detectors

In order to calibrate our detectors we sought to use an 80 kV beam with an HVL, and therefore effective energy, equal to that we previously measured for our dual-
energy environment. This was performed by using a Piranha multifunction X-ray meter (RTI Electronics, Mölndal, Sweden) to measure HVL, and adjusting the tube current of the 80 kV beam in order to create a single energy environment with the same HVL we previously measured for the combined dual-energy beams.

![Piranha positioned in the center of the calibration beam to measure HVL](image)

**Figure 4: Piranha positioned in the center of the calibration beam to measure HVL**

Due to limitations of the GE Discovery CT750 HD input parameters we were unable to reduce the HVL of the beam to our desired 5.02 mm Al. The closest we were able to approach to this provided a HVL of 5.42 mm Al, corresponding to an effective energy of 44.7 keV. The calibration beam parameters can be seen in Table 2.
Table 2: Calibration beam parameters

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>kVp</td>
<td>80</td>
</tr>
<tr>
<td>mA</td>
<td>40</td>
</tr>
<tr>
<td>HVL measured by Piranha</td>
<td>5.42</td>
</tr>
</tbody>
</table>

The MOSFET detectors were placed adjacent to the active area of a 6cc ion chamber (10x5-6, Radcal, Monrovia, CA). The detectors and the ion chamber were placed on a Styrofoam slab and suspended off the edge of the patient table in order to produce a calibration in free air that would minimize the effects of backscatter.

Figure 5: MOSFET calibration setup
The MOSFETs and ion chamber were then irradiated using the 80 kV calibration beam in order to calculate calibration factors for the 18 individual MOSFET detectors. These calibration factors convert the raw millivolt readings of the MOSFET detectors into absorbed dose in soft tissue (cGy). The equation for these calibration factors can be seen in equation 3.

$$\Delta mV = \frac{\Delta mV}{(\text{ion chamber reading}) \cdot (\text{ion chamber correction factor}) \cdot (f-factor)} \quad (3)$$

Where $\Delta mV$ represents the change in voltage read by each of the MOSFET detectors.

### 1.2.4 Experimental Validation of MOSFET Calibration

The calibration method was tested by inserting two MOSFET detectors along with an 0.18cc ion chamber (10x5-0.18, Radcal, Monrovia, CA) in the center cavity of a CTDI body phantom (West Physics, Atlanta, GA).
The 0.18cc ion chamber was chosen based on two criteria. The first was its fairly small energy dependence over the energy range produced in our dual-energy protocol. The second benefit of the 0.18cc ion chamber is that its small size allows for the detector to be easily placed into our CTDI phantom. In Figure 5 we can see the scout image from the GSI abdomen/pelvis protocol.
Figure 7: Localizing scout showing the location of MOSFETs and ion chamber in the CTDI body phantom

The CTDI body phantom was then scanned three times with the dual-energy GSI abdomen/pelvis protocol in order to validate the MOSFET calibration. The results can be seen below in Table 3.
Table 3: Calibration validation results

<table>
<thead>
<tr>
<th>Dose (cGy)</th>
<th>Ion Chamber</th>
<th>MOSFET 1</th>
<th>MOSFET 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Run 1</td>
<td>1.72</td>
<td>1.73</td>
<td>1.64</td>
</tr>
<tr>
<td>Run 2</td>
<td>1.71</td>
<td>1.69</td>
<td>1.64</td>
</tr>
<tr>
<td>Run 3</td>
<td>1.72</td>
<td>1.77</td>
<td>1.79</td>
</tr>
<tr>
<td>Avg.</td>
<td>1.71 ± 0.01</td>
<td>1.71 ± 0.06</td>
<td></td>
</tr>
</tbody>
</table>

These results experimentally confirm dose equivalence between our calibrated MOSFET detectors and ion chamber (as gold standard).

1.2.5 Measuring Organ Dose.

After verifying our calibration the MOSFETs were inserted into an adult anthropomorphic phantom (Model 702-BR190, CIRS). This phantom is composed of 39 slabs of thickness 2.5 cm, and simulates the radiological behavior of different tissue types including soft tissue, lung, bone, brain, and spinal cord and disks.
Figure 8: Anthropomorphic phantom loaded with MOSFET detectors

The locations of the MOSFETs were chosen to sample organs of interest that could be exposed during the process of the abdomen/pelvis protocol. The parameters of dual-energy GSI abdomen/pelvis protocol can be seen in Table 4.

<table>
<thead>
<tr>
<th>Table 4: Dual-energy protocol</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>kVp</strong></td>
</tr>
<tr>
<td>mA</td>
</tr>
<tr>
<td>CTDI Vol. (mGy)</td>
</tr>
<tr>
<td>DLP (mGy-cm)</td>
</tr>
</tbody>
</table>
The loaded phantom was scanned four times using the dual-energy GSI abdomen/pelvis protocol. In order to accurately reflect dose to specific organs/tissues several corrections were necessary. The dose to bone marrow was corrected using published estimations of the distribution of active bone marrow in adults. [15] This allowed us to scale our measured dose in locations corresponding sites of bone marrow storage. Dose to bone surface was corrected using percentages of skeletal mass for the bone structures measured in our study. [16]

The dose to skin surface was calculated using “the rule of nines” [17], which is traditionally in medicine to estimate the percentage of a person’s skin affected by a burn. This rule separates the skin surface into nine anatomical regions and applies a surface area percentage fraction to each. The surface area factors used for this experiment were chest-0.09, abdomen- 0.09, and back of torso- .18. Using this method we calculated 36% of the total skin surface area was exposed during the scan.

Volumetric corrections [18] were also performed for dose measurements of the lung and esophagus. These were performed by cross-referencing our images with anatomical information of the adult phantom in order to calculate what percentage of each organ was present in our images.
1.3 Results

The effective energy of the combined dual-energy beams was found to be 42.8 kV, corresponding to an f-factor of 0.923. The effective energy of the 80 kV beam used in calibration was found to be 44.7 kV. When testing our calibration using the CTDI body phantom the dose measurements were 1.71 ± 0.06 cGy and 1.71 ± 0.01 cGy from MOSFET and ion chamber respectively, as was shown previously in Table 3.
Organ dose measurements recorded by MOSFETs can be seen below in Figure 9. Using these measurements combined with ICRP 103 tissue weighting factors effective dose was calculated as $17.42 \pm 0.05$ mSv for adult female, and $12.62 \pm 0.04$ mSv for adult male. Effective dose estimated using AAPM Dose Length Product (DLP) method was 14.6 mSv.

![Organ Dose (cGy)](image)

**Figure 10: MOSFET organ dose measurements**

### 1.4 Conclusion

This study shows that using the effective energy of the combined dual-energy beams is a viable way to calibrate MOSFET detectors in a dual-energy environment. The method is useful particularly in fast-kV switching systems where the high and low kV beams cannot be separated for individual treatment.
1.5 Discussion

The discrepancy in effective energy of the dual-energy environment and our calibration beam can be seen in the table 5. The difference can be seen to be 4.4% which was considered acceptable due to the accuracy of the MOSFETs in this experimental design being nearly 10%. [19]

Table 5: Effective energy of GSI protocol vs calibration beam

<table>
<thead>
<tr>
<th>Effective Energy</th>
<th>Dual-Energy</th>
<th>Calibration Beam</th>
</tr>
</thead>
<tbody>
<tr>
<td>% difference</td>
<td>4.4</td>
<td></td>
</tr>
</tbody>
</table>

Our estimation of effective dose for an adult female is higher than estimates using DLP method by 19%. Other studies comparing effective dose measurements using Monte Carlo methods and DLP method for dual-energy have shown a similar disagreement when using ICRP 103 tissue weighting factors [20]

1.6 Future Work

Moving forward we would like to repeat the experiment using a different CT scanner for calibration that will allow us to produce an 80 kV beam with effective energy closer to that of the dual-energy environment. We would also seek to apply this method to other dual-energy systems such as Dual Source CT.
2. Evaluation of Phillips DoseAware Personal Dose Meter System

The Phillips DoseAware system is a novel approach to providing real-time information on dose to hospital radiation workers. The system consists of personal dosimeters worn by the workers wirelessly connected to a wall mounted LED display. The display shows a real-time readout of radiation dose rate of each of the connected dosimeters, and catalogs this quantity vs time. The primary purpose of the system is to allow radiation workers to directly observe the radiation dose rate they are receiving, in hopes that this information will encourage safer work practices. Publications on this topic have shown a sharp reduction in dose to interventional radiology staff when given real-time feedback on their radiation exposure [21]. However there are currently no studies providing information on the physical performance of the system. It is important to investigate the radiological performance of the DoseAware Personal Dose Meter (PDM) in order to ascertain the validity of the exposure information being relayed to the hospital staff. The goal of this study was two-fold: (1) to initially test the manufacturer listed performance specifications of the system, and (2) to compare the performance of the DoseAware system with other commercially accepted detectors in a radiology environment.
2.1 Verification of Manufacturer Listed Specifications

2.1.1 Intro

The manufacturer listed performance specifications of interest to this study were dose rate response range, energy dependence, and angular dependence of the personal dosimeters. These specifications can be seen in Figure 11.
Dose Rate Range

- ±10% 40 μSv/hr – 150 mSv/hr
- ±20% 150 mSv/hr - 300 mSv/hr

Energy Dependence

- ±20% $\bar{E} = 33 - 84$ kev
- ±30% $\bar{E} = 84 - 101$ kev

Angular Dependence

- ±5% within within ±5°
- ±30% within within ±50°
- ±100–200% within within ±90°

Figure 12: Phillips DoseAware PDM manufacturer listed specifications [22]

2.1.2 Materials and Methods

The Precision X-ray X-RAD 320 (Precision X-ray, North Branford, CT) was used to irradiate the PDM for our experiment. An additional filtration of 0.1 mm Cu + 2.5 mm Al + 0.5 mm Be (F4 Filter) was used during the course of our experiment in order to increase the mean energy of the various X-ray beams. Both the ion chamber and PDM were placed at the bottom of the irradiation chamber at a distance of 100 cm.
Five sets of beam parameters were used in testing the exposure rate and energy dependence of the PDM. For each of these beams the tube current and run time were held constant at 1.0 mA and 30 seconds respectively. The kVp was changed between sets to acquire data for 40, 60, 80, 90, and 100 kVp. This had the effect of both changing dose rate and energy simultaneously. Exposure rate was not held constant during the investigation of energy dependence due to system lower limits on tube current setting.

In order to find the mean energy of each beam we used the SpekCalc spectrum analyzing software [23] to simulate the X-ray spectrum for each set of parameters. In this software we entered the peak voltage of the beam, along with the amount of filtration to generate out mean energy.
When investigating angular dependence four sets of irradiations were performed at 80 kVp, 1.0 mA, and 30 second run time. Styrofoam wedges were cut to place the PDM at angles representative of 15, 20, 30, and 45 degree displacements. The figure below gives a view of how the wedges were implemented.
### 2.1.3 Results

Our measurements for the dose rate dependence portion of our experiment can be seen in the Table 6. Both the ion chamber measurements and PDM measurements in the table are averages over five runs at each dose rate.

#### Table 6: Exposure rate dependence test. Ion chamber vs Phillips PDM

<table>
<thead>
<tr>
<th>Dose Rate (mSv/hr)</th>
<th>IC Exposure (mR)</th>
<th>PDM Exposure (mR)</th>
<th>% Error</th>
</tr>
</thead>
<tbody>
<tr>
<td>12.66</td>
<td>10.55</td>
<td>9.08</td>
<td>13.08</td>
</tr>
<tr>
<td>64.75</td>
<td>53.96</td>
<td>54.64</td>
<td>1.57</td>
</tr>
<tr>
<td>151.65</td>
<td>126.38</td>
<td>123.00</td>
<td>3.15</td>
</tr>
</tbody>
</table>
Due to limitations of the X-RAD 320 system we were only able to test energy ranges from 31-54 kV. Over this range we observe the PDM operates within the manufacturer’s specifications. Our study shows the PDM operates within ±6% for energy within the range specified by the manufacturer, and operates within ±13.1% slightly under this specified range.

**Table 7: Energy dependence test. Ion chamber vs. Phillips PDM**

<table>
<thead>
<tr>
<th>kVp</th>
<th>Mean Energy</th>
<th>IC Exposure (mR)</th>
<th>PDM Exposure (mR)</th>
<th>% Error</th>
</tr>
</thead>
<tbody>
<tr>
<td>40</td>
<td>31.1</td>
<td>10.55</td>
<td>9.08</td>
<td>13.08</td>
</tr>
<tr>
<td>60</td>
<td>40.1</td>
<td>53.96</td>
<td>54.64</td>
<td>1.57</td>
</tr>
<tr>
<td>80</td>
<td>47.6</td>
<td>126.38</td>
<td>123.00</td>
<td>3.15</td>
</tr>
<tr>
<td>90</td>
<td>52</td>
<td>175.28</td>
<td>169.50</td>
<td>4.16</td>
</tr>
<tr>
<td>100</td>
<td>54</td>
<td>229.42</td>
<td>220.60</td>
<td>5.17</td>
</tr>
</tbody>
</table>

The results of our angular displacement runs can be seen below. Again, the exposure values listed are averages over 5 exposures at each angle. We see that for displacements of 15–45° that the PDM accuracy is within ±15.4%, but there does not
seem to be a clear correlation between the accuracy of the PDM and the angle at which it is placed in the beam.

Table 8: Angular dependence test. Ion chamber vs. Phillips PDM

<table>
<thead>
<tr>
<th>Angle Displacement</th>
<th>IC Dose (mR)</th>
<th>PDM Dose (mR)</th>
<th>% Error</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>126.38</td>
<td>123.00</td>
<td>3.15</td>
</tr>
<tr>
<td>15</td>
<td>127.8</td>
<td>214.2</td>
<td>2.83</td>
</tr>
<tr>
<td>20</td>
<td>128.1</td>
<td>108.4</td>
<td>15.4</td>
</tr>
<tr>
<td>30</td>
<td>127.2</td>
<td>109.6</td>
<td>13.8</td>
</tr>
<tr>
<td>45</td>
<td>128.7</td>
<td>125.8</td>
<td>2.22</td>
</tr>
</tbody>
</table>

2.2 Performance Testing in Radiology Environment

2.2.1 Materials and Methods

A Philips Allura XPERA FD 20/20 (Phillips, Amsterdam, Netherlands) fluoroscopy system was used to evaluate the performance of the Phillips DoseAware system in an interventional radiology environment where it would most likely be used. An adult anthropomorphic phantom torso was used to provide a realistic scattering medium to best recreate dose from scatter radiation in a typical procedure. Three ion chambers (451, Fluke Biomedical, Everette, WA) and three RadEye G (Thermo Fisher Scientific, Waltham, MA) exposure rate meters were used to compare the dose readings of the Phillips PDMs at each of three locations. Three locations were chosen to represent
locations where radiology personnel would likely be stationed during a typical procedure. The three locations chosen were physician standing beside patient, a nurse station, and technician work station

![Diagram showing three locations: Physician, Nurse Station, Technician]  

**Figure 16: Geometry of experiment**

At each location one Phillips PDM and one RadEye were placed side by side with the respective active areas normal to direction of incoming scatter. A low dose fluoroscopy scan was performed on the phantom torso 3 times to achieve readings for the detectors. The parameters of the fluoroscopy scan can be seen in Table 9. In this table DAP refers to Dose Area Product. This quantity is defined as the absorbed dose in air from the machine multiplied by the area irradiated. The displayed DAP output is
internally calculated by the machine. The PDM and RadEye detectors were then replaced with ion chambers and the scans were repeated.

Table 9: Low dose fluoroscopy parameters

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>kVp</td>
<td>78</td>
</tr>
<tr>
<td>mA</td>
<td>8.9</td>
</tr>
<tr>
<td>DAP (mGy⋅cm^2/s)</td>
<td>80</td>
</tr>
<tr>
<td>Exposure Time (seconds)</td>
<td>120</td>
</tr>
</tbody>
</table>

The Phillips PDM and RadEye detectors were then repositioned and three scans were performed with the Philips Allura XPERA in cine angiograph mode. The
parameters of the cine scans can be seen in Table 10. Once again the three scans were repeated with the ion chambers in place of the Phillips PDM and RadEye detectors. Due to the nature of the cine angiograph scan the displayed values on the ion chamber changed over the course of the exposure. Each value was recorded and averaged for each of the three ion chambers.

By taking measurements at three distances from the phantom, and operating the machine in both low dose fluoroscopy, and cine angiography mode we were able to measure a wide range of dose rate in order to assess the Phillips PDM

<table>
<thead>
<tr>
<th>Table 10: Cine-Angiograph mode parameters</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>kV</strong></td>
</tr>
<tr>
<td>mAs</td>
</tr>
<tr>
<td>Exposure Time (seconds)</td>
</tr>
</tbody>
</table>

2.2.2 Results

The detector response of the Phillips PDM and RadEye dosimeters was plotted against dose rate recorded by ion chamber (as gold standard). The performance of both detectors can be seen in Figure 18 and Figure 19.
Figure 18: Phillips PDM detector response vs ion chamber
2.3 Discussion

Our examination of dose rate and energy dependence were performed using the same set of data. This is a less than ideal scenario due to the fact that we were unable to hold one of these constant while examining the other. This could produce a result that is less scientifically significant for both dose rate dependence and energy dependence of the PDM. However, we felt for the scope of the initial specification testing, which is to obtain a baseline idea of the PDM accuracy that this experimental setup would suffice.

The results of our interventional radiology experiment show unexpected behavior in the RadEye detectors. While our intentions were not to test the performance
of the RadEye specifically, our data suggests that the detector fails in the presence of pulsed beam radiation above a dose rate of 180 mR/hr. The RadEye G specifications state a dose rate range of up to 10 R/hr, which is clearly not echoed in our experiment. This finding is troubling due to the widespread commercial use of the RadEye dosimeter.

2.4 Conclusions

Throughout this chapter we have discussed our investigation into the technical and radiological performance of the Phillips DoseAware PDM. We feel the system operates in a satisfactory manner within the range of parameters advertised by the manufacturer, and could be a valuable tool for physicians and health staff working with radiation sources. It is important to note that the Phillips DoseAware system is not intended to be used as a primary personal dosimeter, but rather serve as ancillary device to provide real-time feedback during procedures. While the PDM does not show perfect agreement with ion chamber, we feel it performs adequately for it’s intended function.

2.5 Future Work

Moving forward we would like to further investigate the behavior of the RadEye detector in the presence of high dose rate radiation. Specifically testing this in both pulsed beam and continuous fields of radiation in order to diagnose the cause of the detector response plateau we observed.

3.1 Introduction

The purpose of this project was to estimate the effective dose from novel digital radiography protocols used to image renal stones. Currently renal stones are generally diagnosed using Computed Tomography (CT). Ongoing research at Duke Urology has proposed four digital tomography protocols to image these stones in hopes of lowering dose to patients. Effective dose was estimated by first measuring the skin entrance dose to an adult anthropomorphic phantom (Model 702-BR190, CIRS) for each protocol. After this effective dose for each protocol was simulated using PCXMC Monte Carlo software (STUK, Helsinki, Finland).

3.2 Materials and Methods

3.2.1 Experimental Setup

The protocols for this study were performed using a GE Discovery XR656 (GE Healthcare, Chicago, IL). This machine consists of an X-ray tube that sweeps over the patient in a cranio-caudal manner, acquiring projections over the course of the movement of the tube.

Due to the ongoing nature of the study by Duke Urology this report will not go into complete detail when discussing the parameters of the protocols, but will instead focus on the tools and methods employed in estimating effective dose for each. Table 11 shows details of the protocols that were important in our study.
Table 11: Renal stone protocol parameters

<table>
<thead>
<tr>
<th>Protocol #</th>
<th>Protocol Type</th>
<th>Sweep Angle</th>
<th># of Projections</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Abdomen</td>
<td>20</td>
<td>25</td>
</tr>
<tr>
<td>2</td>
<td>Abdomen (large patient)</td>
<td>20</td>
<td>25</td>
</tr>
<tr>
<td>3</td>
<td>Pelvis</td>
<td>40</td>
<td>60</td>
</tr>
<tr>
<td>4</td>
<td>L Spine</td>
<td>30</td>
<td>60</td>
</tr>
</tbody>
</table>

The sweep angle parameter refers to the full range of motion of the X-ray tube. When the tube is directly overhead of the image receptor it is considered at angle 0. Therefore a sweep angle of 20 corresponds to the X-ray tube starting at an angle of 10 and ending at angle -10 with respect to the original centered position.

A 0.18cc ion chamber (105x-0.18, Radcal, Monrovia, CA) was placed on the torso surface of an adult anthropomorphic phantom (Model 702-BR190, CIRS) positioned supine on the patient table. The ion chamber was placed roughly in the center of the scanning region for each protocol, in order to measure skin entrance exposure. Total exposure measurements for 3 positioning scouts and three scans were recorded for all of the protocols and later divided by three to gain skin entrance exposure for each protocol.

Table 12: Ion chamber entrance exposure

<table>
<thead>
<tr>
<th>Protocol</th>
<th>Entrance Exposure (mR)</th>
</tr>
</thead>
</table>

35
3.2.2 Characterizing protocols for use in PCXMC

PCXMC is a computer program that uses Monte Carlo techniques to estimate patient organ dose and effective dose during medical X-ray examinations. In order to estimate effective dose using the PCXMC software each protocol needed to be characterized as a series of individual projections. The geometric input parameters of PCXMC can be seen in the figure below.

<p>| | |</p>
<table>
<thead>
<tr>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>47.28</td>
</tr>
<tr>
<td>2</td>
<td>104.8</td>
</tr>
<tr>
<td>3</td>
<td>47.05</td>
</tr>
<tr>
<td>4</td>
<td>64.05</td>
</tr>
</tbody>
</table>
The important parameters that change as the X-ray tube sweeps over the patient are field size and cranio-caudal angle. The collimation of the beam is constant as it sweeps over the patient, however as the projection angle changes the field shape at the surface of the patient will geometrically distort. To account for this we characterized each tomography protocol as a summation of individual X-ray projections.

The projections were chosen to be at integer cranio-caudal angles over the range of the total sweep angle for each protocol. A MATLAB (MathWorks Inc., Natick, MA) script was written to calculate the equivalent square field size at the surface of the
phantom for each projection angle. This resulted in 21, 21, 41, 31 projection files for protocols 1, 2, 3, and 4 respectively.

The next step in using the PCXMC software was to enter information on the X-ray tube used to produce each projection. This information included X-ray tube potential, anode angle, and inherent filtration. The anode angle for the Discovery XR656 was 12.5 and the inherent filtration was 2.7 mm Al [24] Figure 21 shows the X-ray spectrum input screen. This information was input for all of the projection files as none of the parameters changed between protocols.

![Figure 21: PCXMC X-ray spectrum input window](image)

With the X-ray spectrum calculated the next value needed was a dose quantity measurement for each projection file. The available options for this quantity are incident air kerma, dose-area product, entrance exposure, exposure-are product, and current-
time product. Our experimental setup recorded skin entrance exposure using the 0.18cc ion chamber. The measured skin entrance exposure for each protocol was divided evenly among the projection files it was composed of. This provided an estimate of skin entrance for each of the projection files to be simulated using PCXMC.

![Figure 22: PCXMX dose quantity input window](image)

The above process was performed on all projection files for each protocol. PCXMC outputs a simulation of organ dose and effective dose using ICRP 103 tissue weighting factors. This information was exported to an Excel (Microsoft, Richmond, WA) spreadsheet where the effective dose of the projection files were combined for each of the 4 protocols.
3.3 Results

The effective dose of the four protocols can be seen in Table 13.

Table 13: Effective dose for renal stone protocols

<table>
<thead>
<tr>
<th>Protocol</th>
<th>ICRP 103 Effective Dose (mSv)</th>
<th>Error (mSV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.98</td>
<td>0.002</td>
</tr>
<tr>
<td>2</td>
<td>2.18</td>
<td>0.006</td>
</tr>
<tr>
<td>3</td>
<td>2.02</td>
<td>0.004</td>
</tr>
<tr>
<td>4</td>
<td>2.02</td>
<td>0.005</td>
</tr>
</tbody>
</table>

3.4 Conclusions

Our results show that the effective dose from the proposed renal stone imaging protocols is lower than typical values seen in CT examinations for the same purpose. [25] [26]. Provided that these images prove to be of adequate quality to diagnose the presence of renal stones this could be a beneficial low dose alternative to CT.

3.5 Future Work

Moving forward our study could benefit from a more accurate model of each of the protocols. This could be accomplished by splitting each up into non-integer projection angles that correspond more closely to the actual behavior of the X-ray tube while performing the protocol. This alteration would likely effect the resulting effective dose only marginally, but could perhaps more accurately report individual organ dose
Appendix A: MATLAB code for calculating field size of digital radiography renal stone protocols

clear all

sid= 73.025; % [cm] shortest value recorded, assumed central projection
% We know the FOV is 35x41 at the original SID of 41 inches (104.14 cm)
% to get our initial, central projection field size we simply have to
% adjust to what this field size would be at our measured ssd of 73.025

colx=24.6;
coly=28.8;

numproj=25; % Number of projections, changed for each protocol
sweep_ang=20; % changed for each protocol

rot=linspace(0,20,21);

div_ang_35=atand(colx/2/sid);
sid_ang=sid./cosd(rot); % distance from center line of beam to image receptor

left_ang_35= abs(rot-div_ang_35);
dis_l35=sid./cosd(left_ang_35);
x2l=(dis_l35.^2-sid^2).^0.5;

right_ang_35= rot+div_ang_35;
dis_r35=sid./cosd(right_ang_35);

% distance from point source to right beam edge along x axis
x2r=(dis_r35.^2-sid^2).^0.5;

for i=1:numel(rot)
    if rot(i) > div_ang_35
        projection_35_x(i)=x2r(i)-x2l(i);
    else
        projection_35_x(i)=x2r(i)+x2l(i);
    end
end

width_l=coly/dis_l35(1).*dis_135;
width_r=coly/dis_r35(1).*dis_r35;

%now to find our equivalent square
% using ES = (4*Area)/(Perimeter)

area=width_l.*projection_35_x + (width_r-width_l).*projection_35_x;

perimeter= width_l+width_r + 2.*(((width_r-
width_l)./2).^2+projection_35_x.^2).^5;

ES= 4.*area./perimeter;

Angle= rot.';
Equiv_Sq=ES.';

T= table(Angle,Equiv_Sq)
References


[22] Phillips DoseAware Bay Station Instructions of Use, Document version 1.2, 2009


