The Use of a Novel Radiation Detector on Quantifying PET/Nuclear Medicine
Occupational and Non-occupational Doses and Calibration of MOSFET Radiation Detectors against Effective Energy

by

Lei Ding

Graduate Program in Medical Physics
Duke University

Date:_______________________

Approved:

___________________________
Terry Yoshizumi, Supervisor

___________________________
Rathnayaka Gunasingha

___________________________
Neil Petry

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

2013
ABSTRACT

The Use of a Novel Radiation Detector on Quantifying PET/Nuclear Medicine Occupational and Non-occupational Doses and Calibration of MOSFET Radiation Detectors against Effective Energy

by

Lei Ding

Graduate Program in Medical Physics
Duke University

Date:_______________________

Approved:

___________________________
Terry Yoshizumi, Supervisor

___________________________
Rathnayaka Gunasingha

___________________________
Neil Petry

An abstract of a thesis submitted in partial fulfillment of the requirements for the degree of Master of Science in Medical Physics in the Graduate School of Duke University

2013
Copyright by
Lei Ding
2013
Abstract

Project 1: Dose reduction for PET technologists by the automatic dose draw/injection system

Purpose: To evaluate the dose reduction by the installed automatic dose draw/injection machine.

Materials & Methods: Six RadEye detectors were given to six PET technologists. A RadEye detector recorded data every 25 seconds throughout the day. Technologists logged their activities as follows: dose draw/injection, patient positioning, patient transport, patient care and non-specific. One technologist performed dose drawing/injection manually while others used the Trasis system. The Trasis machine was monitored with a Radeye detector during the period as well.

Results: The average dose reduction brought by Trasis is 75% for dose draw and 70% for dose injection. Qualitatively, instead of dose draw/injection, patient positioning has become the most significant contributing factor to overall PET technologist dose. In addition, the current average daily dose for a PET technologist is about 0.03 mSv, which on average is 36% less than before [12]. PET technologists typically received a dose of 0.007 mSv from dose draw/injection, 0.005 mSv from patient transport, 0.013 mSv from patient positioning, 0.001 mSv from patient care, and 0.003 mSv from non-specific per working day. This would result in an annual dose of 8 mSv which is approximately 16% of occupational dose limit (50 mSv).
**Conclusions:** The installation of automatic dose draw/injection machine has clear benefits to the PET technologists. The radiation doses for PET technologists are well within the annual limit of doses to occupational radiation workers.

**Project 2: Validation of ceiling shielding in CT/PET room with RADEYE**

**Purpose:** To measure and the magnitude of scattered radiation levels in CT/PET suite and to evaluate the effectiveness of shielding of the ceiling.

**Materials & Methods:** Six RadEye detectors were placed in the CT/PET room, four in the ceiling, and two at one meter above floor. A RadEye detector recorded data every 25 seconds throughout the day. The detector was turned on at the beginning of the day (6 am) and the doses were transferred to a laptop for analysis at the end of the day (5 pm). The dose to a non-radiation worker above the CT/PET room was estimated based on the ceiling data. The magnitude of transmitted CT radiation in the room above was measured separately with RadEye.

**Results:** The CT dose contributed about 80% of the total dose while the PET contributed 20% within the scanning room. No dose contribution was measured above the floor from CT scanning. The combined dose from both PET and CT scan in a room above at 2.2 meter was $2.4 \times 10^{-6}$ mSv per week, assuming an occupancy factor of 1.

**Conclusions:** This study quantified the CT and PET doses contributions separately in the clinical CT/PET room. An analytical model was developed to calculate the non-occupational personnel dose above the CT/PET room and the calculated results were
confirmed by physical measurements. The actual physical dose was much smaller than the NCRP design goals of 0.02 mSv/wk.

Project 3: Evaluating MOSFET dependency on effective energy over diagnostic energy range

**Purpose:** To characterize MOSFET calibration factors (CF) as a function of effective energy over diagnostic energy range.

**Materials & Methods:** Five new MOSFETs were used in the study. The calibration factors were measured in two ways: 1) fixed kVp, fixed SSD, fixed FOV, and varying filtration; and 2) fixed filtration, fixed SSD, fixed FOV, and varying kVp. Effective energy was computed as a function of kVp and filtration by SpekCalc.

**Results:** CF was independent with HVL in the range of HVL = 5 to HVL = 9mm Al at a fixed 120 kVp. CF depended linearly with kilo-voltage (kVp) from 80 to 140 kVp at a fixed filtration. In addition, a strong non-linear correlation of average CF versus effective energy was generated for effective energies in the diagnostic range (Goodness of fit of 0.98).

**Conclusions:** A correlation of second degree polynomial was seen between calibration factor and effective energy over diagnostic range. Hence, we created a calibration curve so that under a given fixed kVp or filtration, the calibration factor is automatically generated. A high correlation between CF versus effective energy was found over the diagnostic energy from 45 keV to 65 keV. This suggests that we could estimate the
calibration factor with in-house generated MOSFET aging data, which would have a
direct impacted CF linearly.
## Contents

Abstract ......................................................................................................................................... iv

List of Tables ................................................................................................................................ xii

List of Figures ............................................................................................................................ xiii

Acknowledgements .................................................................................................................... xv

1. Introduction to Radiation Dosimetry ..................................................................................... 1
   1.1 Overview ........................................................................................................................... 1
   1.2 Basic Quantities and Units in radiation dosimetry ...................................................... 1
   1.3 Ion Chamber ...................................................................................................................... 4
   1.4 Geiger Muller Counter ..................................................................................................... 5
   1.5 Metal-Oxide Semiconductor Field-Effect Transistor (MOSFET) ............................... 5
   1.6 Film badges ....................................................................................................................... 6
   1.7 RadEye Detector ............................................................................................................... 7
   1.8 Piranha Detector ............................................................................................................... 8

2. Project 1: Evaluation of dose reduction by the automatic dose draw/injection machine at Duke PET/Nuclear Medicine facility ..................................................................................... 9
   2.1 Introduction ..................................................................................................................... 10
   2.2 Materials and Methods .................................................................................................. 11
   2.3 Results and Discussion .................................................................................................. 15
      2.3.1 Average dose reduction by procedure ................................................................... 16
      2.3.2 Qualitative dose distribution from each tasks ....................................................... 18
      2.3.3 Daily dose reduction .............................................................................................. 19
      2.3.4 Comparison with radiation film badge data .......................................................... 19
      2.3.5 Additional doses brought upon by the Trasis machine ........................................ 20
Appendix D Experimental set-up used by Das et al. to study backscatter from high Z materials ....................................................................................................................................... 36

References .................................................................................................................................... 38
List of Tables

Table 2.1: Example entry from the log sheet. ................................................................. 12
Table 2.2: The main tasks for each technologists during the day ......................... 13
Table 2.3: Estimated Annual Doses for technologists. .............................................. 22
Table 3.1: Constants used in the Berger’s formulas.................................................. 30
Table 3.2: Build-up factors calculated by Berger’s formulas ...................................... 30
Table 3.3: Broad beam transmission factors at 511 keV comparing to AAPM TG108 .... 30
Table 3.4: Average Daily dose (mR) recorded by the six detectors....................... 31
Table 3.5: Calculated radiation doses after the ceiling shielding............................ 34
Table 3.6: Summary of dose parameters................................................................. 35
Table 3.7: Comparison of calculated PET doses and measured PET doses inside the room .................................................................................................................. 35
Table 3.8: Doses calculated with 90 incident angle .................................................. 36
Table 3.9: Doses calculated with 15 degree incident angle ...................................... 37
Table 4.1: HVL and kVp used in the study with the measured CF ....................... 47
Table 4.2: Calculated backscattered photon energies lowest and highest incident effective energies based on Equation 4.3 ............................................................. 49
List of Figures

Figure 1.1: Radiation quantities used in radiation dosimetry ........................................ 2

Figure 1.2: Simple schematic for an ion chamber .......................................................... 4

Figure 1.3: Simple schematic of a MOSFET ................................................................. 6

Figure 1.4: A snap shot of RadEye detector ............................................................... 8

Figure 1.5: A snap shot of Piranha detector ............................................................... 9

Figure 2.1: Position of RadEye detector on the Trasis machine ............................... 14

Figure 2.2: Average dose reduction by procedures .................................................... 16

Figure 2.3: Mean daily dose contributed from each task type ................................. 17

Figure 2.4: Average daily doses before and after implementation of Trasis system . 18

Figure 2.5: Whole body dose data collected by RadEye and film badge separately before and after using the Trasis system ................................................................. 20

Figure 2.6: Finger dose data collected by RadEye and film badge separately before and after using the Trasis system ................................................................. 20

Figure 2.7: Dose distributions from each task for Technologist #5. .......................... 21

Figure 3.1: Geometry indicating the location of detectors and the PET/CT suite layout ..................................................................................................................................... 27

Figure 3.2: A sample spectrum from a typical PET/CT scan ........................................ 28

Figure 3.3: Average daily doses recorded by the six detectors within the room .... 31

Figure 3.4: Dose received by the six detectors within the room from PET radiation, normalized to each scanning procedure ......................................................... 32

Figure 3.5: Dose received by the six detectors within the room from CT radiation, normalized to each scanning procedure ......................................................... 33

Figure 3.6: Illustration of difference of thickness calculated by AAPM and in reality ......................................................................................................................................... 38
Figure 4.1: Experiment set-up using the Phillips Diagnostic x-ray tube ...............42
Figure 4.2: Experimental set-up using x-rad 320 irradiator..........................43
Figure 4.3: A sample spectrum calculated by SpecCalc software ......................44
Figure 4.4: Calibration factors versus HVL (mm Aluminum) at 120 kVp ..........45
Figure 4.5: Calibration factors versus kVp at a fixed filtration (HVL=6.6 mm Al).......46
Figure 4.6: Calibration Factors as a function of effective energy .......................47
Acknowledgement

First, I would like to thank the following people, for their kindness and efforts that ensured the successful completion of this thesis:

- Dr. Terry Yoshizumi Ph.D.
- Dr. Gunasingha Rathnayaka Ph.D.
- Neil Petry M.S.
- Giao Nguyen M.S.

I would also like to thank the PET technologists participated in this study for their support.

Furthermore, I would like to thank my fellow students, for their helpful discussions whenever I have questions.

Finally, I want to thank my parents deeply for their trust and confidence in me that carry me through puzzled times and enlighten me with the path ahead.
1. Introduction to Radiation Dosimetry

1.1 Overview

Radiation dosimetry involves the measurement and assessment of deposited energy by indirect or direct exposure to ionizing radiation in a medium. It is highly applicable both in radiation protection and radiation therapy. In this thesis, we first discuss two common issues encountered in PET/Nuclear Medicine radiation protection, which include doses to nuclear medicine technologists and the effectiveness of ceiling shielding for a PET/CT room; and an issue of MOSFET calibration methodology in order to accurately determine radiation doses at various half value layer.

Before going into details of experimental set-up and results of these projects, it is important to provide background theories and discuss principles of instruments used in the study. These include basic quantities and units used in radiation dosimetry, the basic operational principles of radiation detectors, including ion chamber, RadEye detector and MOSFET (Metal Oxide Semiconductor Field Effect Transistor) dosimeter, film badge, and Piranha, a filtration measuring device.

1.2 Basic Quantities and Units in radiation dosimetry

This section describes basic quantities and units of radiation involved in these projects. The relationship between quantities used in radiation dosimetry can be described in Figure 1.1 [1,2].
**Exposure, X**

The exposure measures the amount of charges liberated as x ray or γ ray (under 3 MeV) interact in a small volume of air. The unit for exposure is Roentgen (R), which is expressed by Coulombs per kilogram [1].

\[
1R = 2.58 \times 10^{-4} \frac{C}{kg} \quad (1.1)
\]

**Absorbed dose, D**

The absorbed dose is the energy absorbed per unit mass from any type of radiation in any type of material. The unit for absorbed dose is Gy (gray), and

\[
1 \text{ Gy} = 1 \frac{J}{kg} \quad (1.2)
\]
The absorbed dose in a medium can be calculated from exposure in air and the f-factor, where f-factor is proportional to the ratio between mass attenuation coefficients of medium and air [2].

\[ D_{med} = f \times X_{air} \]  \hspace{1cm} (1.3)

\[ f = 0.876 \times \frac{\mu_{medium}}{\mu_{air}} \]  \hspace{1cm} (1.4)

**Equivalent Dose, \( H_R \)**

The equivalent dose takes into account of the biological effect of ionizing radiation, which is differ for each type of radiation. The SI unit of equivalent dose is the Sievert (Sv). It is given by

\[ H_{R,T} = \sum_R W_R D_{R,T} \]  \hspace{1cm} (1.5)

where \( W_R \) is the radiation weighting factor. The radiation weighting factors used are currently given by ICRP 103 [3].

**Effective Dose, \( E \)**

The effective dose represent an overall dose that would produce an equivalent detriment to the health of the individual taking into account the different radiosensitivities of different types of tissues of the human body. The SI unit of effective dose is also Sievert (Sv). It is determined as
\[ E = \sum T W_T H_{R,T} \] (1.5)

where \( W_T \) is the tissue weighting factor [3].

**1.3 Ion Chamber**

Ion chambers are widely used air-filled radiation detectors, and it serves as the gold standard for radiation dosimetry for photons and x-rays [2]. Simple schematic for ion chamber is provided in Figure 1.2.

![Figure 1.2 Simple schematic for an ion chamber [1]](image)

As shown in Figure 1.2, charges are collected as radiation passes through the detector. A potential difference is applied across the chamber in order to collect charges. The current from generated charges is proportional to the energy of the radiation. Task group 61 (TG61) of the American Association of Physicists in Medicine (AAPM) describes a protocol in using ion chamber for x-ray dosimetry at diagnostic range (40 – 300 kV) [4].
The ion chamber used was calibrated to National Institutes of Standards and Technology (NIST) traceable source and applied correction factors to include 1) temperature pressure correction, 2) ion chamber collection efficiencies, 3) electrometer correction factor, and 4) polarity effects [4].

1.4 Geiger Muller Counter

Geiger Muller Counter is a gas-filled detector that operates similarly to the ion chamber but at a different voltage [2]. Geiger Muller Counter operates at a high voltage such that gas multiplication occurs. Gas multiplication is a process where the liberated electrons from ionization have enough energy to create other ionizations along their paths. Thus, an electron cascade is created and the resulting pulse is measured. The process is terminated with the use of quenching gases, which are usually halogens (Cl, Br, etc) [2]. The Geiger Muller counter is usually used as radiation counter because they will produce the signal of detection of radiation regardless the amount of ionization deposited [2].

1.5 Metal-Oxide Semiconductor Field-Effect Transistor (MOSFET)

Metal-Oxide Semiconductor Field-Effect Transistor (MOSFET) is another type of radiation detector. A schematic of the P channel MOSFET detector is shown in Figure 1.3. The source and the drain are two terminals located on top of a positively doped (p-type) silicon area. The substrate is a negatively doped (n-type) silicon substrate. When a sufficiently large negative voltage is applied to the polysilicon gate a significant number
of holes will be attracted to the oxide/silicon surface from both the bulk silicon substrate and the source and drain regions. Once sufficient holes have accumulated there, a conduction channel is formed, allowing current to flow between the source and drain. The voltage necessary to initiate current flow is known as the device threshold voltage ($V_{TH}$) [5]. Ionizing radiation produced electron-hole pairs in the silicon substrate, which move to the oxide-silicon interface where they become trapped causing a negative threshold voltage shift ($\Delta V_{TH}$) [5]. Therefore, the threshold voltage will change after each irradiation. Hence, the voltage shift is measured before and after exposure, and is proportional to dose [5].

![Figure 1.3 Simple schematic of a MOSFET [5]](image)

### 1.6 Film badges

Film badges are used as radiation detectors to measure occupational dose in diagnostic or therapeutic radiology at Duke University Medical Center (DUMC). Film badge detectors consist of a piece of personal monitoring film to allow for integrated dose measurements over a period [6]. To simulate different tissue depths, the radiation
will penetrate five different filters, resulting doses in the form of shallow dose equivalent (SDE) and deep dose equivalent (DDE) [6]. Shallow dose equivalent is the external exposure dose equivalent to the skin or an extremity at a tissue depth of 0.007 centimeters averaged over an area of 1 square centimeter [7]. Deep dose equivalent is the external whole-body exposure dose equivalent at a tissue depth of 1 cm [7].

**1.7 RadEye Detector**

RadEye Detector is a GM detector manufactured by Thermo Scientific (RadEye G, Franklin, MA). The detector reads exposure rates from 0.05 μSv/hr (5 μR/hr) to 0.1 μSv/hr (10 R/hr) [8]. The size of device is 9.6 by 6.1 centimeters on its face, and 3.1 centimeters thick [8]. The weight is 160 grams and is powered for 600 hours by three AAA alkaline batteries [8]. The Radeye unit records the mean and maximum values for exposure rate over a sampling period user specified. The device’s internal history maintains a history of the most recent 1600 sampling periods [8]. With proper settings, the device can store dose measurements every 25 seconds during a 8 hour working day.
1.8 Piranha Detector

Piranha detector (RTI electronics, Towaco, NJ) is a device used that’s frequently used in x ray QA measurements (Figure 1.5). In this thesis, the main purpose of using Piranha is to measure x ray beam quality, expressed in terms of half value layer (HVL). The unit of HVL is usually set as mm Aluminum (Al). The device has two internal detectors where one is on top with some fixed filtration in between. The doses were measured for both detectors and beam quality is determined by these values [9].
Figure 1.5 A snap shot of Piranha detector [9]
2. Project 1: Evaluation of dose reduction by the automatic dose draw/injection machine at Duke PET facility

2.1 Introduction

Positron emission tomography (PET) has become an established nuclear imaging modality that has proved its usefulness in many aspects. PET was invented at the Mallinckrodt Institute of Radiology at Washington University in the mid 1970s and is commonly in practice in areas of neurology, cardiology and oncology today [10]. Nowadays, thousands of PET scanners are in service around the world. However, the popularity of PET scans gave rise to concerns about the radiation exposure of staff members, in particular the technologists making contacts with the radionuclides. It is common that the dose to a PET technologist in a hospital setting is highest among all staff members [11]. At Duke medical center, a former student had categorized the doses to PET technologists due to technologists’ different activities or tasks using RadEye detectors [12]. These activities were broken down into dose draw/injection, patient position, patient transport, patient care and non-specific. Previously, it was found that the highest dose contribution of overall dose to the technologist was from the dose draw and injection task [12]. With the purchase of an automatic dose draw/injection machine, this study was repeated to determine the dose reduction from each task types as a result of the new machine.

The data and previous data were both measured by the RadEye detector as mentioned in section 1.5. The RadEye detector, which composes of a sensitive GM tube
with measuring range from 5 µR/hr to 10 R/h, was used for gamma/x-ray measurement. The biggest advantage for RadEye detector was that it could store the last 1600 measurements internally, which gave us a spectrum of doses throughout the day.

The automatic dose draw/injection machine has a trade name of Unidose® and is manufactured by Trasis Pharmacy (Trasis s.a., Belgium). The machine has several components allowing automatic radioactive solutions to dispense into a shield vial used for dose injection. Briefly, the machine includes 1) an area for transferring and mixing radioactive solutions with saline solutions 2) an area for attaching and storing the radioactive solution 3) an area for measuring and delivering the prepared radioactive solutions 4) a graphical interface and a label printer. The machine sits in the hot cell, and the maximum activity it can handle is up to 10 GBq of F18 while the measured activity 5 cm away from the machine records only 25 μSv per hour [25]. After the dose is prepared, the solution is filled in a cartridge which is then inserted into a shielded vial with 24 mm tungsten in total. Dose injection is facilitated with a push button, in order to help inject doses and rinse the cartridge [25].

2.2 Materials and Methods

Similar to previous study, five technologists in the PET clinic volunteered to wear a Radeye unit as they performed their daily duties. The Radeye detectors were placed at the lower pocket of the lab coat at approximately the wrist level. Technologists 1, 2 & 3 were involved in the previous study. They agreed to maintain a log of their interactions
with radioactive material as before. They were provided log sheet (Table 2.1) on which they specified:

1. The task they were performing
2. The radionuclide they were dealing with and the activity of the radionuclide
3. The starting time and ending time of the current task
4. Any notes that might help to clarify what they were doing

**Table 2.1 Example entry from the log sheet**

<table>
<thead>
<tr>
<th>Dose draw</th>
<th>Patient transport</th>
<th>F-18 mCi</th>
<th>Patient Care</th>
<th>Tc-99m mCi</th>
<th>Time Start</th>
<th>Time End</th>
<th>General Notes</th>
</tr>
</thead>
<tbody>
<tr>
<td>Patient Position</td>
<td>Patient Care</td>
<td>o</td>
<td>Other</td>
<td>o</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

As shown in Table 2.1, each interaction with radioactive material was placed in one of the following categories:

1. Dose Draw – Drawing prescribed dosage of FDG.
2. Dose Injection -Injecting the patient, and returning to the hot lab to perform the residual assay.
3. Patient Positioning – Any act of positioning the patient within the scanner and setting the patient up for IV contrast. This included re-positioning the patient during the scan, whether to adjust the headboard or to rotate the patient for a scan of the legs.
4. Patient Transport – All actions where the technologist escorted the patient in transit, such as escorting patients to the bathroom.
5. Patient Care – All direct contacts with the patient not covered by the previous categories. Examples included taking the patient a blanket or pillow or bringing the patient a drink of water.

Any exposures not logged as one of these categories was placed into a fifth category: unknown or nonspecific source. Nonspecific contribution was an important value because it represented exposure received without the employee knowing he/she was being irradiated.

Because different technologists had different job functions, their daily tasks would be slightly different. Table 2.2 summarizes the daily tasks of each technologist. Their tasks were the same as before as shown in Table 2.2, Tech #1 was only responsible for dose drawing & injection of F-18 radionuclides. Tech #2 & Tech #3 were technologists that involved in every tasks. Tech #4 was a technologist that handled draw/injection without using the automatic dose draw/injection machine.

**Table 2.2 The main tasks for each technologist during the day**

<table>
<thead>
<tr>
<th>Task Type</th>
<th>Tech 1</th>
<th>Tech 2&amp;3</th>
<th>Tech 4</th>
<th>Tech 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dose Drawing/Injection(F-18) using Trasis</td>
<td>•</td>
<td>•</td>
<td>•</td>
<td>•</td>
</tr>
<tr>
<td>Dose Drawing/Injection(F-18) <strong>without</strong> using Trasis</td>
<td></td>
<td></td>
<td>•</td>
<td></td>
</tr>
<tr>
<td>Dose Drawing for both F-18 &amp; Tc-99m</td>
<td></td>
<td></td>
<td></td>
<td>•</td>
</tr>
<tr>
<td>Patient Transport</td>
<td>•</td>
<td>•</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Patient Position</td>
<td>•</td>
<td></td>
<td>•</td>
<td></td>
</tr>
</tbody>
</table>
Patient Care

| Attach F-18 source into the Trasis machine | • | • | • |

Technologist 1 recorded for 9 business days, Technologist 2 & 3 recorded for 16 days, Technologist 4 recorded 5 days and Technologist 5 recorded for 3 days. In addition, a RadEye detector was placed onto the Trasis Machine outside the source holding region, where the detected radioactivity was the largest inside the hot cell (Figure 2.1). The purpose of placing this detector was to determine the maximum possible daily dose inside the hot lab.

![Figure 2.1 Position of RadEye detector on the Trasis machine](image)

The data were downloaded to a hard drive at the end of each 8 hour shift. Exposure rate plots were then exported to analysis software and analyzed with the handwritten logs. The times for each logged activity were highlighted. The exposure rate curve within each highlighted event would then be integrated to calculate the total exposure the individual received for that event according to the formulae below.

14
Equation 2.1 displays the information stored as data points by the device per measurement.

$$\dot{X} \left( \frac{\mu R}{hr} \right) = \frac{1}{K} \sum_{i=1}^{K} \dot{X}_i \left( \frac{\mu R}{hr} \right) \quad (2.1)$$

Here $i$ represents each second the device is measuring, and $K$ represents the sampling time of the instrument ($K$ is 25 seconds). $\dot{X} \left( \frac{\mu R}{hr} \right)$ is the average dose rate recorded for the 25 second sampling period. By choosing 25 seconds as the sampling interval allows the detectors to store measurements up till a total period of 11 hours.

The total exposure for a given interaction is calculated manually using equation 2.2.

$$X (mR) = \sum_{j=1}^{N} \left[ \dot{X} \left( \frac{\mu R}{hr} \right) \times \frac{1}{3600 \text{ sec}} \times \frac{1 \text{ mR}}{1000 \text{ } \mu R} \times K (sec) \right] \quad (2.2)$$

where $N$ is the number of measurements recorded during a given interaction, and $j$ represents each measurement for the 25 seconds interval.

Finally, the uncertainties of measured dose were determined by the relative errors at each dose rate characterized previously with the use of ion chamber [12].

**2.3 Results and Discussion**

**2.3.1 Average dose reduction by procedure**

Since Tech #4 was doing the dose draw and injection without using the Trasis dose draw/injection system, and Tech #2 and Tech #3 were doing dose draw and injection with the Trasis dose draw/injection system, we determined the dose reduction brought about by the automatic system by examining the doses received by these technologists for dose draw and dose injection tasks, separately.
As shown in Figure 2.1, the total dose received by Tech #4 from dose draw and injection tasks without using the Trasis system were about 0.25 mRem (0.0025 mSv), which was consistent with values recorded previously [12]. On average, the dose draw procedure without using the Trasis machine would incur the technologists 0.04 mRem (0.004 mSv) per procedure, and was 75% higher than the incurred dose of 0.01 mRem (0.0001 mSv) per procedure while using the Trasis machine. For the dose injection procedure, the dose reduction was about 70% between without using the Trasis machine (0.16 mRem or 0.0016 mSv) and using the Trasis machine (0.05 mRem or 0.0005 mSv). For the dose injection procedure, dose reduction was about 70% when comparing dose injection for a manual technique (with shielding) to using Trasis machine. The manual technique incurred the technologist a dose of 0.16 mRem (0.0016 mSv) while the with Trasis machine the technologist got a dose of 0.05 mRem (0.0005 mSv).

2.3.2 Qualitative dose distribution from each tasks

Figure 2.2 Average dose reduction by procedures
Previously, the dose draw/injection tasks gave the technologists highest dose among all the tasks [12]. The situation changed since the introduction of the Trasis system (Figure 2.2).

![Graph showing daily dose contributed from each task type]

**Figure 2.3 Mean daily dose contributed from each task type**

As shown in Figure 2.3, the most contributing factor to technologist overall dose was from the patient positioning task. With the Trasis machine, PET technologists typically received a dose of 0.7 mR (0.007 mSv) from dose draw/injection, 0.5 mR (0.005 mSv) from patient transport, 1.3 mR (0.013 mSv) from patient positioning, 0.1 mR (0.001 mSv) from patient care, and 0.3 mR (0.003 mSv) from non-specific per working day. When examining Figure 2.3, please keep in mind that Tech #1 (chief technologist) only handled dose draw/injection. As shown in Figure 2.3, there was non-trivial contribution from other or non-specific category. This could be due to unknown radiation received by the technologists, or technologists forgetting to log their activities. For example, these

17
activities could come from cleaning radioactive spills or dealing with patient uncontrolled urination during the procedure.

### 2.3.3 Daily dose reduction

Since the previous study quantified the daily doses to Tech #1, Tech #2, and Tech #3, we were able to determine the daily dose reduction for these three technologists (Figure 2.4). [12]

![Figure 2.4 Average daily doses before and after implementation of Trasis system](image)

**Figure 2.4 Average daily doses before and after implementation of Trasis system**

The green bars represent the data collected (2011) previously when dose were drawn up by hand with a hand shield, the blue bars represent the data collected with Trasis system implemented. The average daily dose in mRem was shown above each bar. On average, there was a 36% daily dose reduction for the three technologists together. It is important to notice that the daily doses involved with both technologists show
significant variations without using Trasis machine. This was not uncommon because the doses to technologists depended on their technical skills and experiences. In contrast, the doses to Tech #2 and Tech #3 were comparable, because of the involvement of automatic machine in dose draw and injection.

2.3.4 Comparison with radiation film badge data

In order to verify our RadEye measurements, film badge data from four other PET technologists were compared from September, 2011 to December, 2012 [13]. Note that the Trasis system was implemented on Feb, 2012. A discussion of film badge data was mentioned in section 1.6. The finger dose from badge was correlated with dose reduction by dose draw/injection procedure because fingers were directly exposed to such activities. The whole body dose was correlated with the overall daily dose reduction. The comparison of whole body dose was plotted in Figure 2.4 and the comparison of finger dose was plotted in Figure 2.5 while the raw data were included in Appendix A. According to the badge results, the reduction for whole body dose on average was 26%, while the reduction for finger dose was on average 64%. Given that the dose values from radiation badges could have error as large as 20% [26], thus, it was safe to conclude that the dose reduction measured by RadEye detectors were comparable to those determined from radiation badge results.
2.3.5 Additional doses brought upon by the Trasis machine

Figure 2.5 Whole body dose data collected by RadEye and film badge separately before and after using the Trasis system (Sept 2011 to Dec 2012).

Figure 2.6 Finger dose data collected by RadEye and film badge separately before and after using the Trasis system. (Sept 2011 to Dec 2012).
As mentioned in introduction, the Trasis machine is in a secure location with limited access where the raw F-18 and Tc-99m solution was stored. Since the activity of F-18 has a short half life (2 hours), the source needed replenished 2 to 3 times per day to meet clinical requirements. In this study, Technologist #5 was responsible for attaching F-18 source container to the Trasis machine and this could potentially increase the technologist’s exposure. Technologist #5 was responsible for Tc-99m drawing by hand as well. The dose distribution for Tech #5 was shown in Figure 2.7.

![Figure 2.7 Dose distributions from each task for Technologist #5](image)

As shown in Figure 2.7, the biggest dose contribution for Technologist #5 was from attaching F-18 source onto the Trasis machine. This was due to the high activity of F-18, and even though the time spent on attaching the source was less than 1 minute, the
exposure was significant at 2.3 mRem (0.023 mSv) per day. Thus, all other duties for Technologist #5 was insignificant as compared to attaching the source.

### 2.3.6 Comparison to annual dose limits for radiation workers

To put things into perspective, the average daily doses for the technologists were converted to annual doses in Table 2.3. In addition, the survey done with RadEye placed outside the source holding were recorded as well in Table 2.3. The purpose of the survey was to examine the maximum dose that could occur surrounding the Trasis machine inside the hot lab.

**Table 2.3 Estimated Annual Doses for technologists**

<table>
<thead>
<tr>
<th></th>
<th>Average Daily Dose (mRem)</th>
<th>Estimated Annual Dose (Rem)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tech 1</td>
<td>1.1</td>
<td>0.3</td>
</tr>
<tr>
<td>Tech 2</td>
<td>3.1</td>
<td>0.8</td>
</tr>
<tr>
<td>Tech 3</td>
<td>3.2</td>
<td>0.8</td>
</tr>
<tr>
<td>Tech 5</td>
<td>3.4</td>
<td>0.8</td>
</tr>
</tbody>
</table>

As shown in Table 2.3, Tech #1 was estimated to have an annual dose of 0.3 Rem or 3 mSv while Tech #2, Tech #3, Tech #5 estimated to have annual dose of 0.8 rem or 8 mSv. The estimated annual dose for the survey was 1.8 Rem or 18 mSv. The annual dose limit for radiation worker is 5 Rem (50 mSv). Therefore, all the radiation workers were well within the annual dose limit.
2.4 Conclusion

The results of the study have several important implications for PET clinics. First, the study proved that there was significant dose reduction by the automatic dose draw/injection system. Second, the most contributing task type for PET technologists’ doses shifted from dose draw/injection to patient positioning after the introduction of Trasis machine. This would suggest to technologists to change their practices in order to work quickly and efficiently around patients.
3. Project 2: Validation of ceiling shielding for CT/PET room with RADEYE

3.1 Introduction

The shielding of a CT/PET room is of great concern to health physicist because the 511 keV photons are much more penetrating than other radionuclides used in nuclear medicine. Thus, the CT/PET room must be adequately shielded in order to reduce doses to clinic workers and the general public. The recommended requirements for shielding CT/PET rooms are described by NCRP report 147 and AAPM task group 108 [14][15][16]. However, these recommendations were based on hypothesized scenarios that could be somewhat different than the cases in real life.

To be more specific, the AAPM TG108 report failed to take into account of the effect of scattered radiation within the room [14][17]. They used a parallel broad beam to simulate the F-18 emission from the patient. But in reality, the F-18 can accumulate in certain parts in the patient, and act as a point source. In addition, some factors that AAPM TG108 used in determining the thickness of ceiling shielding may be site specific. These factors include the number of patients imaged, the activity of radiotracer administered per patient, the length of time each patients remained in the facility, and loss of F-18 activity due to patient voiding [14].

Due to these complications, we proposed to determine the doses in the room above CT/PET suite based on the detector readings placed on the ceiling. In this way, both primary and scattered radiation inside the room can be taken into account. Furthermore, the RadEye detectors described in Section 1.7, could measure doses
continuously throughout the day, and it provided two immediate benefits. First, it could distinguish the doses from CT and PET separately and second, the doses it measured was real doses in a clinical setting and these were not based on estimated values.

In short, the transmitted dose behind the ceiling could be calculated based on the doses immediately before the shielding, without worrying about other factors, such as scattered radiation, amount of activity injected, etc. In this study, we measured the doses inside the CT/PET room for three consecutive days, in order to determine the radiation doses inside the room and evaluate the effectiveness of the current shielding. If the calculated dose behind the shielding was significantly lower than the regulated limit, this meant that we could potentially propose an alternate shielding method. Considering the construction of a CT/PET suite could be expensive, this study might bring a new concept in CT/PET shielding.

The shielding requirement for a CT/PET combined modality was the issue of interest in this article. One would assume that any shielding required for PET would be sufficient for CT. However, the amount of doses given by CT modality could be significantly larger than the doses given by PET modality. Thus, we would want to segregate the doses given by CT and PET modalities separately before doing the shielding calculations. For a CT/PET modality, the CT scanner was typically in the front of the gantry with the PET scanner located in the back of the gantry about half a meter away [2]. PET scans could be acquired through either 2D or 3D modes. The 2D mode had collimators while the 3D mode did not have collimators. There was additional shielding
provided by 2D mode and this was another reason to use the real doses in clinic for shielding calculations.

The regulatory limit for non-occupational doses was set by the Nuclear Regulatory Commission (NRC). The federal code of regulation (10 CFR20) established the dose limits in controlled radiation areas and uncontrolled areas open to general public [7]. Under such regulations, the facility must be shielded so that the effective dose equivalent in uncontrolled areas did not exceed 1 mSv/year or 20 µSv in any 1 hour [7]. The 1 mSv/year limit implied a weekly dose limit of 0.02 mSv, which is equivalent to 2 mR.

3.2 Materials and Methods

RadEye detectors were placed at various positions (Figure 3.1) inside a CT/PET room at Duke Cancer Center for three business days. The exact blue print of the room was listed in Appendix B. As mentioned previously in section 1.7, RadEye detectors could store measured doses every 25 seconds during the eight hour working period. The radiation doses were then analyzed at the end of each day.
Figure 3.1 Geometry indicating the location of detectors and the PET/CT suite layout

A spectrum of daily doses could be plotted for the detectors in the room. A typical spectrum was shown in Figure 3.2 during a scanning interval. It contained a plateau region with a distinctive spike region. Based on the results retrieved from PET and CT scans, the spike region was the result of short, burst of photons coming from the CT scans while the plateau region was the result of PET scans. In Figure 3.2, the CT component was colored green and the PET component was colored blue. The areas under the green curve were summed to represent the dose contributions from CT scan, and the areas under the blue curve were summed to represent dose contributions from PET scan.
3.2.1 Analytical model

Once we determined the measured doses separately from CT and PET contributions, we could determine analytically the doses after the shielding using an analytical model. For the analytical calculations, we needed to take into account 1) inverse square law 2) attenuation through the barrier 3) angle of incidence and 4) build-up factors due to broad-beam geometry. Hence, Equation 1 was used to compute the dose contribution from CT radiation after passing through the shielding material, which was consisted of 3 inch of steel and 3.25 inch of concrete.

\[
I_{B,CT} = \frac{I_{A,CT}}{d^2} \times e^{-\mu t} \cos(\theta) \times \epsilon
\]  

(3.1)

Where:

\( I_{B,CT} \) = the shielded dose for CT

\( I_{A,CT} \) = the initial dose for CT
\( b_{1, \text{steel}} = \) dose build up factor due to steel for CT

\( b_{2, \text{concrete}} = \) dose build up factor due to concrete for CT

\( \mu_{1, \text{steel}} = \) attenuation coefficient of steel for CT

\( \mu_{2, \text{steel}} = \) attenuation coefficient of concrete for CT

Similarly, the shielded dose of PET was calculated according to Equation 2.

\[
I_{B, PET} = I_{A, PET} \times b_{1, \text{steel}} \times e^{-\mu_{1, \text{steel}} \times x_1} \times b_{2, \text{concrete}} \times e^{-\mu_{2, \text{concrete}} \times x_2}
\] (3.2)

Where:

\( I_{B, PET} = \) the shielded dose for PET

\( I_{A, PET} = \) the initial dose for PET

\( b_{1, \text{steel}} = \) dose build up factor due to steel for PET

\( b_{2, \text{concrete}} = \) dose build up factor due to concrete for PET

\( \mu_{1, \text{steel}} = \) attenuation coefficient of steel for PET

\( \mu_{2, \text{steel}} = \) attenuation coefficient of concrete for PET

The build-up factor was calculated using Berger’s formula [18] due to the broad beam geometry of radiation (Equation 3.3 and 3.4). The Berger’s formula was an empirical formula based on physical measurements [18].

\[
B(x) = 1 + axe^{bx}
\] (3.3)

\[
x = \mu r
\] (3.4)

The appropriate a and b values used in the Berger’s formulas were tabulated in Table 3. These values were found according to the Table in Appendix C based on the energies of photons. For the PET case, only photons with energy of 511 keV were considered. For
the CT case, an effective energy of 60 keV was used, which was approximated from a 120 kVp scan.

Table 3.1 Constants used in the Berger’s formulas

<table>
<thead>
<tr>
<th></th>
<th>PET</th>
<th></th>
<th>CT</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>a</td>
<td>b</td>
<td>a</td>
<td>b</td>
</tr>
<tr>
<td>steel</td>
<td>1.16</td>
<td>0.036</td>
<td>0.07</td>
<td>-0.039</td>
</tr>
<tr>
<td>concrete</td>
<td>1.73</td>
<td>0.055</td>
<td>0.78</td>
<td>-0.008</td>
</tr>
</tbody>
</table>

Thus, according to Table 3.1 and Equation 3.3 and 3.4, the build-up factors used in the shielding calculations were determined and entered in Table 3.2. The build-up factors were checked against the build-up factors obtained by Monte Carlo simulation in the AAPM TG108 report [14]. The comparison was listed in Table 3.3.

Table 3.2 Build-up factors calculated by Berger’s formulas

<table>
<thead>
<tr>
<th>Build-up factors</th>
<th>PET</th>
<th>CT</th>
</tr>
</thead>
<tbody>
<tr>
<td>steel(PET)</td>
<td>17.82</td>
<td></td>
</tr>
<tr>
<td>steel(CT)</td>
<td>1.04</td>
<td></td>
</tr>
<tr>
<td>concrete(PET)</td>
<td>12.36</td>
<td></td>
</tr>
<tr>
<td>concrete(CT)</td>
<td>11.20</td>
<td></td>
</tr>
</tbody>
</table>

Table 3.3 Broad beam transmission factors at 511 keV comparing to AAPM TG108 [19]

<table>
<thead>
<tr>
<th></th>
<th>Berger</th>
<th>AAPM TG108</th>
</tr>
</thead>
<tbody>
<tr>
<td>Steel (PET)</td>
<td>0.000666</td>
<td>0.000666</td>
</tr>
<tr>
<td>Concrete (PET)(^a)</td>
<td>0.071812</td>
<td>0.155000</td>
</tr>
</tbody>
</table>

\(^a\)The density of normal concrete (2.35 g/cm\(^3\)) was used in AAPM TG108, while a high density concrete (3.35 g/cm\(^3\)) was used in the calculation.
### 3.3 Results and Discussion

#### 3.3.1 Radiation dose levels within the PET/CT scanning room

The results for average daily doses were tabulated in Table 3.4 and Figure 3.3. As mentioned in 3.2.1, the daily doses were divided into PET and CT components.

<table>
<thead>
<tr>
<th></th>
<th>Det#1</th>
<th>Det#2</th>
<th>Det#3</th>
<th>Det#4</th>
<th>Det#5</th>
<th>Det#6</th>
</tr>
</thead>
<tbody>
<tr>
<td>Total (mR)</td>
<td>8.66</td>
<td>5.89</td>
<td>1.67</td>
<td>10.29</td>
<td>4.31</td>
<td>6.26</td>
</tr>
<tr>
<td>PET (mR)</td>
<td>0.85</td>
<td>0.71</td>
<td>1.20</td>
<td>1.03</td>
<td>0.69</td>
<td>1.12</td>
</tr>
<tr>
<td>CT (mR)</td>
<td>7.72</td>
<td>5.30</td>
<td>0.37</td>
<td>10.21</td>
<td>3.63</td>
<td>5.17</td>
</tr>
<tr>
<td>Background (mR)</td>
<td>0.09</td>
<td>0.14</td>
<td>0.10</td>
<td>0.09</td>
<td>0.13</td>
<td>0.08</td>
</tr>
</tbody>
</table>

*The location of detectors were shown in Figure 3.1. Detectors 2 to 5 were placed on the ceiling while detectors 1 &6 were placed at 1 meter above the ground.*

![Figure 3.3 Average daily doses recorded by the six detectors within the room](image)

As shown in Table 3.4 and Figure 3.3, within the scanning room, the dose contributions from CT were on average significantly larger than dose contributions from PET. The reason why detector 3 had such low contributions from CT was that the gantry...
of machine shielded most of the radiation. Depending on the locations of detectors, the doses recorded would be different for both CT doses and PET doses. For the four detectors on the ceiling (detector #2 to #5), the highest dose recorded was 11.10 mR per day for CT radiation while the highest dose recorded for PET radiation was 1.36 mR per day. In addition, the doses received by the six detectors normalized per scan were plotted in Figure 3.4 and Figure 3.5.

![Graph showing the dose per scan for different detector locations and days.](image)

**Figure 3.4** Dose received by the six detectors within the room from PET radiation, normalized to each scanning procedure.
As shown in Figure 3.4 and Figure 3.5, the doses received by each detector per scanning procedure were consistent from day to day. The slight difference would be due to different amount of radionuclide used during each day.

### 3.3.2 Transmitted doses after ceiling shielding

The calculation of transmitted doses after ceiling shielding was estimated based on the four detectors at the ceiling. The calculations were described in the previous section. The data were presented in Table 3.5, with the first row showing dose estimated by PET contributions, and the second row showing dose estimated by CT contributions. It could be seen that the radiation doses as a result of CT scan after the...
shielding was negligible. If we were to measure the doses from CT scans after the shielding, there should be no radiation at all except for background radiation.

The weekly doses were estimated based on the average daily doses for the three consecutive days and were tabulated in the fourth row. Furthermore, assuming a person was sitting at the room above in a chair at a height of 0.5 meter, the radiation doses to the person was calculated with an assumed occupancy factor of 1 (row 5 & 6). As shown in Table 3.5, the calculated weekly doses were about $2.2 \times 10^{-4}$ mR per week, about 10000 times less than the regulated limit of 2 mR per week.

Table 3.5 Calculated daily radiation doses after the ceiling shielding

<table>
<thead>
<tr>
<th></th>
<th>Det #2 (mR)</th>
<th>Det #3 (mR)</th>
<th>Det #4 (mR)</th>
<th>Det #5 (mR)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Dose PET</strong>&lt;sup&gt;a&lt;/sup&gt;</td>
<td>4.2E-05</td>
<td>7.2E-05</td>
<td>6.1E-05</td>
<td>4.1E-05</td>
</tr>
<tr>
<td><strong>Dose CT</strong>&lt;sup&gt;a&lt;/sup&gt;</td>
<td>5.2E-69</td>
<td>6.4E-70</td>
<td>8.9E-69</td>
<td>3.7E-69</td>
</tr>
<tr>
<td><strong>Dose per day</strong></td>
<td>4.2E-05</td>
<td>7.2E-05</td>
<td>6.1E-05</td>
<td>4.1E-05</td>
</tr>
<tr>
<td><strong>Dose per week</strong></td>
<td>2.1E-04</td>
<td>3.6E-04</td>
<td>3.1E-04</td>
<td>2.1E-04</td>
</tr>
<tr>
<td><strong>At 0.5 meter high</strong></td>
<td>1.5E-04</td>
<td>2.6E-04</td>
<td>2.2E-04</td>
<td>1.5E-04</td>
</tr>
<tr>
<td><strong>with Occupancy</strong>&lt;sup&gt;b&lt;/sup&gt;</td>
<td>1.5E-04</td>
<td>2.6E-04</td>
<td>2.2E-04</td>
<td>1.5E-04</td>
</tr>
</tbody>
</table>

<sup>a</sup> Doses are calculated separately for PET and CT contributions separately

<sup>b</sup> Occupancy factor is assumed to be 1 as recommended by NCRP151

3.3.3 Comparison with AAPM TG108
Dose calculation was performed according to AAPM TG108’s formalism to check with our measured data and calculation. Parameters based on AAPM TG108’s formalism were tabulated in Table 3.6, and the values of these parameters were based on real clinical data at Duke University Medical Center.

Table 3.6 Summary of dose parameters

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Definition</th>
<th>Formulation</th>
<th>Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>( A_0 )</td>
<td>Administered activity (mCi)</td>
<td></td>
<td>12</td>
</tr>
<tr>
<td>( T_U )</td>
<td>Uptake time (hr)</td>
<td></td>
<td>0.5</td>
</tr>
<tr>
<td>( T_I )</td>
<td>Imaging time (hr)</td>
<td></td>
<td>0.5</td>
</tr>
<tr>
<td>( D(0) )</td>
<td>Initial dose rate (( \mu )Sv-m(^2)/MBq-hr)</td>
<td></td>
<td>0.092</td>
</tr>
<tr>
<td>( T_{1/2} )</td>
<td>Radionuclide half life (min)</td>
<td></td>
<td>110</td>
</tr>
<tr>
<td>( F_U )</td>
<td>Uptake time decay factor</td>
<td>( \exp\left[-0.693T_U/T_{1/2}\right] )</td>
<td>0.83</td>
</tr>
<tr>
<td>( R_{ti} )</td>
<td>Dose reduction factor over imaging time</td>
<td>( 1.443(T_{1/2}/t_i)[1-\exp(-0.693t_i/T_{1/2})] )</td>
<td></td>
</tr>
<tr>
<td>( d )</td>
<td>Distance from source to detector (m)</td>
<td></td>
<td>various</td>
</tr>
</tbody>
</table>

Based on Table 3.6, and the formalism by the AAPM TG108, the calculated radiation doses from PET scan per patient were calculated and tabulated in Table 3.7.

Table 3.7 Comparison of calculated PET doses and measured PET doses inside the room

<table>
<thead>
<tr>
<th>Dose (mR)</th>
<th>( \text{Det#1} )</th>
<th>( \text{Det#2} )</th>
<th>( \text{Det#3} )</th>
<th>( \text{Det#4} )</th>
<th>( \text{Det#5} )</th>
<th>( \text{Det#6} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>measured PET</td>
<td>0.85</td>
<td>0.71</td>
<td>1.20</td>
<td>1.03</td>
<td>0.69</td>
<td>1.12</td>
</tr>
<tr>
<td>calculated PET</td>
<td>1.10</td>
<td>0.72</td>
<td>0.90</td>
<td>0.90</td>
<td>0.20</td>
<td>1.10</td>
</tr>
</tbody>
</table>
As shown in Table 3.7, large variations were shown for Detector #1, #3, and #5. These variations could be due to two factors. First, in the calculation, the assumption was that the patient was immobilized and stayed at the center of the gantry during the scanning. However, in reality, the patient would spend quite a lot of time moving inside the scanning room, either by himself or on the table. Thus, the reading by detectors should show a difference, especially for detector located far away from the gantry. Second, the shielding from gantry was ignored in the calculation. In fact, the shielding by CT/PET gantry could range from 15% to about 600% [14][31]. Therefore, the important message here was that the measured radiation doses inside the room due to PET scan was reasonable and the value was about 1 mR per day, according to Table 3.7.

Next, we would like to compare the doses after shielding based on AAPM TG108 calculation and compared with our hand calculations. The shielding between the ceiling and floor upstairs were 3 inch of steel and 3.25 inch of concrete. Yet the radiation path length depended strongly on the incident angle. So we performed the calculation for two extreme cases, one directly above the gantry head and the other at an incident angle of 15 degree. The highest and lowest doses measured were used in the calculating doses behind shielding. The calculation was based on the transmission factor provided by AAPM TG108.

<table>
<thead>
<tr>
<th>% difference</th>
<th>28.94</th>
<th>1.43</th>
<th>25.14</th>
<th>12.42</th>
<th>71.03</th>
<th>1.90</th>
</tr>
</thead>
</table>

Table 3.8 Doses calculated with 90 incident angle

<table>
<thead>
<tr>
<th>Dose per day (mR)</th>
<th>Dose after shielding per day (mR)</th>
</tr>
</thead>
</table>

36
<table>
<thead>
<tr>
<th>Dose per day (mR)</th>
<th>Dose after shielding per day (mR)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.20</td>
<td>2.6×10^{-2}</td>
</tr>
<tr>
<td>0.69</td>
<td>1.5×10^{-2}</td>
</tr>
</tbody>
</table>

Table 3.9 Doses calculated with 15 degree incident angle

<table>
<thead>
<tr>
<th>Dose per day (mR)</th>
<th>Dose after shielding per day (mR)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.20</td>
<td>1.1×10^{-9}</td>
</tr>
<tr>
<td>0.69</td>
<td>6.0×10^{-10}</td>
</tr>
</tbody>
</table>

Our calculated dose (Table 3.7) based on an incident angle of 30 degree was between $4×10^{-5}$ and $7×10^{-5}$ mR per day. This agreed with the values in Table 3.8 and Table 3.9.

In summary, we showed that the doses before the shielding inside the CT/PET room calculated with AAPM TG108 formalism was in agreement with our measured data. However, the recommended calculation methodology by AAPM TG108 may contain inaccurate assumption when performing dose calculation behind a shielding barrier. In their methodology, they calculated the shielding thickness reversely based on the transmission factors at a distance. Thus, their shielding thicknesses were all perpendicular to the incident radiation (shown by orange circle, in Figure 3.6). In reality, the shielding should not be perpendicular to the incident radiation (red rectangle, in Figure 3.6). Due to this difference, the actual thickness of the shielding should be less than the thickness by AAPM formalism. Indeed, as shown in Table 3.8 and 3.9, the
difference in dose behind shielding could be orders of magnitude different due to different incident angles, by AAPM’s own simulated transmission factors.

![Figure 3.6 Illustration of difference of thickness calculated by AAPM and in reality](image)

**3.4 Conclusions**

This study quantified the CT and PET doses contributions separately in the clinical CT/PET room. An analytical formula was developed to calculate the non-occupational personnel dose above the CT/PET room and the calculated results were confirmed by physical measurements. The actual physical dose was much smaller than the NCRP design goals of 0.02 mSv/week.
4. Project 3: Evaluating MOSFET calibration factor dependency on effective energy over diagnostic energy range

4.1 Introduction

Recently, the application of metal-oxide-semiconductor field effect transistor (MOSFET) detectors for determination of radiation dose in diagnostic range is gaining popularity. Compared to other detectors, MOSFETs have unique advantages because they are small (3 mm diameter, 25 mm length), wireless and provide immediate dose readings [20]. The basic operational principle of MOSFET was already discussed in section 1.4.

The dosimetric characteristics of commercially available MOSFET detectors for \textit{in vivo} dosimetry in the kV x-ray range with respect to fading, temperature dependence and directional dependence were well studied [21,22]. Additionally, it was shown that MOSFET sensitivity is energy dependent [5]. As the incident energy decreases from 150 keV to 10 keV, MOSFET sensitivity increases in device [5]. This is due to increased photoelectric cross section at lower energies, resulting additional release of electron-hole pairs and a greater threshold voltage shift [5].

To characterize the energy dependency of MOSFET dosimeters, both peak voltage (kVp) and total filtration needed to be considered. Clinically, before using MOSFET for dosimetric measurement, we would calibrate it under the real measurement conditions for a given kVp and total filtration. The process is time consuming and it reduces MOSFET lifetime. Therefore, the purpose of this study was to quantify the energy dependence of MOSFET sensitivity over diagnostic range. If there was a clear
dependency, we could estimate the calibration factors based on given effective energy and the in-house generated MOSFET aging data [22]. Furthermore, this study would answer the question whether or not MOSFET can be used to measure doses under fluoroscopy when there was a change in kVp and filtration on the fly. Finally, an analysis was given to show the degree of uncertainty for dose measurement without doing dose calibration.

4.2 Materials and Methods

Five new MOSFETs (SN: 21406, 21407, 21408, 21409, 21410, Best Medical) were used in the study. The calibration factors were measured in two ways: 1) fixed kVp, fixed FOV, and varying filtration; and 2) fixed filtration, fixed FOV, and varying kVp. Effective energy was computed as a function of kVp and filtration by SpekCalc [23].

The MOSFET detectors were calibrated with an ion chamber (model 10x5-6, Radcal, Monronvia, CA) to determine MOSFET calibration factors. The exposure measurement from ion chamber, X, was converted to absorbed dose in air, D, according to the protocol from the American Association of Physicist in Medicine (AAPM) radiation therapy task group 61 (TG61) (equation 4.1) [27].

\[
D(Q)_{air} = 0.876 \times X \times N_x^{120kVp} \times P_{stem}^{air}
\]  

(4.1)

where 0.876 cGy R \(^{-1}\) was the exposure to dose in air coefficient, and X (R) was the ionization in air reading that had been temperature and pressure corrected, \(N_x\) was the ion chamber calibration coefficient obtained from the University of Wisconsin Accredited Dosimetry Calibration Laboratory, and the chamber stem correction factor
\( P_{stem,air} \) was assumed to be unity since the calibration setup met AAPM TG-61 defined criteria (i.e., collimated field size differed <50% when compared with the reference calibration field size) [27]. As a result, the MOSFET calibration coefficient (mV cGy\(^{-1}\)) was calculated using following equation:

\[
CF = \frac{\text{MOSFET reading}}{D(Q)_{air}}
\]  

(4.2)

where the MOSFET reading was the measured voltage shift, and \( D(Q)_{air} \) was the calculated absorbed dose in air.

The MOSFET was first calibrated in-air using conventional radiographic x-ray tube (Philips Diagnostic). The tube potential was kept at 120 kVp, and the filtration was varied (Figure 4.1). The filtration values were determined by Piranha detector (RTI electronics, Towaco, NJ). Caution was taken to warm up x ray tube and actual measurements were not performed until consistent filtration and kVp values were recorded. In this study, the following filtration values were measured (from smallest to largest): 5.13 mm Aluminum, 6.94 mm Aluminum, 7.98 mm Aluminum, 8.68 mm Aluminum. As shown in Figure 4.1 (a), MOSFETs were secured to a foam pad with their sensitive region facing main axis of the x-ray beam. In addition, the center of MOSFET detectors was aligned with center of ion chamber as shown in Figure 4.1 (b), where for cylindrical chamber this was the effective point of measurement [27]. The 6 cc ion chamber was used because previous experiments indicated that there were no dose differences between MOSFET measurements and 6 cc ion chamber for at least 5 cm away from the edge of the ion chamber [28]. The calibration factor for the ion chamber
at 120 kVp was 1.022 R/Rdg (University of Wisconsin – ADCL, Cal. Date: 04 Aug 2011).

The source to ion chamber distance was set at 61 cm and the field size was set to 4 cm by 2 cm. Same measurements were also performed at a different field size (7 cm by 7 cm).

Figure 4.1 Experiment set-up using the Phillips Diagnostic x-ray tube; (a) shows the horizontal radiation towards detectors; (b) shows the location of MOSFETs compared to ion chamber

Next, MOSFETs were calibrated with ion chamber at a given filtration but with varying kVp. Ideally, this study should be performed with the same x-ray tube. Unfortunately, it was not possible to do so due to x-ray tube over-heating. Instead, the experiment was carried by the x-rad 320 irradiator (Precision x-ray, Connecticut). The experimental set-up was shown in Figure 4.2. The source to ion chamber distance was set at 50 cm and the field size was 20 cm by 20 cm. The filtration was set at a constant value of 6.8 mm Aluminum while the peak voltage was varied at: 140 kVp, 120 kVp, and 80 kVp. Note however in this case a smaller ion chamber (model 10x5-0.18, Radcal, Monrovia, CA) was used because it was easier to pass through the opening from
behind. Ideally, three different ion chamber calibration factors should be used to calculate the absorbed dose based on Equation 4.1. However, only the calibration factor at 120 kVp was provided by the University of Wisconsin ADCL and thus the same ion chamber calibration factors were applied to all three kilovoltage values.

![Experimental setup](image)

**Figure 4.2** Experimental set-up using x-rad 320 irradiator; foam pads were used to set the detector-to-source distance at 50 cm

To determine the effective energy, the combined effect of peak voltage (kVp) and total filtration were converted into a single quantity, effective energy, through the SpekCalc software [23]. SpekCalc was a software package designed for x-ray spectrum calculation based on tungsten anode. In short, SpekCalc relied on both deterministic data and Monte Carlo generated data to obtain x-ray spectrum. In addition, SpekCalc results were checked in agreement with results generated by BeamNRC and IPEM78,
which were considered gold standard in generating x-ray spectrum. A sample spectrum obtained with SpekCalc was shown in Figure 4.3. As shown in Figure 4.3, the user could input parameters such as peak energy, angle of incidence, and filtrations by several materials and the SpekCalc would generate spectrum along with effective energy.

![A sample spectrum calculated by SpekCalc software](image)

**Figure 4.3** A sample spectrum calculated by SpekCalc software

### 4.3 Results and Discussions

#### 4.3.1 Calibration factors versus total filtration

The respective calibration factors were plotted against half value layer values in Figure 4.4. All the measurements were taken at 120 kVp and a fixed distance while the filtration values were adjusted.
As shown in Figure 4.4, there was no difference for calibration factors from HVL = 5 mm Al to HVL = 9 mm Al with uncertainties included.

### 4.3.2 Calibration factors versus peak voltage (kVp)

The calibration factors versus peak voltage were obtained under a given filtration (HVL = 6.6 mm Aluminum). Four data sets were maintained and were plotted in Figure 4.5. As shown in Figure 4.5, there was a linear trend between these two quantities.
4.3.3 Calibration factors versus effective energies

The effective energies calculated by SpekCalc were tabulated in Table 4.1 using a combination of filtration and kVp. Percent difference for a given calibration factor was automatically produced for the five MOSFET detectors under a given condition based on Equation 4.3.

\[
\%D = \frac{CF_{\text{max}} - CF_{\text{min}}}{CF_{\text{mean}}} \times 100
\]  

In Equation 4.3, \(CF_{\text{max}}\) was the maximum calibration factor recorded, and \(CF_{\text{min}}\) was the minimum calibration factor recorded. \(CF_{\text{mean}}\) was the average of the recorded calibration factors under a given kVp and filtration.
In Table 4.1, one data point (6.6 mm Aluminum, 74 kVp) was measured by a student for the Phillips Xper CT under fluoro mode with low filtration set-up to expand the lower energy range of this curve. A brand new MOSFET was used for this measurement.

**Table 4.1** HVL and kVp used in the study with the measured CF

<table>
<thead>
<tr>
<th>HVL (mm Al)</th>
<th>kVp</th>
<th>( E_{\text{eff}} ) (keV)</th>
<th>CF (mV/cGy)</th>
<th>% D</th>
</tr>
</thead>
<tbody>
<tr>
<td>6.6</td>
<td>74(^a)</td>
<td>45.8</td>
<td>36.1</td>
<td>14.4%</td>
</tr>
<tr>
<td>6.8</td>
<td>80</td>
<td>48.2</td>
<td>33.8</td>
<td>10.3%</td>
</tr>
<tr>
<td>6.8</td>
<td>120</td>
<td>59.8</td>
<td>28.8</td>
<td>3.3%</td>
</tr>
<tr>
<td>5.13</td>
<td>120</td>
<td>57.5</td>
<td>29.3</td>
<td>9.2%</td>
</tr>
<tr>
<td>6.94</td>
<td>120</td>
<td>59.9</td>
<td>29.6</td>
<td>7.6%</td>
</tr>
<tr>
<td>8.68</td>
<td>120</td>
<td>61.8</td>
<td>27.8</td>
<td>7.4%</td>
</tr>
<tr>
<td>6.8</td>
<td>140</td>
<td>64.3</td>
<td>27.8</td>
<td>6.6%</td>
</tr>
<tr>
<td>7.24</td>
<td>120</td>
<td>60.3</td>
<td>29.06</td>
<td>3.7%</td>
</tr>
</tbody>
</table>

\(^a\) Measurement taken by Chu Wang for the Xper CT under fluoro mode

The data in Table 4.1 were plotted in Figure 4.6 and fitted with a second degree polynomial. In general, at lower effective energy, the calibration factor was higher as expected. This was consistent with what we expected because at lower energy the photoelectric effect would take place with a higher cross section.
In addition, as shown in Figure 4.6, uncertainties for calibration factors also depended on effective energies. From 65 keV to 57 keV, the associated uncertainties with MOSFET measurements were from 3% to 9% while below 48 keV, the associated uncertainties with MOSFET were from 10% to 14%.

4.3.4 Sources of errors

At this stage, an analysis would analyze data uncertainties induced by different calibration methodologies. This would estimate the overall uncertainties for estimating MOSFET calibration factors without doing actual calibration.

As shown above, the in-air calibration performed by conventional x-ray tube was shot with no materials at the back of the detector while the in-air calibration done by the irradiator was shot with stainless steel 50 cm lower than the detectors. Therefore, hand calculations were performed to calculate the shift in effective energy for backscattered photons and to calculate possible dose enhancement due to back-
scattering. In addition, literature references were provided for dose enhancement for similar materials as shown below.

The effective energies from the study for different peak potentials and filtrations were listed in Table 4.1. The lowest effective energy was 45.8 keV while the highest effective energy was 60.3 keV. Based on Compton scattering formula (Equation 4.3), the backscattered photon energies were calculated for two angels (180 degree and 90 degree).

\[ E' = \frac{E}{1 + \frac{E}{m c^2} (1 - \cos \theta)} \]  

In Equation 4.3, \( E' \) was the energy of the scattered photon, and \( E \) was the energy of the incident photon. \( \theta \) was the scattering angle, and \( m c^2 \) was the electron rest energy.

The 180 degree backscatter represented the highest energy for the scattered photon while the 90 degree backscatter was a conservative estimation for the scattered photon. The calculated backscattered photon energies were listed in Table 4.2.

<table>
<thead>
<tr>
<th>Incident ( E_{\text{eff}} )</th>
<th>( E' ) (180 degree)</th>
<th>( E' ) (90 degree)</th>
</tr>
</thead>
<tbody>
<tr>
<td>45.8</td>
<td>38.8</td>
<td>42.0</td>
</tr>
<tr>
<td>60.3</td>
<td>48.8</td>
<td>53.9</td>
</tr>
</tbody>
</table>

As shown in Table 4.2, the change in effective energy for backscattered photons was significant and this could lead to an increase in sensitivity as large as 12%.
However, the measured calibration factors by the detector would depend on the amount of primary radiation and scattered radiation. Although there was a shift in effective energy between the primary photon energy and scattered photon energy, the amount of primary photon could outweigh the amount of scattered photon significantly. The effect of change in effective energy would be negligible for the purpose of calibration factor computation. The amount of scattered radiation reaching the detector would not be easy to calculate because it depended on various parameters, including beam energy, field size, and the thickness, width, position, and atomic number of the material. Therefore, only literature references were provided to estimate the calibration factor change as a result of scattered radiation.

Das et al. had quantified the backscatter dose perturbation in kilo-voltage photon beams at high atomic number interfaces. His experimental set-up was shown in Figure D1, Appendix D and a quantity called backscatter dose factor (BSDF) was defined to take into account the dose enhancement in the presence of high-Z medium. The BSDF was a function of beam energy, field size, geometry, and material of the scattering medium. His finding indicated that the BSDF depended on incident energy, the distance between the high-Z medium and the detector, and the material of the high-Z medium. It found the BSDF approached 1 as the distance between the detector and the lead was greater than 10 cm for a 10 cm by 10 cm field size (Figure D2, Appendix D). Since steel had a lower Z number (26) than lead (82), it was expected the BSDF at greater than 10 cm should also be close to 1. In addition, he characterized dose enhancement as a result of backscatter based on the field size. In short, BSDF factor was greater at smaller field
size than at larger field size (Figure D3, Appendix D). The exact difference as a result of field size varied based on difference beam energies. If the high Z material was lead, based on Das et al data, the dose difference between a field size of 7 by 7 cm and a field size of 20 by 20 cm could be as large as 6%. In summary, simulated Monte Carlo data would be needed eventually to quantify the effect of backscattered radiation.

Therefore, for a conservative estimate, where scattering did happen and the amount of scattered radiation compared to primary radiation was significant, overall uncertainty would comprise of uncertainties from inherent statistical radiation variations (3% to 15%), and uncertainties as a result of presence of scattered radiations (0% to 18%). Hence, the combined overall uncertainty could be in between 3% to 33%.

4.4 Conclusions

A clear trend of calibration factor versus effective energy was observed over diagnostic range. Hence, we created a calibration curve so that under a given fixed kVp or filtration, the calibration factor is automatically generated. The uncertainties for estimating MOSFET calibration factors were estimated based on statistical nature of radiation and geometrical variation between standard in-air calibration and actual experimental set-up. The findings are significant because 1) this proved a means to estimate MOSFET calibration factors under emergency condition 2) enabled MOSFET study for dynamic dose delivery when there would be a kVp change on the fly, such as fluoro machine in the interventional department.
Appendix A Badge doses recorded for the PET technologists

Table A1. Badge results (whole body) for four random technologists in PET facility from Sept, 2011 to Dec, 2012

<table>
<thead>
<tr>
<th></th>
<th>DDE</th>
<th>Tech 1</th>
<th>Tech 2</th>
<th>Tech 3</th>
<th>Tech 4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sep-11</td>
<td>N/A</td>
<td>192</td>
<td>133</td>
<td>165</td>
<td></td>
</tr>
<tr>
<td>Oct-11</td>
<td>N/A</td>
<td>89</td>
<td>116</td>
<td>216</td>
<td></td>
</tr>
<tr>
<td>Nov-11</td>
<td>147</td>
<td>119</td>
<td>116</td>
<td>248</td>
<td></td>
</tr>
<tr>
<td>Dec-11</td>
<td>56</td>
<td>129</td>
<td>71</td>
<td>125</td>
<td></td>
</tr>
<tr>
<td>Jan-12</td>
<td>93</td>
<td>119</td>
<td>84</td>
<td>126</td>
<td></td>
</tr>
<tr>
<td>Feb-12</td>
<td>99</td>
<td>137</td>
<td>52</td>
<td>161</td>
<td></td>
</tr>
<tr>
<td>Mar-12</td>
<td>87</td>
<td>132</td>
<td>64</td>
<td>90</td>
<td></td>
</tr>
<tr>
<td>Apr-12</td>
<td>103</td>
<td>96</td>
<td>62</td>
<td>122</td>
<td></td>
</tr>
<tr>
<td>May-12</td>
<td>99</td>
<td>95</td>
<td>69</td>
<td>88</td>
<td></td>
</tr>
<tr>
<td>Jun-12</td>
<td>55</td>
<td>126</td>
<td>31</td>
<td>84</td>
<td></td>
</tr>
<tr>
<td>Jul-12</td>
<td>62</td>
<td>82</td>
<td>90</td>
<td>90</td>
<td></td>
</tr>
<tr>
<td>Aug-12</td>
<td>90</td>
<td>N/A</td>
<td>81</td>
<td>131</td>
<td></td>
</tr>
<tr>
<td>Sep-12</td>
<td>63</td>
<td>N/A</td>
<td>86</td>
<td>131</td>
<td></td>
</tr>
<tr>
<td>Oct-12</td>
<td>79</td>
<td>N/A</td>
<td>108</td>
<td>128</td>
<td></td>
</tr>
<tr>
<td>Nov-12</td>
<td>76</td>
<td>N/A</td>
<td>53</td>
<td>115</td>
<td></td>
</tr>
<tr>
<td>Dec-12</td>
<td>N/A</td>
<td>N/A</td>
<td>54</td>
<td>82</td>
<td></td>
</tr>
</tbody>
</table>

* N/A means that either the technologist left the job or the technologist had not started working at PET facility
Table A2. Badge results (finger) for four technologists in PET facility from Sept, 2011 to Dec, 2012

<table>
<thead>
<tr>
<th>SDE</th>
<th>Tech 1</th>
<th>Tech 2</th>
<th>Tech 3</th>
<th>Tech 4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sep-11</td>
<td>N/A</td>
<td>1126</td>
<td>699</td>
<td>1074</td>
</tr>
<tr>
<td>Oct-11</td>
<td>N/A</td>
<td>956</td>
<td>402</td>
<td>1645</td>
</tr>
<tr>
<td>Nov-11</td>
<td>133</td>
<td>755</td>
<td>517</td>
<td>1043</td>
</tr>
<tr>
<td>Dec-11</td>
<td>88</td>
<td>906</td>
<td>462</td>
<td>1056</td>
</tr>
<tr>
<td>Jan-12</td>
<td>69</td>
<td>803</td>
<td>751</td>
<td>713</td>
</tr>
<tr>
<td>Feb-12</td>
<td>559</td>
<td>847</td>
<td>353</td>
<td>745</td>
</tr>
<tr>
<td>Mar-12</td>
<td>265</td>
<td>355</td>
<td>154</td>
<td>396</td>
</tr>
<tr>
<td>Apr-12</td>
<td>129</td>
<td>166</td>
<td>107</td>
<td>126</td>
</tr>
<tr>
<td>May-12</td>
<td>222</td>
<td>132</td>
<td>238</td>
<td>75</td>
</tr>
<tr>
<td>Jun-12</td>
<td>151</td>
<td>183</td>
<td>97</td>
<td>132</td>
</tr>
<tr>
<td>Jul-12</td>
<td>138</td>
<td>146</td>
<td>151</td>
<td>156</td>
</tr>
<tr>
<td>Aug-12</td>
<td>130</td>
<td>N/A</td>
<td>141</td>
<td>182</td>
</tr>
<tr>
<td>Sep-12</td>
<td>188</td>
<td>N/A</td>
<td>131</td>
<td>284</td>
</tr>
<tr>
<td>Oct-12</td>
<td>98</td>
<td>N/A</td>
<td>190</td>
<td>322</td>
</tr>
<tr>
<td>Nov-12</td>
<td>159</td>
<td>N/A</td>
<td>160</td>
<td>168</td>
</tr>
<tr>
<td>Dec-12</td>
<td>115</td>
<td>N/A</td>
<td>154</td>
<td>153</td>
</tr>
</tbody>
</table>

* N/A means that either the technologist left the job or the technologist had not started working at PET facility
Appendix B  CT/PET room at Duke Cancer Center

Figure B1 Blue print of the PET/CT suite. Blue circles indicate the positions of detectors within the room (viewed from top)
**Appendix C** Berger’s formula used in approximating build-up factors

\[ B(x) = 1 + axe^{bx}, \quad x = \mu r \]

**Table E.6. Parameters for the Berger form of the air-kerma buildup factor.**

<table>
<thead>
<tr>
<th>Photon energy (MeV)</th>
<th>Air $a$</th>
<th>Air $b$</th>
<th>Water $a$</th>
<th>Water $b$</th>
<th>Concrete $a$</th>
<th>Concrete $b$</th>
<th>Iron $a$</th>
<th>Iron $b$</th>
<th>Lead $a$</th>
<th>Lead $b$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.015</td>
<td>0.08</td>
<td>-0.034</td>
<td>0.09</td>
<td>-0.036</td>
<td>0.01</td>
<td>-0.029</td>
<td>0.00</td>
<td>0.00</td>
<td>0.00</td>
<td>0.00</td>
</tr>
<tr>
<td>0.020</td>
<td>0.23</td>
<td>-0.032</td>
<td>0.26</td>
<td>-0.032</td>
<td>0.03</td>
<td>-0.041</td>
<td>0.02</td>
<td>0.032</td>
<td>0.00</td>
<td>0.00</td>
</tr>
<tr>
<td>0.030</td>
<td>0.93</td>
<td>-0.009</td>
<td>1.01</td>
<td>-0.006</td>
<td>0.10</td>
<td>-0.036</td>
<td>0.01</td>
<td>0.036</td>
<td>0.00</td>
<td>0.00</td>
</tr>
<tr>
<td>0.040</td>
<td>2.40</td>
<td>0.018</td>
<td>2.58</td>
<td>0.024</td>
<td>0.26</td>
<td>-0.035</td>
<td>0.02</td>
<td>0.032</td>
<td>0.01</td>
<td>0.066</td>
</tr>
<tr>
<td>0.050</td>
<td>4.05</td>
<td>0.050</td>
<td>4.36</td>
<td>0.057</td>
<td>0.52</td>
<td>-0.026</td>
<td>0.04</td>
<td>0.034</td>
<td>0.01</td>
<td>0.046</td>
</tr>
<tr>
<td>0.060</td>
<td>5.27</td>
<td>0.075</td>
<td>5.59</td>
<td>0.082</td>
<td>0.78</td>
<td>-0.009</td>
<td>0.07</td>
<td>0.039</td>
<td>0.01</td>
<td>0.028</td>
</tr>
<tr>
<td>0.080</td>
<td>6.11</td>
<td>0.102</td>
<td>6.47</td>
<td>0.108</td>
<td>1.42</td>
<td>0.007</td>
<td>0.14</td>
<td>0.034</td>
<td>0.02</td>
<td>0.029</td>
</tr>
<tr>
<td>0.100</td>
<td>5.93</td>
<td>0.113</td>
<td>6.11</td>
<td>0.120</td>
<td>1.83</td>
<td>0.028</td>
<td>0.24</td>
<td>0.030</td>
<td>0.20</td>
<td>0.479</td>
</tr>
<tr>
<td>0.150</td>
<td>4.70</td>
<td>0.121</td>
<td>4.88</td>
<td>0.125</td>
<td>2.19</td>
<td>0.054</td>
<td>0.52</td>
<td>0.015</td>
<td>0.21</td>
<td>0.075</td>
</tr>
<tr>
<td>0.200</td>
<td>3.94</td>
<td>0.133</td>
<td>4.13</td>
<td>0.118</td>
<td>2.20</td>
<td>0.065</td>
<td>0.77</td>
<td>0.064</td>
<td>0.08</td>
<td>0.054</td>
</tr>
<tr>
<td>0.300</td>
<td>3.10</td>
<td>0.094</td>
<td>3.18</td>
<td>0.096</td>
<td>2.03</td>
<td>0.067</td>
<td>1.06</td>
<td>0.022</td>
<td>0.08</td>
<td>0.040</td>
</tr>
<tr>
<td>0.400</td>
<td>2.61</td>
<td>0.079</td>
<td>2.67</td>
<td>0.089</td>
<td>1.87</td>
<td>0.061</td>
<td>1.15</td>
<td>0.033</td>
<td>0.11</td>
<td>0.033</td>
</tr>
<tr>
<td>0.500</td>
<td>2.29</td>
<td>0.067</td>
<td>2.32</td>
<td>0.068</td>
<td>1.73</td>
<td>0.055</td>
<td>1.16</td>
<td>0.036</td>
<td>0.15</td>
<td>0.028</td>
</tr>
<tr>
<td>0.600</td>
<td>2.05</td>
<td>0.058</td>
<td>2.07</td>
<td>0.059</td>
<td>1.60</td>
<td>0.049</td>
<td>1.14</td>
<td>0.036</td>
<td>0.19</td>
<td>0.024</td>
</tr>
<tr>
<td>0.800</td>
<td>1.71</td>
<td>0.045</td>
<td>1.74</td>
<td>0.045</td>
<td>1.41</td>
<td>0.049</td>
<td>1.09</td>
<td>0.032</td>
<td>0.25</td>
<td>0.019</td>
</tr>
<tr>
<td>1.000</td>
<td>1.50</td>
<td>0.035</td>
<td>1.50</td>
<td>0.036</td>
<td>1.27</td>
<td>0.032</td>
<td>1.03</td>
<td>0.028</td>
<td>0.30</td>
<td>0.015</td>
</tr>
<tr>
<td>1.500</td>
<td>1.16</td>
<td>0.021</td>
<td>1.16</td>
<td>0.021</td>
<td>1.02</td>
<td>0.021</td>
<td>0.88</td>
<td>0.020</td>
<td>0.36</td>
<td>0.007</td>
</tr>
<tr>
<td>2.000</td>
<td>0.97</td>
<td>0.013</td>
<td>0.97</td>
<td>0.013</td>
<td>0.89</td>
<td>0.014</td>
<td>0.76</td>
<td>0.018</td>
<td>0.38</td>
<td>0.004</td>
</tr>
<tr>
<td>3.000</td>
<td>0.75</td>
<td>0.005</td>
<td>0.74</td>
<td>0.005</td>
<td>0.71</td>
<td>0.007</td>
<td>0.66</td>
<td>0.014</td>
<td>0.37</td>
<td>0.019</td>
</tr>
<tr>
<td>4.000</td>
<td>0.61</td>
<td>0.001</td>
<td>0.62</td>
<td>0.000</td>
<td>0.59</td>
<td>0.004</td>
<td>0.56</td>
<td>0.015</td>
<td>0.31</td>
<td>0.038</td>
</tr>
<tr>
<td>5.000</td>
<td>0.53</td>
<td>-0.002</td>
<td>0.52</td>
<td>0.002</td>
<td>0.49</td>
<td>0.004</td>
<td>0.49</td>
<td>0.017</td>
<td>0.24</td>
<td>0.062</td>
</tr>
<tr>
<td>6.000</td>
<td>0.47</td>
<td>-0.004</td>
<td>0.47</td>
<td>-0.005</td>
<td>0.45</td>
<td>0.002</td>
<td>0.42</td>
<td>0.021</td>
<td>0.19</td>
<td>0.082</td>
</tr>
<tr>
<td>8.000</td>
<td>0.37</td>
<td>-0.004</td>
<td>0.38</td>
<td>-0.006</td>
<td>0.36</td>
<td>0.001</td>
<td>0.33</td>
<td>0.028</td>
<td>0.11</td>
<td>0.125</td>
</tr>
<tr>
<td>10.000</td>
<td>0.31</td>
<td>-0.004</td>
<td>0.31</td>
<td>-0.005</td>
<td>0.30</td>
<td>0.003</td>
<td>0.25</td>
<td>0.039</td>
<td>0.07</td>
<td>0.161</td>
</tr>
<tr>
<td>15.000</td>
<td>0.23</td>
<td>-0.006</td>
<td>0.23</td>
<td>-0.008</td>
<td>0.21</td>
<td>0.004</td>
<td>0.15</td>
<td>0.066</td>
<td>0.00</td>
<td>0.000</td>
</tr>
</tbody>
</table>


David K Trubey (1966) *A survey of empirical functions used to fit gamma ray build up factors.*

Oak Ridge National Laboratory, Radiation Shielding Information Center.
Appendix D Experimental set-up used by Das et al. to study backscatter from high Z materials

Figure D1. Experimental Set-up used by Das et al. to study backscatter from high Z materials for kilovoltage photon beams; E is the beam energy, A is the field size, t and w are the thickness and width, respectively, of the square area of the high-Z medium, d is the distance of the interface from the surface, x is the distance of the point of measurement from the interface, θ is the beam angle, \( D_i \) is the absorbed dose with high-Z material and \( D_h \) is the absorbed dose without high-Z material.

The ratio of dose with and without presence of high-Z material was given by backscatter dose factor (BSDF) defined by Equation F1 and Figure F1.

\[
BSDF(E, A, w, d, t, x, Z, \theta) = \frac{D_i}{D_h}
\]  

(F1)
Figure D2. BSDF as a ratio of distance for beams with different energies from a lead interface
Figure D3. BSDF as a function of field size and energy from a lead interface
References


