Evaluation of Patient Effective Dose of Neurovascular Imaging Protocols of a C-arm Cone-beam CT

&

Estimation of Current Source Radioactivity of a Cs-137 Irradiator

by

Chu Wang

Graduate Program in Medical Physics
Duke University

Date:_______________________

Approved:

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Terry Yoshizumi, Supervisor

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RathnayakaGunasingha

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James Colsher

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

2012
ABSTRACT

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Abstract

Purpose:

(Project 1) The purpose of this study was three-fold: 1) to estimate the organ doses and effective dose (ED) for patients undergoing neurovascular imaging protocols, 2) to study the effect of beam collimation on ED for 3-D imaging protocols, and 3) to derive protocol-specific DAP-to-ED conversion factors.

(Project 2) The Cs-137 irradiator is one of the most commonly used irradiation device in radiobiological research. The purpose of this study is to develop a simple method to estimate the current source radioactivity of a Cs-137 irradiator (Mark I-68A, JL Shepherd).

Methods:

(Project 1) A cone-beam CT system (Philips AlluraXper FD20/20) was used to measure the organ doses for seven 3-D (cone-beam CT and 3-D Rotational Angiography protocols) and eight 2-D (fluoroscopy and digital subtraction angiography) imaging protocols. Organ dose measurements were performed on an adult male anthropomorphic phantom (CIRS, Norfolk, VA) with 20 MOSFET detectors (Best Medical Canada, Ottawa, Canada) placed in selected organs. The dose area product (DAP) values were recorded from console. The ED values were computed by multiplying measured organ doses to corresponding ICRP 103 tissue weighting factors.
The ED of four 3-D imaging protocols were also measured with standardized beam collimation to compare with the ED associated with the same protocols without beam collimation.

(Project 2) Three positions along the peak-dose irradiation direction within the irradiation chamber were picked as the reference dosimetry positions. Individual dose rate at each of these positions was measured by an ion chamber in “Gy/sec”, as well as estimated by Monte Carlo simulation in “Gy/primary event”. The source activity, “disintegration/sec”, was then derived from these two sets of values and corrected by the branching ratio of the main 662 keV emission.

Results:

(Project 1) For the seven 3-D imaging protocols with uncollimated setting, the EDs ranged from 0.16 mSv to 1.6 mSv, and the DAP-to-ED conversion factors range from 0.037 to 0.17 mSv/Gy·cm². For four protocols with beam collimation, ED was reduced approximately by a factor of 2, and the DAP-to-ED conversion factors by approximately 30%. For the eight 2-D imaging protocols, the ED rates ranged from 0.02 mSv/sec to 0.04 mSv/sec (for DSA) and from 0.0011 mSv/sec to 0.0027 mSv/sec (for fluoroscopy), and the DAP-to-ED conversion factors range from 0.045 to 0.068 mSv/Gy·cm² (for DSA) and factors range from 0.0029 to 0.059 mSv/Gy·cm² (for fluoroscopy).
(Project 2) For the irradiator in question, the source activity, as of Nov. 17, 2011, was estimated to be 2770 Curies. The current activity from the manufacturer was calculated to be 5900 Curies.

Conclusion:

(Project 1) We have measured ED for standard adult neuro imaging protocols in a C-arm cone-beam CT system. Our results provide a simple means of ED estimation using DAP values from console in the C-arm cone-beam CT system.

(Project 2) Our method offers a convenient means to estimate the source activity. The result was compared to the value computed from the manufacturer. We have found discrepancies between the two: 41%, 86%, and 97%, assessed at location 1, 2, and 3, respectively.
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Acknowledgements

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

Duke Radiation Safety and Duke University Medical Center:

- Terry Yoshizumi Ph.D.
- Rathnayaka Gunasingha Ph.D.
- James Colsher Ph.D.
- Greta Toncheva M.S.
- Giao Nguyen M.S.
- Ann Paschall B.S.

Philips Healthcare:

- Xianxian Jiang M.Eng.

Finally, I owe my deepest gratitude to my parents for the confidence they have vested in me – their love and blessing have always given me strength to move forward in life.
1. Introduction

1.1 Overview

Radiation dosimetry is a scientific subject that involves the measurement and assessment of deposited energy (or “radiation dose”), by indirect and direct exposure to ionizing radiation, in a medium. In its many areas of applications, particularly in medicine and radiobiological research, radiation dosimetry plays a key role in ensuring patient safety and the accurate delivery of radiation.

In this thesis, two research projects in radiation dosimetry are discussed: 1) Evaluation of patient effective dose of neurovascular imaging protocols of a C-arm Cone-beam CT (CBCT) system, and 2) Estimation of source activity of a Cs-137 irradiator.

When designing a radiation dosimetry study, two fundamental questions are always asked first: What needs to be measured or evaluated, and what instrumentation or evaluating methodology is best suited for the purpose. Thereby in the following sections, descriptions and explanations are given for: basic quantities and units in radiation dosimetry (Section 1.2); basic operation principles of radiation detectors used in the studies, ion chamber and MOSFET (Metal Oxide Semiconductor Field Effect Transistor) dosimeters (Sections 1.3 and 1.4, respectively); basic rationale of Monte Carlo
dose-estimation methods (Section 1.5); and MC-based dose-estimation software, PCXMC (Section 1.6).

1.2 Basic Quantities and Units in kV Radiation Dosimetry

This section gives descriptions to the basic quantities and units of radiation dosimetry involved in this thesis. The general relationship among some of the key quantities in human dosimetry is shown in Figure 1.1.

**Figure 1.1 Key quantities in human radiation dosimetry**

![Diagram of key quantities in human radiation dosimetry]

- **Exposure, X**

  A measure of the amount of ionization induced in air by x-rays and gamma rays under the energy of 3 MeV. The SI unit of exposure is *coulomb per kilogram* (C/kg).
**Absorbed Dose, D**

It is defined as the energy deposition in a medium by ionizing radiation per unit mass, expressed in the formula below:

\[
D = \frac{d\epsilon}{dm}
\]  

(1-1)

where \(\epsilon\) is energy imparted by ionizing radiation to any matter with mass \(m\) in a finite volume. The SI unit of absorbed dose is Gray (Gy), where 1 Gy = 1 J/kg.

In practice, the absorbed dose in a medium are often indirectly calculated from in-air exposure measurements, using the following formula:

\[
D_{med} = 33.85 (\mu_{en})_{air}^{med} X_{air}
\]  

(1-2)

\[
D_{med} = f X_{air}
\]  

(1-3)

where \(X\) is the exposure in C/kg, \((\mu_{en})_{air}^{med}\) is the ratio of the mass energy absorption coefficients of the medium and air, and \(f\) is called the “f-factor” which equals to \(33.85(\mu_{en})_{air}^{med}\).

**Equivalent Dose, \(H_e\)**

It is an intermediate measure of biological effect of ionizing radiation that takes into account the different biological effects of different types of radiation. The SI unit of equivalent dose is the **Sievert** (Sv). It is determined as

\[
H_R = \sum R w_R D_R
\]  

(1-4)
where \( w_R \) is the *Radiation Weighting Factors*, convert the absorbed doses of different types of ionizing radiation to a single summated value of equivalent dose and can also be used for a single type of radiation. The Radiation Weighting Factors currently in use were recommended by Publication 103 of the International Commission on Radiological Protection (ICRP) (See Table 1.1) and were mainly based on the *relative biological effectiveness* (RBE) of different types of radiation with some corrections from experimental results from radiobiological research. (ICRP 103, 2007)

<table>
<thead>
<tr>
<th>Radiation type</th>
<th>Radiation weighting factor, ( w_R )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Photons</td>
<td>1</td>
</tr>
<tr>
<td>Electrons and muons</td>
<td>1</td>
</tr>
<tr>
<td>Protons and charged pions</td>
<td>2</td>
</tr>
<tr>
<td>Alpha particles, fission fragments, heavy ions</td>
<td>20</td>
</tr>
<tr>
<td>Neutrons</td>
<td>A continuous function of neutron energy</td>
</tr>
</tbody>
</table>

*Effective Dose, \( H \)*

The *effective dose*, ED, is a more advanced measure of biological effect of ionizing radiation that takes into account the different radiosensitivities of different types of tissues of the human body. Like the equivalent dose, The SI unit of effective dose is also the *Sievert* (Sv). It is determined as

\[
E = \sum_T w_T H_{R,T} \tag{1-4}
\]
where $w_T$ is the *Tissue Weighting Factors*, which convert the equivalent doses in different organs to a single summated value of effective dose for the entire body. The Radiation Weighting Factors currently in use were recommended by ICRP Publication 103 (See Table 1.2) and were based on likelihoods of cancer induction, hereditary effects and relative length of life lost due to radiation. (ICRP 103, 2007)

**Table 1.2 Tissue Weighting Factors (ICRP 103, 2007)**

<table>
<thead>
<tr>
<th>Tissue</th>
<th>$w_T$</th>
<th>$\sum w_T$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bone-marrow (red), colon, lung, stomach, breast, remainder tissues*</td>
<td>0.12</td>
<td>0.72</td>
</tr>
<tr>
<td>Gonads</td>
<td>0.08</td>
<td>0.08</td>
</tr>
<tr>
<td>Bladder, oesophagus, liver, thyroid</td>
<td>0.04</td>
<td>0.16</td>
</tr>
<tr>
<td>Bone surface, brain, salivary glands, skin</td>
<td>0.01</td>
<td>0.04</td>
</tr>
</tbody>
</table>

* Remainder tissues: adrenals, extrathoracic (ET) region, gall bladder, heart, kidneys, lymphatic nodes, muscle, oral mucosa, pancreas, prostate ($\ddagger$), small intestine, spleen, thymus, uterus/cervix ($\ddagger$).

**Dose-Area Product, DAP**

It is a quantity applied only in medical x-ray imaging modalities that not only reflect the radiation dose but also the total area of exposure, as given by

$$DAP = AD$$ (1-5)

where $D$ is the absorbed dose and $A$ is the area of exposure in; hence, DAP has a unit of Gy cm$^2$, or sometimes in mGy cm$^2$.

DAP, by its mathematical definition, is independent of distance (Figure 1.2). As most modern medical x-ray imaging devices have the capability of reporting DAP values immediately after imaging, making it a readily convenient index for patient dosimetry.
Since the DAP value is available after one x-ray imaging protocol (e.g. conventional radiography, fluoscopy, etc.) for a patient, one can easily estimate the patient-specific ED value using a pre-determined DAP-ED conversion factor of that imaging protocol.

**DAP-ED Conversion Factor, CF**

This conversion factor is determined by the simple equation:

$$CF (mSv Gy^{-1} cm^{-2}) = \frac{ED (mSv)}{DAP (Gy cm^2)}$$  \hspace{1cm} (1-6)

Since the DAP value is available after one x-ray imaging protocol (e.g. conventional radiography, fluoscopy, etc.) for a patient, one can easily estimate the patient-specific ED value using a pre-determined DAP-ED conversion factor of that imaging protocol.
1.3 Ion Chamber

Ion chambers are the most common type of devices used to measure ionizing radiation. It has long been established as the standard dosimetry tool in clinical dosimetry due to its simplicity in design and low cost of operation.

Upon irradiation, passing radiation creates a number of ions in the radiation-detecting medium (most commonly gaseous, but can also be solid or liquid). The charges are then collected by two conducting electrodes and induce an ionization current, which can be measured by an electrometer. A simplified schematic diagram of an ion chamber is shown in Figure 1.3.

**Figure 1.3 Simplified schematic of an ion chamber**

The American Association of Physicists in Medicine (AAPM) has prescribed two protocols for reference dosimetry by ion chambers for: a) low- to medium- energy x-ray (Ma, AAPM TG 61, 2001) and b) high-energy photon and electron beams (Almond,
AAPM TG 51, 1999). Both of the protocols require the ion chamber to be calibrated by a NIST traceable source and the following electrometer reading corrections:

\[ M(nC) = P_{TP} P_{\text{ion}} P_{elec} P_{pol} M_{\text{raw}}(nC) \]  

where:

a) \( P_{TP} \) is the correction factor for temperature and pressure:

\[ P_{TP} = \left( \frac{273.2 + T(\degree C)}{295.2} \right) \left( \frac{101.33}{P(kPa)} \right) \]  

where \( T \) is temperature in degrees Celsius and \( P \) is pressure in kilo Pascals.

b) \( P_{\text{ion}} \) is the correction factor for the ion chambers collection inefficiency:

\[ P_{\text{ion}} = \frac{1 - \left( \frac{V_H}{V_L} \right)^2}{\frac{M^H}{M_{\text{raw}}} - \left( \frac{V_H}{V_L} \right)^2} \]  

where \( V_H \) and \( V_L \) are the high and low bias voltages (set on the electrometer) used to find the corresponding high and low raw readings \( M^H \) and \( M^L \).

c) \( P_{\text{elec}} \) is the electrometer correction factor provided by an Accredited Dosimetry Calibration Laboratory.

d) \( P_{pol} \) is the correction factor for polarity effects:

\[ P_{pol} = \left| \frac{M^+_{\text{raw}} - M^-_{\text{raw}}}{2 \cdot M_{\text{raw}}} \right| \]  

(1-10)
where $M^+$ and $M^-$ are the raw readings measured at ± bias.

e) $M_{raw}$ is the raw reading of measurement at the positive or negative bias.

### 1.4 Metal Oxide Semiconductor Field Effect Transistor (MOSFET)

The MOSFET technology is only a new member in the radiation detector family, despite having existed in the electronic world for decades. Due to its technological and practical advantages over some of the conventional dosimeters, such as small size, good isotropy, large dynamic range, radiation type sensitivity and near real-time reading, it has become a reliable tool in medical radiation dosimetry.

**Figure 1.4 Simplified schematic of a p-channel MOSFET**

![Simplified schematic of a p-channel MOSFET](image)

The operation principle of a MOSFET dosimeter is fairly straight-forward. A schematic of a p-channel MOSFET dosimeter is shown in Figure 1.4. Application of a
large negatively biased voltage at the polysilicon gate induces a large number of holes to migrate from the bulk silicon substrate and the source and drain regions to the silicon/oxide interface. With a large concentration of holes in the area, a conduction channel can be formed between the source and the drain. The threshold voltage, \( V_{TH} \), is the minimum voltage needed to initiate current through the channel. Upon irradiation, passing radiation produces electron-hole pairs in the oxide. The electrons move out of the gate electrode rapidly, due to their high mobility in Si O\(_2\) whereas the holes move towards the Si/SiO\(_2\) interface and get trapped there indefinitely. The excess positive charge results in a current in the channel and consequently a permanent shift in the threshold voltage (\( \Delta V_{TH} \)), which can be determined by measuring the voltage shift, before and after irradiation. \( \Delta V_{TH} \) is proportional to dose.

Secondary electrons play the major role in energy deposition in the MOSFET dosimeter. For photon irradiation upon silicon, photoelectric absorption is the dominant process at energies below 50 keV and Compton scattering gains co-dominance at energies between 60 keV and 90 keV. Since the photoelectric cross-section is strongly dependent on target material (\( \tau \propto Z^4 \)), different amounts of electron are produced in Si and Si O\(_2\). This property, along with the varying photoelectric absorption dominance at different energies, leads to energy-sensitivity of the MOSFET dosimeter.
Therefore, before actual measurements, MOSFET dosimeters must be energy-calibrated.

### 1.5 Rationales of the Monte Carlo Dose Estimation Methods Used

Monte Carlo methods are computational algorithms that employ large numbers of random sampling to determine the most statistically significant results from distributions of all random samplings. In radiation dosimetry, Monte Carlo methods are also called Monte Carlo simulations; in such an application, a large number of input particles are simulated to transport through numerically set-up geometries, of materials with known physical properties and radiation-interaction probabilities, and the energy deposition, or dose, at a particular position is tallied after each “run” (i.e. a set number of random inputs) and recorded into the final distribution of results.

All Monte Carlo softwares in radiation dosimetry consist of three general input blocks:

1) Random number generator;

2) Experimental geometry;

3) Physical properties and radiation-interaction probabilities of the materials involved in the geometry.
Simulating with the three input blocks listed above, one can ensure the accuracy and reproducibility of the results, as long as a sufficiently large number of samples have been run.

Because of its software-nature, a Monte Carlo algorithm’s key advantage, over physical dosimetric measurement, is flexibility. MC-based softwares have been widely used in the radiation dosimetry works; for instance, shielding designs of radiation facilities – it saves both time and man-power as most of dosimetric estimations and shielding adjustments can be obtained before the actual construction begins. In the two studies discussed in this thesis, two Monte Carlo software packages, PCXMC and FLUKA, were utilized to serve the different purposes of the studies.

1.6 PCXMC

PCXMC (Version 2.0, STUK, Finland) is a MC-based computer software dedicated for estimation of patient dose from medical x-ray exposures, primarily fluoroscopy and radiography. Simulation customizations are allowed in terms of x-ray beam quality, exposure geometry, as well as patient’s size, age group and sex. The software uses a set of mathematical hermaphrodite phantoms of Cristy and Eckerman (Source), with some modifications and user-adjustable phantom sizes (Source). Based upon its Monte Carlo algorithm, the program calculates organ doses (in mGy) as well as the whole-body effective dose (in mSv). If needed, the organ doses can be further
applied to estimate for the radiation-induced cancer risks using the BEIR VII factors (BEIR Report, 2006). More detailed technical information can be found in the software developer’s own technical publication, (STUK A231, 2008)
2. Evaluation of Patient Effective Dose of Neurovascular Imaging Protocols of a C-arm Cone-beam CT

2.1 Introduction

C-arm CBCT is a relatively new x-ray imaging modality in interventional neuroradiology and endovascular procedures. A C-arm CBCT system is capable of performing the traditional 2-D imaging techniques, such as standard fluoroscopy and DSA (digital subtraction angiography), as well as 3-D imaging techniques through the rotation of the C-arm and CT-like reconstruction algorithms. The 3-D imaging capability has shown to be valuable for detecting procedural complications, evaluating low-radiopacity stents and assessing coil packing in small-sized aneurysms (Kamran, 2010). While the clinical benefit is evident, one cannot overlook the radiation-induced health risk for patients undergoing interventional procedures involving a C-arm CBCT system.

In recent decades, ED has become a key measure for quantifying patient radiation dose in medical x-ray scans (e.g. CT, fluoroscopy, etc.). For C-arm CBCT scans, studies have reported the EDs associated with cardiac imaging protocols, which involve irradiations primarily in the chest region of the patient. (Bai, 2011; Wielandts, 2010) However, to the best of my knowledge, there are very limited information about the ED associated with neurovascular imaging protocols of C-arm cone-beam CT.

In this study, we measured the ED values of neurovascular imaging protocols, both 2-D and 3-D, of a C-arm CBCT system currently in use at the Department of Interventional Radiology, Duke University Medical Centre. In addition, protocol-specific
DAP-ED conversion factors (CF) for the neurovascular imaging protocols were also derived for the convenience in estimating patient-specific ED during neurovascular interventional procedures in the future.

2.2 Material and Methods

Organ dose measurements were conducted using an adult male anthropomorphic phantom (model 701-D; CIRS, Norfolk, VA), with high-sensitivity MOSFET dosimeters (Model TN-1002RD, Best Medical, Ottawa Canada) placed in selected organ locations for the appropriate coverage of the exposures involved.

2.2.1 C-arm CBCT System and Neurovascular Scan Protocols Investigated

The C-arm CBCT system we investigated is an AlluraXper FD20/20 (Philips Medical Systems, the Netherlands). (Figure 2.1) Two groups of neurovascular scan protocols, 3-D and 2-D (traditional planar fluoroscopy and DSA), were investigated, by exposing an adult anthropomorphic phantom (phantom information is presented in Section 2.3) simulating actual patient scans.
2.2.1.13-D Imaging Protocols

Seven 3-D neurovascular protocols, including five cone-beam protocols (called XperCT) and two 3-D Rotational Angiography (3DRA) were investigated (Table 2.1). The direct results were evaluated as ED per individual scan (mSv), as these protocols have defined exposure times and numbers of images. Two sets of patient dose data were obtained: all seven 3-D neuro protocols with uncollimated beam (FOV: entire head) and four (of the seven overall) with beam collimation (FOV: roughly from the base to the top of the skull). (Sample scan images with and without collimation are shown in Figure 2.2) Effect of beam collimation was examined by comparing ED values of the four 3-D protocols that were measured both with and without beam collimation. Finally,
protocol-specific DAP-ED conversion factors were determined for all seven 3-D neuro protocols, including the cases with beam collimation.

**Figure 2.2** Sample images for collimated (left) and uncollimated (right) 3-D scans

### Table 2.1 Neuro 3-D Protocols (manufacturer default)

<table>
<thead>
<tr>
<th>Protocol</th>
<th>XPERCT CEREBRAL HD</th>
<th>XPERCT CEREBRAL LD</th>
<th>XPERCT CEREBRAL FAST HD</th>
<th>XPERCT INTRACRANIAL STENT HD</th>
<th>XPERCT INTRACRANIAL STENT LD</th>
<th>3DRA CEREBRAL PROPELLER</th>
<th>3DRA CEREBRAL ROLL</th>
</tr>
</thead>
<tbody>
<tr>
<td>kV</td>
<td>120</td>
<td>120</td>
<td>120</td>
<td>80</td>
<td>80</td>
<td>80</td>
<td>80</td>
</tr>
<tr>
<td>mAs</td>
<td>250</td>
<td>250</td>
<td>170</td>
<td>260</td>
<td>260</td>
<td>137</td>
<td>138</td>
</tr>
<tr>
<td>ms</td>
<td>5</td>
<td>5</td>
<td>7</td>
<td>7</td>
<td>7</td>
<td>7</td>
<td>7</td>
</tr>
<tr>
<td>fps</td>
<td>39</td>
<td>39</td>
<td>39</td>
<td>39</td>
<td>39</td>
<td>39</td>
<td>15</td>
</tr>
<tr>
<td>#image</td>
<td>621</td>
<td>312</td>
<td>624</td>
<td>621</td>
<td>621</td>
<td>120</td>
<td>120</td>
</tr>
<tr>
<td>scan time (s)</td>
<td>20.8</td>
<td>10.4</td>
<td>10.4</td>
<td>20.8</td>
<td>20.8</td>
<td>4</td>
<td>8</td>
</tr>
<tr>
<td>C-arm motion</td>
<td>&quot;propeller&quot; movement starting at -120° and end at 120°</td>
<td>&quot;roll&quot; movement from (-120°) to (120°)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>degree/s</td>
<td>10</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>detector field (FD)</td>
<td>28.8 x 38.4 cm (diagonal 48 cm)</td>
<td>16 x 16 cm (diagonal 22 cm)</td>
<td></td>
<td>28.8 x 38.4 cm (diagonal 48 cm)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>filters</td>
<td>0.4 mm Cu + 1.0 mm Al</td>
<td>0.4 mm Cu + 1.0 mm Al</td>
<td>0.4 mm Cu + 1.0 mm Al</td>
<td>0 mm Cu + 0 mm Al</td>
<td>0 mm Cu + 0 mm Al</td>
<td>0 mm Cu + 1.0 mm Al</td>
<td>0.1 mm Cu + 1.0 mm Al</td>
</tr>
</tbody>
</table>
2.2.1.22-D Imaging Protocols

Eight DSA and standard fluoroscopy protocols were investigated (Table 2.2). These 2-D scan protocols vary in patient orientations (AP vs. LAT), intensity ("FluoroFlavours" of High, Normal, or Low) and Source-to-Image distances (SID). The direct results were evaluated as ED rate (mSv/sec), as the scanning times of these protocols are manually controlled with footpads and/or push buttons, and therefore are subject to the preferences of clinical practice of individual operators.

The 2-D protocols were also simulated with the software, PCXMC, for an independent estimation of protocol-specific ED rates. The results from MOSFET measurements and PCXMC simulations were compared. Meanwhile, however, the difference in the phantoms used by the two methods must be taken into account: the PCXMC software uses a mathematical hermaphrodite near-MIRD phantom, whereas the MOSFET measurement was performed on a CIRS adult male phantom. Finally, protocol-specific DAP-ED conversion factors were determined for all eight protocols.
Table 2.2 2-D Imaging Protocols. Top: Fluoroscopy; Bottom: DSA.

<table>
<thead>
<tr>
<th>Protocol</th>
<th>FLUORO LAT FLAVOR LOW</th>
<th>FLUORO AP FLAVOR LOW</th>
<th>FLUORO AP FLAVOR NORMAL</th>
<th>FLUORO AP FLAVOR HIGH SID=120 CM</th>
<th>FLUORO AP FLAVOR HIGH SID=100 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>kV</td>
<td>70/77</td>
<td>88</td>
<td>80</td>
<td>77</td>
<td>74/88</td>
</tr>
<tr>
<td>mA</td>
<td>7.5/10.3</td>
<td>12.9</td>
<td>17.7</td>
<td>13.6</td>
<td>10.8/54</td>
</tr>
<tr>
<td>fps</td>
<td>2</td>
<td>2</td>
<td>2</td>
<td>2</td>
<td>2</td>
</tr>
<tr>
<td>SID (mm)</td>
<td>1200</td>
<td>1200</td>
<td>1200</td>
<td>1200</td>
<td>1200</td>
</tr>
<tr>
<td>FD (mm)</td>
<td>420</td>
<td>420</td>
<td>420</td>
<td>420</td>
<td>420</td>
</tr>
<tr>
<td>ROT/ANG</td>
<td>-90°/0°</td>
<td>0°/0°</td>
<td>0°/0°</td>
<td>0°/0°</td>
<td>0°/0°</td>
</tr>
<tr>
<td>detector field (FD)</td>
<td></td>
<td></td>
<td>30 x 30 cm (diagonal 42 cm)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>filters</td>
<td>0.5 mm Cu + 1.0 mm Al</td>
<td>0.4 mm Cu + 1.0 mm Al</td>
<td>0.1 mm Cu + 1.0 mm Al</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

MOSFET Dosimeters and Calibration with Ion Chamber

As discussed in Section 1.4, MOSFET dosimeters require appropriate energy calibrations under irradiations of different beam qualities. In this study, an ion chamber was used to calibrate the 20 MOSFETs for all beam qualities involved with the scan protocols. (Figure 2.3) In particular, the kVp values of the 2-D imaging protocols are automatically modulated according to patient thickness and calibration were performed with added aluminum plates in the FOV so that the kVp value was driven up to the actual patient-imaging value for each protocol.
2.2.3 Adult Anthropomorphic Phantom and MOSFET dosimeter placement

An adult male anthropomorphic phantom (model 701-D; CIRS, Norfolk, VA) was used to simulate for an adult patient. The phantom is made of materials equivalent to bone, lung, and soft-tissue, with compositions based on published data of the reference man (ICRP, 1975) and of the tissue substitutes in radiation dosimetry. (ICRU, 1989) The phantom consisted of 39 contiguous 2.5-cm-thick slices, with a number of 5-mm-diameter holes in each slice for dosimeter placement. Its specifications are listed in Table 2.3.
The 20 MOSFET dosimeters were placed in selected organs/tissue types in the phantom. (Table 2.4) Due to the fact that all neurovascular scan protocols studied involve exposures to the head region of the patient, the majority of the MOSFETs were placed in the head/neck region (e.g. brain, skull, thyroid, etc.) for the direct irradiation in the FOV and the several others in the chest region for scattered irradiation outside of the FOV.
Table 2.4 MOSFET dosimeter placement in the phantom

<table>
<thead>
<tr>
<th>MOSFET #</th>
<th>Phantom Slice</th>
<th>Location Number</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Skin</td>
<td>3</td>
</tr>
<tr>
<td>2</td>
<td>Skin</td>
<td>3</td>
</tr>
<tr>
<td>3</td>
<td>BM skull</td>
<td>3</td>
</tr>
<tr>
<td>4</td>
<td>BM skull</td>
<td>3</td>
</tr>
<tr>
<td>5</td>
<td>Brain Lt</td>
<td>3</td>
</tr>
<tr>
<td>6</td>
<td>Brain Lt</td>
<td>3</td>
</tr>
<tr>
<td>7</td>
<td>Brain</td>
<td>4</td>
</tr>
<tr>
<td>8</td>
<td>Brain</td>
<td>4</td>
</tr>
<tr>
<td>9</td>
<td>Lens of the eye</td>
<td>5</td>
</tr>
<tr>
<td>10</td>
<td>Lens of the eye</td>
<td>5</td>
</tr>
<tr>
<td>11</td>
<td>Salivary glands (Bone surface)</td>
<td>7</td>
</tr>
<tr>
<td>12</td>
<td>BM - c-spine</td>
<td>8</td>
</tr>
<tr>
<td>13</td>
<td>Thyroid</td>
<td>10</td>
</tr>
<tr>
<td>14</td>
<td>Oesophagus</td>
<td>12</td>
</tr>
<tr>
<td>15</td>
<td>Lung</td>
<td>12</td>
</tr>
<tr>
<td>16</td>
<td>BM - clavicle</td>
<td>12</td>
</tr>
<tr>
<td>17</td>
<td>BM - sternum</td>
<td>13</td>
</tr>
<tr>
<td>18</td>
<td>Thymus</td>
<td>13</td>
</tr>
<tr>
<td>19</td>
<td>BM - T/L spine</td>
<td>13</td>
</tr>
<tr>
<td>20</td>
<td>Breast</td>
<td>17</td>
</tr>
</tbody>
</table>

2.2.4 Effective Dose Evaluation with Partial-organ Irradiation

The ICRP tissue weighting factors were initially recommended for wholebody irradiations. In the neuro-imaging protocols of our study, three organs receive partial-organ irradiation: Skin, red marrow and bone surface. Ignoring this issue and applying the full values of the respective tissue weighting factors may lead to significantly overestimated ED and hence radiation-risk. Therefore, partial-volume factors, $v_T$, are applied to the original ED definition:
\[ E = \sum_{T} v_{T} w_{T} H_{R,T} \]  \hspace{1cm} (2-1)

Where:

a) For skin, \( v_{T} = 9\% \). It is according to the percent-skin-area estimation, “Rule of the Nines”, from which estimates, for adults, the skin area in the head, chest (front), abdomen (front), upper back, lower back, each arm, each upper leg and each lower leg, each accounts for 9\% of the total-body skin area and the genitals accounts for the remaining 1\% (Dorland’s Medical Dictionary for Health Consumers, 2007).

b) For bone marrow, \( v_{T} = 51\% \). It is derived from the standard man Bone Marrow distributions. The 51\% include contributions from the skull, scapula, clavicles, upper- and middle-portion spine. In a typical neuro scan, only the skull is directly irradiated while the other portions receive considerably less scattered radiation.

c) For bone surface, \( v_{T} = 20\% \). This is a conservative estimation for the distribution of bone surface in the head region. And

d) For all other organs, \( v_{T} = 100\% \).
2.3 Results and Discussion

2.3.1 Effective Dose Evaluation for 3-D Imaging Protocols

Table 2.5 Effective dose and DAP-ED conversion factors of 3-D scan protocols

<table>
<thead>
<tr>
<th></th>
<th>Uncollimated Beam</th>
<th></th>
<th>Collimated Beam</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Effective Dose</td>
<td>DAP-ED CF</td>
<td>Effective Dose</td>
<td>DAP-ED CF</td>
</tr>
<tr>
<td></td>
<td>(mSv)</td>
<td>(mSv Gy⁻¹ cm⁻²)</td>
<td>(mSv)</td>
<td>(mSv Gy⁻¹ cm⁻²)</td>
</tr>
<tr>
<td>XperCT Cerebral HD</td>
<td>1.6</td>
<td>0.065</td>
<td>0.8</td>
<td>0.048</td>
</tr>
<tr>
<td>XperCT Cerebral LD</td>
<td>0.8</td>
<td>0.062</td>
<td>0.38</td>
<td>0.047</td>
</tr>
<tr>
<td>XperCT Cerebral FAST HD</td>
<td>1.1</td>
<td>0.070</td>
<td>0.51</td>
<td>0.046</td>
</tr>
<tr>
<td>XperCT Intracranial Stent HD</td>
<td>1.7</td>
<td>0.037</td>
<td>0.8</td>
<td>0.025</td>
</tr>
<tr>
<td>XperCT Intracranial Stent LD</td>
<td>0.52</td>
<td>0.17</td>
<td>----</td>
<td>----</td>
</tr>
<tr>
<td>3DRA Cerebral Propeller</td>
<td>0.16</td>
<td>0.043</td>
<td>----</td>
<td>----</td>
</tr>
<tr>
<td>3DRA Cerebral Roll</td>
<td>0.24</td>
<td>0.076</td>
<td>----</td>
<td>----</td>
</tr>
</tbody>
</table>

The measured effective dose and DAP-ED conversion factors of the 3-D scan protocols, with and without collimation, are listed in Table 2.5. The ED values for the “XperCT Cerebral” protocols behaved well according to their difference in scan parameters; for example, the protocol, “XperCT Cerebral HD” had twice the sampling (number of images) rate than the “XperCT Cerebral LD”, and consequently the measured ED value of the former protocol is twice of that of the latter. The “XperCT Intracranial Stent” protocols only differ in detector size by an approximate ratio of 4:1 (48cm-diagonal rectangle vs. 22cm-diagonal square), and their ED values had a evident
4:1 ratio. In addition, Beam collimation reduced the ED by approximately 50% for the four protocols investigated with and without collimation.

In a neurovascular 3-D scan, six organs receive the most significant organ dose: skin, brain, red marrow, lens of the eye, salivary glands and thyroid. (Figure 2.4, using “XperCT Cerebral HD” as example). Furthermore, beam collimation greatly reduced the dose to the salivary glands and thyroid, which became out of the FOV with the applied collimation, and slightly reduced the dose to other organs, which were still within the FOV but received less scatter radiation with collimation. Therefore, if acceptable in procedural circumstances, beam collimation should be applied to reduce the patient’s risks of cancer-induction in the salivary glands and thyroid.

**Figure 2.4 Main organ doses from the protocol “XperCT Cerebral HD”, with and without beam collimation (Note: the skin dose listed is the entrance skin dose)**
Comparing to the traditional Multi-slice CT (MSCT), it is not as easy to compare EDs of different C-arm CBCT protocols, which involve many more scan parameters than MSCT, such as the C-arm motion parameters. The ED of a typical head CT scan is around 1.5 mSv, depending on the specific protocol settings (Mettler, 2000). Specifically for C-arm CBCT, Bai et al have reported the ED of head protocols of the DynaCT, another C-arm CBCT system, to be 1.18 mSv (70 kV, 216 mA, 20s scan time) and 0.85 mSv (90 kV, 115 mA, 8s scan time). (Bai, 2011) Our findings are in general agreement with theirs, after normalizing the general scan parameters.

The ICRP tissue weighting factors were initially recommended for wholebody irradiations. In the neuro-imaging protocols of our study, three organs receive partial-organ irradiation: Skin, red marrow and bone surface. Ignoring this issue and applying the full values of the respective tissue weighting factors may lead to significantly overestimated ED and hence radiation-risk. Therefore, partial-volume factors, $\nu_7$, are applied to the original ED definition:
2.3.2 Effective Dose Evaluation for 2-D Imaging Protocols

Table 2.6 Effective dose rates and DAP-ED conversion factors of 2-D scan protocols (fluoroscopy and DSA)

<table>
<thead>
<tr>
<th>Effective Dose Rate DAP-ED CF</th>
<th>Effective Dose Rate (mSv/sec)</th>
<th>DAP-ED CF (mSvGy⁻¹ cm⁻²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>XperCT DSA protocols</td>
<td></td>
<td></td>
</tr>
<tr>
<td>DSA AP CEREBRAL 2FPS SID=120cm</td>
<td>0.044</td>
<td>0.045</td>
</tr>
<tr>
<td>DSA AP CEREBRAL 2FPS SID=100cm</td>
<td>0.029</td>
<td>0.045</td>
</tr>
<tr>
<td>DSA LAT CEREBRAL 2FPS</td>
<td>0.022</td>
<td>0.068</td>
</tr>
<tr>
<td>XperCT FLUORO protocols</td>
<td></td>
<td></td>
</tr>
<tr>
<td>FLUORO LAT FLAVOR LOW</td>
<td>0.0011</td>
<td>0.059</td>
</tr>
<tr>
<td>FLUORO AP FLAVOR LOW</td>
<td>0.0021</td>
<td>0.058</td>
</tr>
<tr>
<td>FLUORO AP FLAVOR NORMAL</td>
<td>0.0039</td>
<td>0.051</td>
</tr>
<tr>
<td>FLUORO AP FLAVOR HIGH SID=120cm</td>
<td>0.0056</td>
<td>0.042</td>
</tr>
<tr>
<td>FLUORO AP FLAVOR HIGH SID=100cm</td>
<td>0.0027</td>
<td>0.029</td>
</tr>
</tbody>
</table>

The measured effective dose rates and DAP-ED conversion factors of the 2-D scan protocols (fluoroscopy and DSA) are listed in Table 2.4. For both DSA and fluoroscopy, scanning in the Anterior-Posterior (AP) direction result in a greater ED than in the Lateral (LAT) direction, which can be explained by the fact that the patient thickness is greater in the AP-direction leading to a more intensive beam due to the automatic exposure modulation. Another observation from Table 2.4 is that a decrease in SID results in significant decrease in ED, since the area of exposure for the patient body gets smaller as the SID is reduced. Finally, our findings of the DAP-ED CF agreed with
reported DAP-ED CF for cerebral angiography for the UK population, which ranges from 0.028 to 0.087 mSv Gy\(^{-1}\) cm\(^2\). (Hart and Wall, 2002).

**Figure 2.5 Comparison of ED evaluations: MOSFET vs. PCXMC (top: DSA protocols; bottom: Fluoroscopy protocols)**

The MOSFET-measured ED of the 2-D scan protocols are compared with PCXMC simulations (Figure 2.5). While PCXMC use Monte-Carlo data to calculate patient dose,
there are two key unrealistic factors for the program: First, the program base their simulations on a mathematical hermaphroditic phantom, instead of an anthropomorphic phantom used in the MOSFET measurement; and second, the PCXMC simulations do not include the presence of the patient table. The difference in anatomy is not significant for the head region across different phantoms, thus the effect of the first factor is considered negligible. However, the second factor, the absence of table attenuation in PCXMC simulations, is potentially significant. As observed in Figure 2.5, ED rates derived from both techniques for protocols in the Lateral direction (no table attenuation involved) are in good agreement with each other, whereas the ED rates for protocols in the AP direction (table attenuation directly involved) displayed significant differences - PCXMC overestimates by approximately a factor of 2. Unpublished data by Fredrickson reported that the table attenuations of two GE CT scanners are roughly 40% (Fredrickson, 2011). With similar function and purpose, the patient table of the Philips C-arm CBCT may have similar attenuation effects, thus leading to the discrepancy in ED estimation between the two techniques.

2.4 Conclusion

C-arm CBCT is playing an increasingly crucial role in interventional neuroradiology procedures, due to its 3-D imaging capabilities. The results reported from this study provide the clinicians important information on the aspect of radiation-induced risk to patients, such as the effect of beam collimation and SID on effective dose.
Furthermore, the determined protocol-specific DAP-ED conversion factors will also aid in patient-specific dose estimation for this system in the future.
3. Estimation of Current Source Radioactivity of a Cs-137 Irradiator

3.1 Introduction

The Cs-137 radioisotope is one of the most commonly used sources in clinical and radiobiological research, due to its long half-life and predominant gamma ray emissions (662 keV, 94.6%). At Duke University, a Cs-137 irradiator is depended on by a number of research groups for their small-animal/cell-sample irradiation experiments.

Dose rates at the three reference dosimetry positions in the irradiation cavity have been previously measured and documented as reference information. However, most of the experiments involve using apparatus with varied geometries (e.g. a metallic holder that hold multiple test tubes) and irradiation orientation (e.g. turntable rotation), the aforementioned dose rates at reference dosimetry positions therefore should be verified for accuracy of the actual irradiated samples. In the cases of such experiments, individual dosimetric measurement, usually with Thermo-Luminescent Dosimeters (TLDs), has to be performed with each different experimental design; such measurements exert additional pressure on both the manpower and resources on the research groups. While physical measurements for the irradiator have shown their limitations, dose estimation by Monte Carlo simulations, on the other hand, can be of great assistance in terms of accuracy and convenience.

A Monte Carlo simulation of a source should always start with a known value of source radioactivity, but unfortunately, the exact activity is unknown for this
irradiatorexcept for an approximated initial activity of 10,000 Curies, documented on the source certificate dated October 1988. Therefore, this study aims to determine the true source activity of this irradiator. The result of this study will aid in the actual dosimetric Monte Carlo simulations for the variable experimental designs using the irradiator.

3.2 Material & Methods

3.2.1 Cs-137 Irradiator

The irradiator of interest is a Mark I-68A Cs-137 Irradiator (JL Shepherd and Associates, San Fernando, CA). The irradiator was housed in a compact, self-shielded module, Figure 3.1a, with an irradiation cavity of the size (37cm tall x 30cm wide x 45cm deep), Figure 3.1b. Three motor shafts are also present on the floor of the irradiation cavity, Figure 3.1c; the positions are 5.5cm, 15cm and 20cm away from the source guide and are labeled 1, 2 and 3, from closest to farthest from the source. An aluminum turntable can be placed on any of the motor shafts to rotate the samples.
3.2.2 Monte Carlo Simulation

A Monte Carlo software package, FLUKA (FLUktuierendeKAskade) (Ferrari et al, 2005; Battitoni et al, 2006), was used to simulate the Cs-137 irradiator. Three cross-sectional views of the modeled irradiator geometry are shown in Fig. 3.2. Doses in water per primary event, $D_{w,\text{primary}}$, at the three reference dosimetry positions, which are 17 cm above three motor shafts on the floor of the irradiation chamber (also indicated in Fig. 3.2). In order to obtain statistically significant data, a simulation of $53 \times 10^8$ primary events, were conducted.
Figure 3.2 Modeled geometry of the irradiator, in X-Y (top plot), Y-Z (middle plot) and X-Z (bottom plot) cross-sectional views. The “X” in the plots mark the three reference dosimetry positions, Position 1, 2 and 3. The different colorings in the plots represent different materials: lead (green), purple (air), orange (CsCl), black (stainless steel) and light blue (aluminum).
3.2.3 Ion Chamber-Derived Dose Rate Measurement

Exposure rates at the three reference dosimetry positions were measured by a 0.18 cc ion chamber (10X5-0.18, Radcal, Monrovia, CA) with an ion chamber monitor (9015, Radcal). The measured exposure rates were then converted to dose rates in water, $\dot{D}_w$, using the following formula:

$$\dot{D}_w \, (\text{Gy/sec}) = \dot{X}_{\text{air}} \times CF \times (f - \text{factor})_{\text{air}}^{\text{water}} \, (3-1)$$

where $\dot{X}_{\text{air}}$ is the exposure rate reading measured by the ion chamber with temperature and pressure correction, CF is the ion chamber correction factor provided by the University of Wisconsin Accredited Dosimetry Calibration Laboratory, and the f-factor is used for the dose conversion from in-air measurements to doses in water for the 662 keV gamma rays by Cs-137.

3.2.4 Derivation of True Source Activity

For each reference dosimetry position, a value of the true source activity was then derived using the following relationship:

$$A \, (\text{Bq}) = \frac{\dot{D}_w \, (\text{Gy/sec})}{D_{\text{w,primary}} \, (\text{Gy/primary}) \times BR} \, (3-2)$$

where $\dot{D}_w$ is the ion chamber-measured dose rate in water, $D_{\text{w,primary}}$ is the FLUKA-derived dose in water per primary event, and BR is the branching ratio of the 662 keV emission of Cs-137, which is 94.6%.
### 3.3 Results and Discussion

The values of ion chamber-measured dose rates ($\hat{D}_w$), FLUKA-derived dose per primary ($D_{w, \text{primary}}$), and derived source activities from the three reference dosimetry positions are shown in Table 3.1. The relative error for ion chamber measurements is less than 3% and that of the FLUKA-derived dose per primary values is approximately 20%.

| Table 3.1 Estimated source activities derived from ion chamber measurements and FLUKA dose simulations at the three reference dosimetry positions |
|---|---|---|
| **Positions** | 1 | 2 | 3 |
| $\hat{D}_w$ (Gy/sec) | 0.263 | 0.102 | 0.0740 |
| $D_{w, \text{primary}}$ (Gy/primary) | 1.79E-15 | 1.27E-15 | 1.16E-15 |
| Derived Source Activity (Bq) | 1.55E+14 | 8.45E+13 | 6.76E+13 |
| Derived Source Activity (Ci) | 4190 | 2280 | 1830 |
| **Average activity of the latter two values (Ci)** | | | 2060 |

The source activities, as derived from data at Position 1, 2 and 3, are 4190, 2280 and 1830 Curies, respectively. The latter two values are in acceptable agreement with each other (~20% difference), but the source activity derived from Position 1 does not agree with that derived from the other two positions.
To examine the discrepancy, a hand calculation was done to determine the effective activity, $A_E$, of the sources, which would provide a ballpark estimate for the true source activity. The effective activity, assuming only finite-length line sources and without considering scatter radiation, could be determined using the following relationships:

$$A_E = \frac{\Gamma G_L (\theta_1 + \theta_2)}{d} \quad (3-3)$$

$$\bar{\chi} = 2G_L h \quad (3-4)$$

where $\bar{\chi}$ is half of measured exposure rate (representing exposure from each of the two sources) at the measurement point (each of the reference dosimetry positions), $\Gamma$ is the gamma-ray constant of the 662 keV emission from a Cs-137 source, which is 58.33 R cm²/Cihr, $G_L$ is the activity per unit length, $d$ is the perpendicular distance between the source plane and the point of measurement, and $\theta_1$ and $\theta_2$ are angles shown in Appendix A, and $h$ is the length of each source.

The hand-calculated effective activities, using exposure rate measurements at the three reference dosimetry positions, are shown in Table 3.2. The effective activities derived from the three reference positions agree with each other, values ranging between 2240 and 2340 Curies, which reinforce the validity of the true source activities derived from Positions 2 and 3, of 2280 and 1830 Curies, respectively. Thus, the most logical estimate of the source activity is the average of the two values, 2060 Curies.
Table 3.2 Effective source activities from hand calculations using exposure rate measured at each of the three reference dosimetry positions

<table>
<thead>
<tr>
<th>Positions</th>
<th>1</th>
<th>2</th>
<th>3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Effective Source Activity (Ci)</td>
<td>2240</td>
<td>2350</td>
<td>2340</td>
</tr>
</tbody>
</table>

The large discrepancy in the Position 1-derived true source activity is obviously a problem to be solved, which may be related to one critical assumption in our Monte Carlo modeling. In our FLUKA code, the geometrical regions of the two sources were modeled to be filled with CsCl powder with even distribution; however, this may not be true in reality. An un-even distribution of source material would more likely result in abnormal dosimetric behaviors at measurement points closer to the source rather than farther away from the source. Moreover, another modeling assumption was that the two sources were assumed to have equal activities – this assumption was in fact justified by a radiochromic film study by Brady et al, which showed symmetric dosimetric distributions on regions closer to the two sources.

Finally, for further investigation, more reference dosimetry positions should be taken in order to determine the true source activity with better accuracy. Another radiation
dosimetric measurement device (one that is independent of ion chamber calibration) can also be applied to ensure the accuracy of the ion camber measurements.

### 3.2 Conclusion

The discrepancy in the results derived at different positions in the irradiator suggested that the manufacturer-provided information on the irradiator may be incomplete. Without accurate specifications, our approach is a practical way for source activity estimation, which can subsequently be used for additional Monte Carlo simulations of the irradiator.
Appendix

Graphical display of geometry used in the hand-calculation equation.
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