Accurate Surface Dose Measurements in CT Examinations Using High Sensitivity MOSFET Dosimeters Calibrated by Monte Carlo Simulations

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Jan. 20 2006

A thesis submitted to McGill University in partial fulfilment of the requirements of the degree of Master of Science in Medical Radiation Physics

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ACKNOWLEDGEMENTS

I would first like to acknowledge my supervisor Dr. Frank Verhaegen for his guidance, support and availability for discussions at all stages of this project. I want to thank my co-supervisor Gyorgy Hegyi for his numerous advices related to measurements in CT.

I would like to thank Dr. Podgorsak for giving me the opportunity to study medical physics at McGill University in such a stimulating environment. Among the teachers and members of the staff, I would especially like to acknowledge Dr. Jan Seuntjens for his help and advice on kilovolt dosimetry and Monte Carlo issues, Dr. Slobodan Devic for introducing me to film dosimetry and Dr. Wamied Abdel-Rahman for helping me solve many of the Monte Carlo and other computer related problems I encountered. I am also thankful to Joe Larkin for his help with the CT scanner, Pierre Leger and Nagy Sharoubim for many helpful discussions on MOSFET issues.

I want to thank Abdelbasset Hallil at Thomson & Nielsen Electronics Ltd. who kindly provided the MOSFETs used in this work, as well as many technical advices.

I would like to thank all the students who contributed to make these last two years very enjoyable despite the hard work! A special thanks goes to Genevieve Jarry for her help on various Monte Carlo and programming issues, and Emily Heath for being there every time my computer gave me problems.

I also want to thank my family for their encouragement, and Sarah for her understanding, and for always being there for me.

Finally, I would like to acknowledge financial support received from the Ministère de la Santé et des Services Sociaux du Québec.
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ABSTRACT

The objective of this work is to use MOSFET dosimeters to accurately measure surface dose delivered during CT examinations in various scanning conditions. To achieve this, the behaviour of MOSFETs under kilovoltage x-ray irradiation first needed to be investigated. A dose-to-dose reproducibility of 4.5%, and a mean change in sensitivity response of 10.4% with accumulated dose were measured. A Monte Carlo model of the x-ray source of a PQ5000 CT simulator was built and validated in order to investigate the MOSFET response characteristics and perform dose calculations. An over-response of 10% was observed when the beam energy was decreased from 140 to 80 kVp, and a slight anisotropy of 8.5% from the mean value over 360° was observed. The dosimeters were calibrated on a solid water phantom using a method involving MC surface dose calculations. Good agreement was found between measurements and simulations of surface dose on a cylindrical PMMA phantom for a stationary tube technique, single axial scan and multiple contiguous axial scans, with generally less than 7.5% discrepancies. Film and MOSFET measurements were then performed for helical adult brain scan parameters using different pitch and collimator settings. The use of five MOSFETs combined in a linear array was found to be suitable to accurately measure surface dose in helical scans for almost all pitch and collimation combinations.
Le but de ce travail de recherche est de mesurer la dose de surface délivrée lors d'examen de tomodensitométrie à l'aide de dosimètre de type MOSFET. Afin d'accomplir cet objectif, le comportement des MOSFETs irradiés à l'aide de rayons X d'énergie se situant dans le domaine des kilovolt a d'abord été étudié. Une reproductibilité de 4.5% de la réponse face à une dose donnée, et une variation moyenne de 10.4% de la réponse avec l'accumulation de dose ont été mesurée. La réponse des dosimètres a également été étudiée à l'aide de simulations Monte Carlo. Pour ce faire, un modèle de la source de rayons X d'un tomodensitomètre PQ5000 a été construit et validé afin de permettre d'effectuer des calculs de dose. Une variation de la réponse avec le changement d'énergie de 10%, de même qu'une légère anisotropie de 8.5% de la réponse moyenne pour une rotation de 360° a été observée. Les dosimètres ont été calibrés sur un fantôme d'eau solide par une méthode impliquant des calculs MC de doses de surface. Les différences observées étaient généralement inférieures à 7.5% entre les mesures et les simulations MC de dose de surface sur un fantôme cylindrique de PMMA pour une technique d'exposition où le tube à rayon X est fixe, un balayage axial simple de même que pour des balayages axiaux multiples et contigus. Des mesures à l'aide de films et de MOSFETs ont finalement été prises pour des paramètres d'examens de cerveau adulte utilisant un balayage hélicoïdal et différentes combinaisons d'ouverture de collimateur de pitch. L'utilisation de cinq MOSFETs combiné en une maille linéaire s'est avérée approprié pour mesurer efficacement la dose de surface lorsque le balayage hélicoïdal était employé, et ce, pour presque toutes les combinaisons d'ouverture de collimateur et de pitch disponibles.
Chapter 1 Overview of Computed Tomography Dosimetry

1.1 Introduction

When it was first commercially introduced in radiology in the early 1970's, computed tomography (CT) was truly revolutionary as it was the first fully digital imaging device, even though it was not even close to conventional imaging system in terms of image quality. However, improvement in CT technology has made it at least ten times more sensitive to change in tissue x-ray attenuation than films. It is even possible to obtain almost real-time images with the most advanced multiple-detector row CT scanners.

The number of CT examinations increased throughout the years as the clinical benefits of CT became more and more evident. In fact, a radiation dose survey conducted in 1989 in the United Kingdom found that the proportion of CT examinations represented only 2% of all the medical x-ray examinations, contributing to approximately 20% to the collective effective dose due to x-ray diagnostic procedures (Shrimpton et al., 1991). Fourteen years later, in 2003, another dose survey in the United Kingdom found that the proportion of CT examinations had reached 9% of all x-ray examinations and that CT's contribution to the collective effective dose had more than double to reach about 47% (Shrimpton et al., 2005).

The increasing use of CT combined to the resulting increase of radiation exposure to population, especially to children, has been of concern to the medical community (Mayo et al., 2003), and efforts are still being made to develop and improve means to evaluate CT dose. A better knowledge of the dose delivered to patients in various scanning conditions will lead to a better understanding of the real risks associated with such examinations.
1.2 Past and Current Practices in CT Dosimetry

1.2.1 Conventional Experimental Methods

The conditions of exposure during CT examinations are very different from those in conventional x-ray procedures (highly collimated beam, rotating tube), thus specific techniques were developed to assess patient dose from CT (Jessen et al., 1999). Head and body polymethyl methacrylate (PMMA) phantoms specially designed for CT dosimetry are usually used to measure dose descriptors. The head and body phantoms are cylinders of respectively 16 cm and 32 cm diameter for a length of 15 cm.

The Computed Tomography Dose Index (CTDI) was one of the first dose descriptor developed for CT procedure (Shope et al., 1981) and is defined for a single-detector row scanner as:

\[ CTDI = \frac{1}{T} \int_{-\infty}^{\infty} D(z) \, dz \quad \text{(mGy)} \quad (1.1) \]

where \( T \) is the slice thickness and \( D(z) \) is the dose at point \( z \) on any line parallel to the \( z \) axis of rotation for a single scan. To account for the effect from multiple scans encountered in typical CT examinations, the Multiple Scan Average Dose (MSAD) that corresponds to the average dose resulting from a series of scans over an interval of length \( I \) was developed and is defined:

\[ MSAD = \frac{1}{I} \int_{-l/2}^{l/2} D_{\text{series}}(z) \, dz \quad \text{(mGy)} \quad (1.2) \]

where \( I \) is the constant distance separating each scans and \( D_{\text{series}}(z) \) is the dose at position \( z \) parallel to the rotational \( z \) axis resulting from the series of CT scans (McNitt-Gray, 2002). For a series of scan such that the first and the last slice of the series do not contribute significantly to the dose over the width of the central slice, Shope et al. (1981) showed that \( MSAD \) could be conveniently derived from \( CTDI \) as:
\[ MSAD = \frac{T}{I} CTDI \] (mGy) \hspace{1cm} (1.3)

where \( T \) is the slice thickness and \( I \) the increment between successive slices.

The Food and Drug Administration (FDA) suggested a particular version of \( CTDI \) in 1984, \( CTDI_{FDA} \), defined as the radiation dose corresponding to 14 contiguous sections, normalized to the beam width:

\[ CTDI_{FDA} = \frac{1}{T} \int_{-T}^{T} D(z) \, d(z) \, (mGy) \] \hspace{1cm} (1.4)

where \( T \) is the slice thickness and \( D(z) \), as defined in equation 1.1, is the dose at point \( z \) on any line parallel to the \( z \) axis of rotation for a single scan. In the United States, manufacturers must report such dose data to the FDA for every type of scanner on the basis of measurements in standard dosimetry phantoms that are expressed in terms of absorbed dose to PMMA (Jessen et al., 1999). According to the definition, \( CTDI_{FDA} \) can only be measured for a maximum slice width of 7 mm with the 100 mm long pencil ionization chamber commercially available. For larger slice width, the radiation dose profile corresponding to the length of the 14 sections have to be measured with TLDs or film, which is not very convenient. To overcome the limitations and difficulties introduced by the FDA definition, the \( CTDI_{100} \) radiation dose index was developed (Leitz et al., 1995). It doesn't have the 14 sections constraints of \( CTDI_{FDA} \) and can easily be measured with a 100 mm pencil ionization chamber. \( CTDI_{100} \), expressed in terms of absorbed dose to air, can be measured in air or in phantom and is defined as:

\[ CTDI_{100} = \frac{1}{T} \int_{-5 \text{ cm}}^{5 \text{ cm}} D(z) \, dz \, (mGy) \] \hspace{1cm} (1.5)

\( CTDI_{100} \) can also be expressed as:
where $T$ is the slice thickness, $f$ is a conversion factor from exposure to dose in air approximately equal to 8.76 mGy/R for low energy x-rays, $c$ is the electrometer calibration factor, $E$ is the measured exposure value in Roentgens acquired for the single scan and $L$ is the active length of the pencil ion chamber (McNitt-Gray, 2002). The CT dosimetric phantoms allow measurement of the $CTD_{100}$ index at the center and at 1 cm below the surface at four different positions at the periphery. Due to attenuation by the phantom, the dose index is clearly dependent on the position and depth at which it is measured inside the phantoms within the scan plane.

In its 1990 recommendations, the International Commission on Radiological Protection (ICRP) introduced the concept of dose constraints for the patient, which contributed towards the introduction of diagnostic reference dose levels for application in relation to typical practice for x-ray departments (Jessen et al., 1999). In 1997, a Working Group of the Commission of the European Communities (CEC, 1997) proposed to use a weighted dose index that would provide an indication of the average dose over a single slice for each setting of nominal slice thickness. They introduced the weighted CT dose index normalized to unit radiographic exposure, $\,_{w}CTDI$, based on the measurement of center and peripheral $CTD_{100}$ values:

$$
\,_{w}CTDI = \frac{1}{C} \left( \frac{1}{3} CTDI_{100,c} + \frac{2}{3} CTDI_{100,p} \right) \quad \text{(mGy/mAs)} \quad (1.7)
$$

where $CTDI_{100,c}$ is the value of $CTD_{100}$ measured at the center of the phantom, $CTDI_{100,p}$ is the average value of $CTD_{100}$ measured at 1 cm below the surface at the four peripheral positions and $C$ is the radiographic exposure (mAs). Based on this new CT dose descriptor, the CEC then proposed two well defined reference dose quantities: the
weighted CT dose index \( \text{CTDI}_w \) for a single slice, and the dose-length-product \( \text{DLP} \) for a complete examination. The first reference dose quantity, \( \text{CTDI}_w \), is defined as:

\[
\text{CTDI}_w = \sum_i \text{CTDI}_w \ C \quad \text{(mGy)}
\]  \( \text{(1.8)} \)

where \( C \) and \( \text{CTDI}_w \) are defined as in equation 1.7. The second reference dose quantity, \( \text{DLP} \), for a complete examination in axial mode is given by:

\[
\text{DLP} = \sum_i \text{CTDI}_w \ T \ N \ C \quad \text{(mGy cm)}
\]  \( \text{(1.9)} \)

where \( i \) represents each scan sequence forming part of an examination, \( N \) is the number of slices of thickness \( T \) (cm), and \( C \) (mAs) is the radiographic exposure in a particular sequence. For a scan in helical mode, equation 1.9 becomes:

\[
\text{DLP} = \sum_i \text{CTDI}_w \ T \ A \ t \quad \text{(mGy cm)}
\]  \( \text{(1.10)} \)

where for each \( i \) helical sequence forming part of an examination, \( T \) is the nominal slice thickness (cm), \( A \) is the tube current (mA) and \( t \) is the total acquisition time.

Reference dose values for \( \text{CTDI}_w \) and \( \text{DLP} \) were determined for various examination conditions. These values are intended to help identify potentially poor dosimetric performance of a scanner but can hardly be related to assess potential risks to patient from radiation exposure. In 1991, the ICRP introduced a new parameter called effective dose \( (E) \) in its publication 60 to take into consideration that the probability of stochastic effects not only depends on the amount of dose absorbed, or equivalent dose, but also on the tissue or organ irradiated. Effective dose is defined as:

\[
E = \sum_T w_T H_T \quad \text{(mSv)}
\]  \( \text{(1.11)} \)

where \( H_T \) is the equivalent dose in tissue or organ \( T \) and \( w_T \) is the tissue weighting factor for tissue \( T \) that represents the relative contribution
of that organ or tissue to the total detriment due to the effects resulting from uniform irradiation of the whole body. Effective dose can therefore serve as a mean to inter-compare whole body detriment associated with different types of localized examinations.

A method based on measured $DLP$ was developed to give broad estimates of effective dose associated with various types of examinations (Jessen et al., 1999):

$$E = E_{DLP} \times DLP \quad {\text{(mSv)}}$$

(1.12)

where $DLP$ is the dose-length-product defined by equations 1.9 and 1.10, and $E_{DLP}$ are constant values associated with general body parts. For example, $E_{DLP}$ for the head is equal to 0.0021 mSv mGy$^{-1}$ cm$^{-1}$, and equal to 0.014 mSv mGy$^{-1}$ cm$^{-1}$ for the chest.

1.2.2 Monte Carlo (MC) Based Methods

Conventional CT dosimetry methods based on cylindrical phantom measurements are mostly used as part of quality assurance programs. In UK for example, a vast study on CT practice throughout the country in 2003 has allowed the National Radiological Protection Board (NRPB) in the UK (Shrimpton et al., 2005) to note that the overall levels of CT exposure were in general 10-40% lower than previous survey data from 1991, even though wide variations were still apparent between medical centres. These methods have no means to assess the actual radiation doses delivered to specific organs, nor evaluating patient specific doses and thus properly evaluate potential risks. Monte Carlo dose simulations offer a powerful tool in attempting to reach this objective and many approaches were studied.

Shrimpton et al. in 1991 were one of the first groups to conduct a Monte Carlo study dedicated to dose delivered in CT exams. They have estimated patient doses for 27 common types of CT scanner for 18 general types of axial examinations based on dose calculations performed in the MIRD V phantom, a mathematical hermaphrodite patient model.
where organs are represented by geometrical shapes (McNitt-Gray, 2002).

In order to make organ dose calculations more accurate, there is a need to use detailed voxelized representation of the human anatomy constructed from actual CT data (Caon et al., 1997). In 1997, Caon et al. used the EGS4 MC code to simulate dose resulting from axial scans in voxel-based computational models of head and chest CT dosimetry phantoms. The simulated dose results for both head and chest phantoms agreed to within 7% with measurements performed with an ion chamber. Two years later, the same group used the tomographic model of a 14 years old female torso (ADELAIDE) constructed from fifty-four consecutive CT scans to calculated average absorbed dose to organs (Caon et al., 1999). They simulated axial CT examinations of the chest and abdomen and calculated the average absorbed doses to 13 different organs. They also compared calculated effective dose results with previous results obtained by Shrimpton et al. (1991) for the MIRD V phantom and found up to 30% discrepancies.

More recently, Jarry et al. (2003) modified the general-purpose MC transport code MCNP4B to simulate complete axial and helical scans. Parameters such as slice thickness and pitch could be selected as desired. They validated their CT model and calculated normalization factors by comparing measured and simulated $CTD_{100}$ in air and in phantom for various energies. They used the mathematical phantom MIRD V and a voxelized phantom derived from CT data to estimate relative and absorbed dose to various organs resulting from axial and helical scans. They found significant discrepancies between organ doses simulated with the two phantoms, possibly due to differences in organ size and mass as well as differences in x-ray attenuation in the phantoms media.
1.3 Proposed work

1.3.1 Motivations

The constant increase in number of CT examinations in medical centres throughout the world combined to the fact that doses delivered during CT are amongst the highest of all diagnostic imaging procedures require close monitoring of dose delivered to patients. The conventional dose indexes used in CT dosimetry are helpful when it comes to evaluate performances of a scanner with respect to accepted reference values but cannot be used to monitor actual patient dose. Furthermore, recent studies have shown that doses measured with these methods were up to 20-30% lower than the actual dose (Dixon 2003; Nakonechny et al., 2005). In fact, the main assumption behind the measurement of the CT dose indexes presented in section 1.2.1 is that most of the radiation profile is contained within the active length of the 100 mm pencil chamber. Evidences have shown this might not be true even for the thinner slices, let aside the very large slices used nowadays with the emergence of the multi-detector row scanners. It was suggested that measurement of such indexes be performed in longer phantoms in order to contain all the profile tail scattered radiation (Nakonechny et al., 2005) using a small volume chamber (Dixon 2003; Nakonechny et al., 2005).

Monte Carlo based methods offer ways to accurately evaluate dose to various organs, but they are time consuming and allow calculations under predefined scan conditions and phantom geometry. Under the same scanning conditions, changing the phantom geometry and size also changes the dose deposited. Thus, evaluating dose for every patient scanned would required having patient specific voxelized phantoms, which is probably too much time consuming for most clinics at the moment. Although offering great insight at quantities such as organ dose and effective dose that could not be otherwise measured directly on patients, MC does not allow fast and easy specific patient dose monitoring.
1.3.2 Surface Dose Versus Internal Dose

It is well known that depth dose profiles for kilovolt x-rays fall rapidly due to attenuation by the absorbing media. It is also a known fact that unlike for mega-voltage beams where depth dose profiles are characterized by a region of dose build-up of a few cm near the surface, the point of maximum dose is right at the surface, or within a few mm, for kilovoltage beams. For these reasons, the surface of a phantom or a patient is exposed to higher doses than internal parts in diagnostic procedures involving kilovolt x-rays like CT.

1.3.3 Objectives of the Study

TLDs have been used in many studies to evaluate surface dose in CT (Adams et al., 1997; Nawfel et al., 2000; Hirata et al., 2005). Their small size makes them very appropriate to evaluate surface dose but they are cumbersome and the preparation required for their proper use is time consuming. In this study, a simple means of monitoring surface dose in CT examinations using MOSFET-type dosimeters calibrated by Monte Carlo simulations is proposed. MOSFETs are small, easy to handle and unlike TLDs, allow almost instantaneous readout. In the first part of the study, the Monte Carlo model of a Picker PQ5000 CT simulator is evaluated. In the second part, MOSFET dosimeters characteristics are evaluated. Measurements and simulations are used to evaluate reproducibility, change in response with accumulated dose, angular and energy response of the dosimeters. The third part of the study is dedicated to surface dose evaluation in various scanning conditions of a cylindrical PMMA head phantom. Finally, a protocol is proposed for routine use of MOSFETs in monitoring patient surface dose in clinical CT examinations.
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Chapter 2 Computed Tomography Imaging

2.1 Generations of CT scanner

The basic principle behind CT imaging is that internal structure of an object can be reconstructed from multiple projections of the object and was first proved in 1917 by the Austrian mathematician J. Radon (Curry III et al., 1990). It took more than 50 years before G.N. Hounsfield could turn this relatively simple mathematical statement into a practical application in radiology in 1972. Design of CT scanners has evolved since then and scanners are now classified in four generations based on geometrical characteristics.

2.1.1 First Generation

The first CT scanner was introduced in radiology 1972 by G.N. Hounsfield, a senior research scientist at EMI Limited in Middlesex in England (Curry III et al., 1990). CT technology represented an enormous step forward in the world of diagnostic imaging: CT cross-sectional tomographic images eliminated the superposition of structures that occurred in plane film imaging due to the compression of three-dimensional structures onto two-dimensions recording system. This first generation scanner used an x-ray pencil beam and a single detector describing a translate-rotate motion around the patient (see Figure 2.1). About five minutes were required to acquire a single slice. Significant motion artefacts resulted of the long scanning time. Thus, the first generation scanners could only be used for head and neck scans, where good immobilization can be achieved.
2.1.2 Second Generation

The long scanning time was the major drawback of the first generation scanners. To increase speed acquisition, second generation scanners were equipped with a fan-shaped beam and a linear array of detectors, but still describing a translate-rotate motion like first generation machines (see Figure 2.2). The acquisition time for a single tomographic section could then be reduced to 10 to 20 seconds, depending on the fan beam angle used by the manufacturer.
2.1.3 Third Generation

Third generation, introduced in 1975 by GE, represents the most common design of modern scanners. It uses a wider fan beam encompassing the whole subject to be imaged (see Figure 2.3). The corresponding array of detectors (between 500 to 1000) is no longer linear but shaped along the arc of a circle whose center corresponds approximately with the x-ray tube focal spot (Curry III et al., 1990). With this new x-ray beam and detectors configuration, the translation motion was completely removed. The source and detectors are always perfectly aligned and describe a rotating motion around the patient. The third generation design, also termed rotate-rotate design, allowed decreasing the scanning time by at least a factor of two, down to around two seconds. The very high rotation speed required to perform such fast scans results in tremendous acceleration (as high as 13 G) applied to the x-ray tube and detectors and have caused stability problems in the early third generation scanners, and still continue to represents a technical challenge today.

![Figure 2.3: Schematic diagram of a third generation CT scanner type](reproduced from Mahesh, 2002)

2.1.4 Fourth Generation

The fourth generation scanners are equipped with a complete ring of fixed detectors (see Figure 2.4). The ring is fixed and uses 2000, and
even up to 4800 detectors. Using the combined technologies of slip ring and brush, the tube alone is allowed to continuously rotate around the patient to acquire images in this rotate-fixed design. Having only the tube moving, stability requirements for the detectors are considerably less than for the third generation scanners. On the other hand, the large amount of detectors required to build the ring, from 2000 to 4800 detectors, directly results in an increase of the cost of the fourth generation over the third generation scanners. Scan time is comparable with third generation scanners with around 2 seconds per axial slice. Ring artefacts caused by malfunctioning detectors being continuously exposed in third generation scanners are avoided in the fourth generation where different detectors are exposed as the tube rotates around the ring. On the other hand, fourth generation scanners are more susceptible to scatter artefacts than third generation scanners and therefore require more efficient object scatter correction algorithm to obtain acceptable image quality. In fact, unlike third generation scanners that use effective anti-scatter collimator sheets placed at the detector array to limit the amount of scattered radiation within the patient to reach the detectors, fourth generation scanners can only use collimator sheets to a very limited extent as not only scattered but also significant parts of attenuated primary radiation would be shielded (Ohnesorge et al., 1999).

Figure 2.4: Schematic diagram of a fourth generation CT scanner type (reproduced from Mahesh, 2002).
2.2 The Picker PQ5000 CT Simulator

The scanner used in this study is a Picker PQ5000 computed tomography simulator (Royal Philips Electronics, Eindhoven, Netherlands), a fourth generation scanner equipped with a ring of 4 800 solid state detectors, that allows scanning in both axial and helical mode. An important parameter the user must select in helical scan mode is the pitch ($P$), defined as:

$$P = \frac{I}{T}$$

where $I$ is the table travel in one 360° rotation of the tube and $T$ is the total collimated width of the x-ray beam (McNitt-Gray, 2002). Pitch is dimensionless and ranges from 0.25 to 2.0 in 0.25 increments on the scanner. It also allows single shot exposure using a stationary tube technique when operated in service mode. It is possible to select nominal slice thicknesses of 1.5, 2, 3, 4, 5, 8 and 10 mm at isocenter. It is equipped with a DUNLEE Rhino 6.5 x-ray tube (DUNLEE, Illinois, USA) that permits selection of tube potentials ranging from 80 to 140 kVp in 10 kVp increments, and tube current of 30, 50, 65 and 100 up to 300 mA in 25 mA increments. It has a 7-degree angle tungsten-rhenium anode with possibility of selecting a small and large focal spot of 0.4 x 0.7 mm² and 0.6 x 1.3 mm² respectively. The specified inherent filtration is equivalent to 2.5 mm Al. Additional filtration of 4.5 and 9.0 mm of aluminium are available to users. Head and body compensators, also called bowtie filters, can be selected to compensate for non-uniform tissue thickness across the various exposed parts of the body, thus achieving more uniform x-ray intensity across the fan beam at the detector level. The scanner has a 70 mm gantry opening and a source-to-isocenter distance of 62.75 cm.

2.3 Doses in CT and other Imaging Procedures

The probability of stochastic effects resulting from x-ray irradiation of localized parts of the body can be assessed with the quantity effective
dose $E$ (ICRP 60, 1991). Therefore, effective dose is the quantity of choice when trying to compare the risks associated with various imaging procedures. The effective dose associated with four groups of diagnostic procedure as well as the relative importance in terms of total number of procedures and their contribution to the total effective collective dose is presented in table 2.1.

Table 2.1: X-ray imaging procedures divided into four dose bands*

<table>
<thead>
<tr>
<th>Effective Dose Range (mSv)</th>
<th>Typical X-Ray Procedures</th>
<th>Percentage of Total Number of Procedures</th>
<th>Percentage of Total Collective Effective Dose</th>
</tr>
</thead>
<tbody>
<tr>
<td>&lt; 0.02</td>
<td>Radiography of chest, limbs and theeth</td>
<td>73</td>
<td>1</td>
</tr>
<tr>
<td>0.02-0.2</td>
<td>Radiography of head, neck and joints</td>
<td>5</td>
<td>1</td>
</tr>
<tr>
<td>0.2-2.0</td>
<td>Radiography of spine, abdomen and pelvis</td>
<td>16</td>
<td>20</td>
</tr>
<tr>
<td>2.0-20</td>
<td>CT, angiography, contrast studies of the GI and urinary tracts</td>
<td>6</td>
<td>78</td>
</tr>
</tbody>
</table>

* Table obtained from a study by Wall, 2004

These are summarized statistics resulting from a vast study performed in 1998 by the NRPB in the UK concerning all medical x-ray imaging (Hart and Wall, 2002). While high dose procedures like CT only account for 6% of the total number of examinations, they are responsible for 78% of the collective dose, from what the necessity comes of closely monitoring dose to patients in these examinations, as discussed in chapter 1.

The variation of effective dose levels observed between different imaging procedures is also observed within various CT examinations as shown in table 2.2. The 2003 UK radiation dose survey results show that effective dose can vary by almost a factor of 10 within CT examinations. The wide range of effective dose observed results from the different levels of radio-sensitivity of organs irradiated during each procedure, but also on
the variability of exposure techniques required to obtain good image quality of the different parts of the body.

Table 2.2: Typical effective doses delivered to patient during CT examinations*

<table>
<thead>
<tr>
<th>Examination</th>
<th>Effective Dose mSv</th>
</tr>
</thead>
<tbody>
<tr>
<td>Routine head (acute stroke)</td>
<td>1.5</td>
</tr>
<tr>
<td>Abdomen (liver metastases)</td>
<td>5.3</td>
</tr>
<tr>
<td>Abdomen &amp; pelvis (abscess)</td>
<td>7.1</td>
</tr>
<tr>
<td>Chest, abdomen &amp; pelvis (lymphoma staging or follow up)</td>
<td>9.9</td>
</tr>
<tr>
<td>Chest (lung cancer: known, suspected or metastases)</td>
<td>5.8</td>
</tr>
<tr>
<td>Chest: Hi-resolution (diffuse lung disease)</td>
<td>1.2</td>
</tr>
</tbody>
</table>

*Table obtained from NRPB report W67 (Shrimpton et al., 2005)

2.4 Surface Dose in CT

In 1997, a group at the North Western Medical Physics, Christie Hospital in Manchester, UK, used TLDs to measure surface dose on 24 patients undergoing routine and pelvis CT scan (Adams et al., 1997). The scans consisted of 10 contiguous axial slices of 10 mm width. They calculated the average dose measured by one TLD positioned on the anterior/posterior surface and one on the lateral surface. The mean surface doses measured range from 50 to 80 mGy.

A study was conducted in 2000 in the Department of Radiology at the Brigham and Women’s Hospital in Boston, USA, on patient and personnel dose during CT fluoroscopy-guided interventional procedures. Nawfel et al. (2000) estimated surface dose resulting from such procedures using TLD placed on cylindrical phantom. They measured surface dose rates ranging from 2.3 to 10.4 mGy/sec. Considering the fact that these examinations can last from seconds to a few minutes, total surface doses to patients can reach as high as 2,000 mGy for a 3 minutes fluoroscopic intervention, which is high enough to initiate deterministic effects such as transient erythema.

More recently, a group from Japan (Hirata et al., 2005) placed TLDs on a human head phantom to measure surface doses in cerebral CT
perfusion, a very high dose procedure used to image blood-flow in the brain. The examination consisted of continuous cine scans (1 sec/revolution x 60 sec) of four 5-mm thick contiguous slices. Various combinations of tube potential and current were used. The lowest dose measured was 162.5 mGy at 80 kVp and 50 mA, and the highest dose measured was 2 264.6 mGy at 140 kVp and 200 mA. Clearly, the exposure settings used have very large impact on the dose delivered to patients.

2.5 Factors Influencing Radiation Dose in CT

2.5.1 Beam Energy

The energy of the beam has direct influence on the patient radiation dose. The energy of an x-ray beam is determined mostly by the tube potential selected (kVp), but also by the filtration used (McNitt-Gray, 2002). Dose in air is typically proportional to the exposure, which is typically proportional to kVp². Therefore, increasing the kVp from 120 to 140 would lead to an increase in dose to air of about 36%.

2.5.2 Photon Fluence

The photon fluence is the amount of photon crossing a given area and is directly related to the current-time product, or mAs. Since dose is directly proportional to the photon fluence, it increases linearly with the mAs selected. To maintain dose as low as possible in CT procedures, it is of primary importance to minimize protocol exposure parameters (kVp and mAs) while maintaining acceptable quality of diagnostic images.

2.5.3 Helical Scan Pitch

In helical scan procedures, the distance between directly exposed parts of a phantom is determined by a parameter known as pitch that was defined in equation 2.1. For example, a pitch of one will correspond to a 10 mm table increment per tube rotation combined with a 10 mm nominal
slice thickness. Similarly, a nominal slice thickness of 5 mm combined with a 10 mm table increment would correspond to a pitch of 2.

McNitt-Gray et al. (1999) measured \( CTDI_{vol} \) values with TLDs. \( CTDI_{vol} \) is simply defined as:

\[
CTDI_{vol} = \frac{CTDI_w}{P}
\]  

(2.2)

where \( CTDI_w \) is the weighted CT dose index that was defined previously in chapter 1 (see eq. 1.8), and \( P \) is the pitch. It is the only CT dose index that takes pitch into account. Their results have shown that radiation dose for a helical scan is approximately proportional to \( 1/pitch \). Therefore, to achieve lower dose, larger pitch are preferable, although this comes to the cost of an increase in volume averaging during image reconstruction which might in term results in reduced object contrast (McNitt-Gray, 2002).

2.5.4 X-Ray Beam Collimation

In their study of 1999, McNitt-Gray et al. also investigated the effect of collimation on dose measured in phantom. They found that collimation only has a small effect on radiation dose for both axial and helical scans. In fact, dose measured using 1, 3, and 5 mm nominal slice thicknesses were all within 15% of the dose measured using 10 mm slice thickness.

2.5.5 Patient or Phantom Size

In a recent study, Huda et al. (2004) have investigated the effect of patient size on radiation doses delivered in CT scan of the head. They based their study on 127 patients CT scan data and divided the patients per age group. A \( H_2O \) equivalent head radius \( r \) was calculated for each group and then every group of patients was modelled as a uniform cylinder of water of radius \( r \). The dosimetric quantity under study were the mean section dose \( D_m \), obtained for a single CT section by dividing the total energy absorbed in a cylindrical water phantom by the mass of the
directly irradiated section, the energy imparted \( E \) to a single section of water phantom and effective dose \( E \). They found that for the same exposure parameters, the mean section doses were 40% higher in infants (0-6 months) than in adults (>18 year) due to reduced attenuation resulting in higher exit dose, in smaller patients. While energy imparted was 50% higher in adults, mostly due to the larger mass of the head, effective dose was 4 times higher in infants. This last observation can be explained by the fact that radiosensitive organs considered in the calculation of effective dose are much closer to the site of primary irradiation for infants, which means these organs are exposed to more scattered radiation. The smaller sizes of organs in infants also lead to larger organ doses, resulting in higher overall calculated effective dose.

### 2.5.6 Other Factors

Avilés Lucas et al. (2001, 2004) have studied the variation of surface dose in CT procedures by measuring the variations of air kerma-length product (AKLP), which is simply the \( CTDI_{100} \) dose index measured at the surface of the phantom multiplied by the nominal slice thickness. A schematic diagram of the configuration they used to measure \( AKLP \) can be found in figure 2.5.

![Schematic diagram of the configuration used by Avilés et al. to measure AKLP](reproduced from Avilés et al., 2004).
It allowed them to study the effect of various parameters on AKLP such as patient size, position within the gantry plane and beam-shaping filter type. They showed that AKLP was mostly dependent on the vertical position of the surface of the phantom within the gantry: it is maximum at the isocenter and decreases as the surface of the phantom vertically moves away from it. They showed that this dependence was mostly due to the presence of the bowtie filter. In fact, as the tube rotates around the gantry, the isocenter is the point within the rotation plane where the attenuation due to the filter is constant and minimum. AKLP measured when the surface of the phantom is at the isocenter was 19% higher than the AKLP measured 12 cm away with the large filter in place.

Beam collimation and reconstruction algorithm together can also influence CT dose (McNitt-Gray, 2002). Noise in a reconstructed slice of thickness $T$ typically depends on $T^{-1/2}$. Therefore, for constant kVp and mAs, thinner reconstructed slices are noisier than larger slices. To account for the resulting decrease in image quality, it is sometimes necessary to increase the exposure technique, which results in a higher doses delivered to patients.

Many techniques to reduce CT dose are under development. Among them are techniques based on mAs modulation. To achieve equivalent image quality, thinner parts of the body require less incident photons than thicker parts, and so lower dose could be achieved by modulating the tube output (mAs) with respect to tissue thickness.

In this work, the focus is put on the measurement of localized surface dose with MOSFET dosimeter resulting from CT head scans, as oppose to the various conventional average dose index measured with a pencil-type ion chamber discussed in chapter 1. The effects of parameters such as the beam energy, collimation and helical pitch on the surface dose are investigated.
References


Chapter 3 Kilovoltage X-Rays: Dosimetry, Radiation Detectors and Monte Carlo Simulations

3.1 Dosimetry in Kilovoltage X-Rays

3.1.1 TG-61 Protocol

The AAPM Radiation Therapy Committee Task group 61 was set-up in 1995 to evaluate the different protocols available at that time for reference dosimetry purposes in kilovoltage x-ray beam (Ma et al., 2001). In 2001, the committee presented a new protocol for 40 to 300 kV x-ray beam dosimetry in radiotherapy and radiobiology known as the TG-61 protocol. Two methods were proposed to determine absorbed dose to water. The first method, known as the in-air method, was developed for low and medium energy x-rays (tube potential ranging from 40 to 300 kV) and allows determination of absorbed dose to water at the surface. The second method, known as the in-phantom method, was developed for medium energy x-rays (tube potential ranging from 100 to 300 kV) and allows determination of absorbed dose to water at a depth of 2 cm.

Even though TG-61 was initially designed to serve radiotherapy and radiobiology purposes, the two methods explained can also be applied to any x-ray machine operating in the 40 to 300 kVp range. Computed tomography is a kilovoltage x-ray procedure typically performed at beam energies ranging from 80 to 140 kVp, which falls in the energy range covered by TG-61. This study focuses on 120 and 140 kVp, the most commonly used beam energies in CT exams, therefore, the in-phantom method for medium energy x-rays is presented here.

3.1.2 Ionization Chamber Calibration

Implementation of TG-61 requires the use of a calibrated ionization chamber. The calibration of the chamber is made in terms of air-kerma free in air ($K_{air}$) in a national standards lab for a number of x-ray beam qualities that are ideally not to dissimilar to the beam qualities in which the
chamber is meant to be used. The air-kerma calibration factor of the user's chamber at a specific beam quality is defined as:

$$N_K = \frac{K_{air}}{M} \quad (3.1)$$

where $K_{air}$ is the known air-kerma value at the reference point in air and $M$ is the chamber reading corrected for temperature, pressure, ion recombination, polarity effect and electrometer accuracy. The chamber reference point must be at the same reference point in air where $K_{air}$ is known. In diagnostic radiology dosimetry, the exposure calibration factor $N_x$ is frequently used and can be derived from $N_K$ according to:

$$N_x = \frac{N_K \cdot \frac{W}{e}_{air} \cdot (1-g)}{(3.2)}$$

where $(W/e)_{air}$ is the average energy expended per unit charge of ionization produced in dry air and has the value of 33.97 J/C, and $(1-g)$ corrects for the effect of radiative losses by secondary charged particles. Since $g$, the radiative fraction, is less than 0.1% for photon below 300 keV in air, the $(1-g)$ factor can be neglected at CT energies (Ma et al., 2001).

### 3.1.3 Chamber Reading Correction Factors

The pressure and temperature correction factor $P_{TP}$ accounts for the fact that the amount of charges collected in the chamber depends on the mass of gas present in the cavity, which varies with ambient temperature and pressure conditions for an unsealed ion chamber. Therefore, the amount of charge collected must be corrected to the reference temperature ($T_{ref} = 22 ^\circ C$) and pressure ($P_{ref} = 101.33$ kPa = 760 mm Hg). $P_{TP}$ can be calculated as:

$$P_{TP} = \frac{P_{ref}}{P} \frac{(T_{ref}[\circ C]+273.2)}{(T[\circ C]+273.2)} \quad (3.3)$$
The ion collection efficiency correction factor, $P_{ion}$, accounts for the fact that some ions created in the chamber cavity might recombine with ions of opposite charge and not be collected by the electrode. For continuous beam, $P_{ion}$ is obtained using the two-voltage approach as:

$$P_{ion}(V_H) = 1 - \left(\frac{V_H}{V_L}\right)^2 \frac{M^H_{raw}}{M^L_{raw}} \left(\frac{V_H}{V_L}\right)^2$$  \hspace{1cm} (3.4)

where $M^H_{raw}$ is the raw chamber reading with the normal operating bias voltage $V_H$, usually 300 V, and $M^L_{raw}$ is the raw chamber reading measured using a lower bias voltage $V_L$, usually 150 V (Ma et al., 2001).

The polarity correction factor $P_{pol}$ accounts for possible difference in signal collected when the chamber polarity is reversed and depends on beam quality and cable arrangement. It can be evaluated as:

$$P_{pol} = \frac{M^+_{raw} + M^-_{raw}}{2M_{raw}}$$  \hspace{1cm} (3.5)

where $M^+_{raw}$ is the chamber reading when positive charges are collected, $M^-_{raw}$ is the reading when negative charges are collected, and $M_{raw}$ is the reading corresponding to the charge collected in the reference dosimetry settings (either $M^+_{raw}$ or $M^-_{raw}$) (Ma et al., 2001).

The ionization chamber is usually sent for calibration along with the electrometer it is going to be used with. Correction for electrometer accuracy, $P_{elec}$, is done by the standards lab as part of the chamber calibration process, so the user shall simply consider $P_{elec}$ as being unity.

### 3.1.4 Absorbed Dose to Water: The in-phantom method

Using the in-phantom method, absorbed dose to water at a depth of 2 cm can be determined according to:
\[ D_{w,z=2\text{cm}} = M \cdot N_K \cdot P_{Q,\text{cham}} \cdot P_{\text{sheath}} \cdot \left( \mu_{en}/\rho \right)_{\text{air-water}}^{\text{water}} \]  \hspace{1cm} (3.6)

where \( M \) is the chamber reading, with the center of the sensitive air cavity positioned at the reference depth \( z_{\text{ref}} = 2 \text{ cm} \), corrected for temperature, pressure, ion recombination, polarity effect and electrometer accuracy, and \( N_K \) is the air-kerma calibration factor for the beam quality used. \( P_{Q,\text{cham}} \) is the overall chamber correction factor that accounts for the change in the chamber response due to the displacement of water by the ion chamber and the presence of the chamber stem, the change in the energy and angular distribution of the photon beam inside the phantom compared to that used for the calibration of the chamber in air. If applicable, \( P_{\text{sheath}} \) corrects for photon absorption and scattering in the waterproofing sleeve, and \( \left( \mu_{en}/\rho \right)_{\text{air-water}}^{\text{water}} \) is the ratio of the mean mass energy-absorption coefficient of water-to-air averaged over the photon spectrum at the reference point in water in the absence of the chamber. Values for \( P_{Q,\text{cham}} \), \( P_{\text{sheath}} \) and \( \left( \mu_{en}/\rho \right)_{\text{air-water}}^{\text{water}} \) are tabulated for various chamber types, waterproofing sleeve materials and beam energies in TG-61.

### 3.2 Radiation Dosimeter Behaviour in Kilovoltage X-Rays

The most commonly used detectors to measure radiation are ionization chambers, thermoluminescent dosimeters (TLDs), films and semiconductors. In order to measure dose accurately, the behaviour of these different types of dosimeters under given irradiating conditions must be well understood. The response of a dosimeter is usually greatly affected by changes in x-ray energy, especially when used in the kilovolt energy range. This change in response can be explained by considering the different modes of interaction of x-rays with matter. The total mass attenuation coefficient, \( \mu/\rho \), of photon in air is plotted in figure 3.1. Data
were obtained from the XCOM photon cross-section database (Berger et al., 1998).

![Graph showing photon total mass attenuation coefficient in air for energies ranging from 1 keV to 1 MeV. The contribution to the total mass attenuation coefficient due to photoelectric and Compton effect is also plotted.](image)

Figure 3.1: Photon total mass attenuation coefficient in air for energies ranging from 1 keV to 1 MeV. The contribution to the total mass attenuation coefficient due to photoelectric and Compton effect is also plotted.

For energies larger than 40 keV, photons mostly interact through Compton effect, which causes the total cross section to slowly increase with decreasing energy. Under 40 keV, the photoelectric effect, which varies as $\frac{1}{(h\nu)^3}$, is the predominant mode of interaction and causes a rapid increase of photon interaction with decreasing energy. The regions of predominance of the photoelectric effect and the Compton effect vary with the atomic number $Z$ of the attenuator, but the general trend of the total photon cross section for any medium is typically similar to what can be seen in figure 3.1, and the most important variations are always observed in the medium kilovolt energy range.

The energy response measured for a Farmer-type Exradin A12 and a Farmer 2505/3 ionization chamber is plotted in figure 3.2. The chamber responses were normalized to the response observed for an effective energy of 120 keV. Even though both chamber cavities are filled with air, they exhibit a different behaviour due to differences in the overall geometry and material used. The Farmer 2505/3 response decreases by
4% between 30 and 120 keV, while the maximum change in response for the Exradin A12 is 2.3%.

![Graph](image)

**Figure 3.2**: Measured energy response of the Exradin A12 and Farmer 2505/3 ionization chambers normalized to the response at 120 keV.

Mota et al. (1990) investigated the response of Kodak X-Omat XV-2 radiographic film. They measured the change in net optical density after irradiation under different conditions: 120 kVp (2.5 mm Al HVL), 280 kVp (1.7 mm Cu HVL) and Cobalt-60. Their results are shown in figure 3.3.

![Graph](image)

**Figure 3.3**: Dose-response curves for Kodak X-Omat XV-2 film. The curves were obtained at energies of 120 kVp (2.5 mm Al HVL), 280 kVp (1.7 mm Cu HVL) and Co-60 (data obtained from Mota et al., 1990).
The film response is much higher at lower energy. For example, an accumulated dose of 10 cGy results in a net optical density of around 2.0 at 120 kVp, 1.0 at 280 kVp (1.7 mm Cu HVL) and only around 0.2 at Co-60. Aside from the energy dependence, the film response with accumulated dose can also be seen in figure 3.3. The film response is only linear over a limited dose range: an over-response is observed in the lower dose regions for all beam energy used.

3.3 MOSFET: Metal Oxide Semiconductor Field Effect Transistor

3.3.1 The MOSFET as a Micro-Electronic Device

A Metal Oxide Semiconductor Field Effect Transistor (MOSFET) is a micro-electronic device characterized by a very high input resistance (tera ohm range), very low gate current (pico ampere range), small size and low production cost. They are widely used in digital circuits, analogue circuit signal treatment, microprocessors and amplifiers. The basic structure of a P-channel MOSFET is depicted in figure 3.4.

Two electrodes of the MOSFET, the source and the drain, are situated on top of a positively doped (p-type) silicon region. The third electrode is called the gate and is positioned over a thin insulating silicon dioxide (SiO₂) layer (around 1 micron thick), and underneath this layer is the n-type silicon substrate. A negative voltage with respect to the substrate applied to the gate, the gate voltage (Vgs), causes a significant number of minority carriers, in this case holes, to be attracted to the silicon dioxide-silicon interface from both the n-type silicon substrate and the source and drain regions. A conduction channel is created as the number of holes
accumulates, allowing an appreciable amount of current to flow between the source and the drain, called the drain-source current \(I_{ds}\). The voltage required to establish a predetermined current \(I_{ds}\) is called the threshold voltage \(V_{TH}\).

### 3.3.2 The MOSFET as a Radiation Detector

The use of MOSFET device as radiation detector is based on the fact that ionizing radiation modifies their intrinsic physical properties. When a MOSFET is irradiated by ionizing radiation, electron-hole pairs are created within the sensitive SiO\(_2\) layer (Soubra et al., 1994). Electron mobility at room temperature is about four orders of magnitude greater than holes in SiO\(_2\), so a large fraction of the electrons quickly moves toward the positively biased contacts, whereas the rest of the electrons created will recombine with holes. Depending on the energy and the nature of the radiation and the applied field, a portion of the holes will escape initial recombination with electrons and remain relatively immobile near their point of generation. In response to a positive bias applied to the gate, holes are pulled in a stochastic hopping transport phenomenon toward the Si-SiO\(_2\) interface where some of them are captured in long-term trapping sites. As a result, these trapped positive charges cause a negative threshold voltage shift \((\Delta V_{TH})\), meaning that a larger gate voltage \(V_{gs}\) is required to re-establish the predetermined drain-source current \(I_{ds}\) (see Figure 3.5). The threshold voltage shift can persist for years and can be experimentally measured. Its magnitude is proportional to the total quantity of charges trapped at the Si-SiO\(_2\) interface, which is proportional to the dose deposited. Therefore, measurement of a radiation dose with MOSFET requires linking the voltage shift \(\Delta V_{TH}\), in units of mV, to a dose deposited, in units of mGy, through precise calibration.
Figure 3.5: Negative threshold voltage shift $\Delta V_{TH}$ resulting from irradiation of a MOSFET dosimeter. $V_{TH,l}$ and $V_{TH,f}$ are the initial and final threshold voltages required to attain a drain-source current $I_{ds}$ (reproduced from Soubra et al., 1994).

3.3.3 Characteristics of the High Sensitivity TN-1002RD MOSFET Dosimeter

In this study, high sensitivity MOSFET model TN-1002RD dosimeters (Thomson & Nielsen Electronics Ltd., Nepean, Ontario, Canada) were used. It is well known that not only radiation, but also heat can induce change in voltage threshold. In fact, a 1°C change in ambient temperature can modify the threshold voltage shift $\Delta V_{TH}$ by as much as 4 to 5 mV (Soubra et al., 1994), which would correspond to an error of 4 to 5% on a typical reading of 100 mV obtained at CT energy. In order to overcome this temperature dependence, Thomson & Nielsen has developed a dual bias dual MOSFET dosimeter. For dual MOSFET detectors, the threshold voltage shift measured is given by:

$$\Delta V_{TH} = (\Delta V_{TH}^1 - \Delta V_{TH}^{Temp}) - (\Delta V_{TH}^2 - \Delta V_{TH}^{Temp})$$

$$= \Delta V_{TH}^1 - \Delta V_{TH}^2$$

(3.7)

where $\Delta V_{TH}^{Temp}$ is the threshold voltage shift due to change in temperature and $\Delta V_{TH}^1$ and $\Delta V_{TH}^2$ are the changes due to irradiation in
MOSFET 1 and 2 respectively. Soubra at al. (1994) found a resultant temperature coefficient of less than 0.015 mV/°C.

The sensitivity of a MOSFET dosimeter, which is basically the amount of holes trapped per unit dose deposited, depends on both the thickness of the sensitive SiO₂ layer, which determines the total amount of electron-hole pairs available, and the positive gate bias applied during irradiation, which determines the efficiency of the hole trapping mechanism. Nominal sensitivity stated by the manufacturer ranges from 10 mV/R for the standard bias supply to 30 mV/R for the high bias supply.

The MOSFET exhibit a small non-linearity with accumulated dose. As the number of trapped holes in the SiO₂ layer increases due to accumulating dose, the magnitude of the effective positive gate bias in the oxide is reduced. The resulting decrease in electric field in the sensitive layer leads to an enhanced electron/trapped-holes recombination effect that diminishes the threshold voltage shift measured (Soubra et al., 1994). Therefore, the amount of incident radiation required to measure the same $\Delta V_{TH}$ increases as dose accumulates in the MOSFET. According to the manufacturer, the linearity of the response with accumulated dose is within +/- 5% throughout the 20 000 mV lifespan of the MOSFET. The 20 000 mV lifespan of the dosimeter corresponds to around 18 Gy when used under standard bias supply, and around 6 Gy when used at high bias supply.

Dong et al. (2002) studied the energy dependence of the LiF:Mg:Cu:P TLD-100H dosimeter and the high sensitivity MOSFET TN-1002RDI (Thomson & Nielsen). They evaluated the energy dependence for tube potentials ranging from 40 to 125 kVp, and their results can be seen in figure 3.6. Almost no dependence on energy is observed for the TLD-100H. On the other hand, MOSFET response changes by more than 20% in the same potential range. The behaviour of MOSFET-type dosimeter is discussed in more details in the next section.
The angular response of a MOSFET dosimeter was studied by Roshau and Hintenlang (2003) under diagnostic x-ray energies. They irradiated the standard sensitivity MOSFET model TN-502RD (Thomson & Nielsen) in air at 70 kVp on a conventional radiographic machine. Note that the MOSFET models TN-502RD and TN-1002RD only differ by the thickness of the sensitive oxide layer, so they should have very similar angular response. The MOSFET was rotated in increments of 15° for a full 360° rotation and the response at each angle was normalized to the overall mean reading. They found a nearly isotropic response with a standard deviation of only 2.7% from the mean value and observed no significant deviations from the mean at any of the measured angles.

The threshold voltage shift resulting from x-ray irradiation of a MOSFET has been found to depend on the readout time interval following irradiation. Bower and Hintenlang (1998) observed a rapid rise of the response by about 14%, occurring during the first 26 hours after exposure, followed by a slower but steady decline. No levelling off was observed over a 500 hour period. This post-exposure drift effect is not well

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Figure 3.6: TLD-100H and MOSFET TN-1002RDI energy response expressed as energy dependent correction factors, normalized to that at 90 kVp, and measured for tube potentials ranging from 40 to 125 kVp (data obtained from Dong et al., 2002).
understood but is thought to be attributable to mechanisms such as positive ion diffusion, relaxation of charges residing in deep traps in the SiO$_2$, capacitance-voltage hysteresis, both thermal and non-thermal annealing of trapped charges and enhanced $1/f$ noise (Bower and Hintenlang, 1997). $1/f$ noise is responsible for variations in the measurement given by a single instrument and exhibits a spectral dependence that is inversely proportional to the frequency $f$ of the noise (Benson et al., 2004).

Another effect, known as the creep-up effect or the multiple query effect, can also alter the MOSFET response following irradiation. The action of reading out the threshold voltage shift results in an injection of charges in the MOSFET by the measuring circuit, thus creating a temporary perturbation in the charge distribution that normally decays in a few minutes if no subsequent readout is performed (Ramani et al., 1997). Bower and Hintenlang (1997) quantified this effect by reading out the response of a MOSFET every 30 seconds over a 20 minutes period following exposure. They observed an artificial increase of the response of about 11% after nearly 40 consecutive readouts.

While the post-exposure drift effect and the creep-up effect can lead to significant discrepancies between the dose read and the actual dose deposited, these two effects can be easily overcome by making only one dose readout at a fixed time interval following irradiation.

3.4 Monte Carlo Simulations
3.4.1 Introduction to the EGSnrc Monte Carlo Code

The EGSnrc MC code developed at the National Research Council Canada (NRC) was used for all the simulations performed in this work. It incorporates many different user codes that will further be discussed in more details such as BEAM, BEAMDP, DOSXYZ, FLURZ and g.

In the field of radiation physics, a MC simulation usually refers to the detailed simulation of individual particle interactions and tracks inside a
given geometry of interest. The EGSnrc code can accurately model coherent Rayleigh scattering, incoherent bound Compton scattering events occurring at low energies as well as Klein-Nishina Compton scattering (Kawrakow and Rogers, 2003). Thick target bremsstrahlung photon spectra generated in x-ray tubes can be generated, but only one of the two phenomena responsible for the production of the characteristic x-ray peaks can be effectively modeled. In fact, EGSnrc takes into account atomic relaxation leading to the creation of fluorescent photons from the K, L and M shells, Auger and Coster-Kronig electrons resulting from photoelectric or Compton interactions. On the other hand, the production of characteristic photons through electron impact ionization (EII) is not handled, as a result of what the simulated characteristic peaks fluence is usually smaller than expected. Nevertheless, Verhaegen et al. (1999) found that the uncertainty on the photon peaks had only a limited effect on dosimetric quantities simulated such as HVL because the majority of the characteristic x-rays in thick target were created through photoelectric emission after re-absorption of bremsstrahlung photons in the target.

3.4.2 EGSnrc User Codes

EGS/BEAMnrc is a user code that allows the modeling of complex geometries such as an x-ray tube assembly. Each part of the tube is considered to be a single component module (CM) and is assigned to a specific horizontal slab portion of the whole geometry (Rogers et al., 2003). A typical CT geometry modeled with BEAMnrc is shown in figure 3.7. BEAMnrc is typically used to accumulate particle information in phase space files scored at various plane levels of an x-ray tube geometry, which can then be used by the user for a variety of applications. To perform BEAMnrc simulation, the user must create input files, either using a text editor like emacs or the graphical user interface available with BEAM (beam_gui), where all the parameters concerning the geometry of interest are specified based on manufacturer information or direct measurements.
Other simulation information like particle transport parameters, cross section data, number of histories and variance reduction techniques also need to be specified in the input file.

Phase space files obtained with BEAM can be analyzed using the BEAMDPnrc user code. Information such as energy fluence, mean energy and angular distribution of photons, electrons or positrons can be easily extracted from a phase space file. It is also possible to inter-compare characteristics of different phase space files on the same graph.

The DOSXYZnrc code is used to calculate dose distribution in a rectilinear voxel phantom from full phase space files or particle spectra (Walters et al., 2005). The user has the possibility to manually specify the dimensions of the voxels in the x, y and z orientation as well as the density and material for any given voxel. Phantoms for use in DOSXYZnrc can also be created from a CT scan data set using the ctcreate program. The code was used in this work to model and calculate dose distributions in a MOSFET dosimeter model, a cylindrical PMMA phantom and a solid water
phantom. One of the sources available in DOSXYZnrc offers the possibility to have particles generated from multiple directions and was used to simulate axial CT scans.

The FLURZnrc code is based on a cylindrical RZ geometry and was used to calculate particle fluence at different depths in solid water and PMMA phantoms. It was used in conjunction with the g program that calculates air-kerma, mass energy-absorption coefficient and mass energy-transfer coefficients.

3.4.3 Variance Reduction Techniques

Monte Carlo simulation makes it possible to model almost every interaction a particle might undergo in a given medium, allowing complex situations to be simulated very accurately. While modeling every single interaction of every particle set in motion is technically possible, it would be incredibly time consuming, even for a simple situation like x-ray generation by electrons incident on a tungsten target. Moreover, in many cases, a lot of time would be wasted tracking down particles that would not even contribute to the dosimetric quantity of interest.

The efficiency $\varepsilon$ of a MC simulation can be calculated as:

$$\varepsilon = \frac{1}{\sigma_{rel}^2 T}$$

(3.8)

where $T$ is the total simulation time, $\sigma_{rel}$ is the relative statistical error on the mean over the batches and is approximately proportional to $1/N^2$, with $N$ corresponding to the total number of particles simulated. It is usually desired to obtain calculated dosimetric quantities with the smallest error possible, which comes at the cost of an increased number of particles and longer simulation time that ultimately translates in a decrease in the simulation efficiency. Techniques that allow reducing the relative error without decreasing the efficiency exist and are usually called variance reduction techniques.
Among the various techniques available, the transport energy cut-off of photons (PCUT) and electrons (ECUT) was systematically used in all the simulations to terminate particle histories when their energy dropped under a predetermined value. For example, it was found during test runs that terminating electron transport in an x-ray target at 10 keV instead of 1 keV had no impact on photon spectra generated while significantly decreasing simulation time.

When using electron range rejection, the residual range of an electron is calculated before it is effectively transported. If the residual range does not exceed the distance to the nearest boundary, the history is terminated, avoiding the production of unnecessary low energy secondary electrons.

Bremsstrahlung splitting and Russian roulette were used together to increase photon production in the x-ray target, thus increasing the number of photons that actually reached the region of interest. A splitting factor of 100 was used for the BEAMnrc x-ray source simulation. Every bremsstrahlung photon created was split into 100 photons, and their weight was decreased by the same factor 100. However, having more photons created also means that more unwanted secondary electrons will be created, thus increasing simulation time. To avoid this problem, Russian roulette is used, where all the electrons created are assigned a survival threshold of 1/100, and a random number is chosen for each of them. If the random number is lower than the threshold, the electron survives and its weight is increased by a factor of 100, otherwise it is killed (Rogers et al., 2003).

The MOSFET sensitive region being very small, the probability of having a photon interaction taking place in, or close to this region is very low. To increase the probability of interaction, a photon splitting factor of 50 was used in conjunction with Russian roulette in the DOSXYZnrc MOSFET dose calculations. The efficiency of simulations performed with
photon splitting factor of 50 is almost 10 times greater than simulations where no photon splitting is used.

3.4.4 Modeling of the Picker PQ5000 CT Simulator X-Ray Source

The x-ray tube assembly of a Picker PQ5000 CT simulator was modeled using the EGS/BEAMnrc MC code based on information provided by the manufacturer and on actual measurements. The 7° angle x-ray target was modeled using the component module XTUBE. The actual target contains tungsten and rhenium, but since the exact proportion of rhenium was not known, and is usually very small, the target was modeled as pure tungsten. Photon spectra were generated using parallel rectangular mono-energetic electron beams incident from the side (ISOURCE 13) of the XTUBE.

The x-ray tube housing was not considered for the model, but the exit window was modeled as a thin aluminium slab using the SLABS component module. The thickness of the slab was originally set to the nominal inherent filtration of 1.5 mm specified by the manufacturer, but modifications were required to properly match the simulated HVL with the HVL measured at various tube potentials.

No precise information was available on the CT beam limiter aside from the fact that it is made of copper. It was modeled using the JAWS component module. The dimensions were determined using an iterative method until the simulated x-ray profile in-air obtained with the beam limiter in place would match the measured profile.

The CT collimator was also modeled using the JAWS component module. It was directly accessible by simply opening the bore, so the dimensions of the collimator opening along both the x-axis (slice thickness axis) and the y-axis (fan beam axis) could be measured and implemented in the model.

When it came to model the beam shaping filters for head and body scan, it was not possible to obtain any information from the manufacturer,
nor was it possible to have direct access to it. Thus, the two compensators could not be explicitly included in the model geometry. The effect of the compensators on the shape of the beam was taken into account by modifying the weights of particles with respect to their position in a phase space file simulated in the open beam configuration. To this end, a C-program was written that applied a position-dependent weight correction factor $W(y)$ to each particle of the phase space file. The correction factor was determined for the two filters based on the measured compensated profile and the simulated open beam profile as:

$$W(y) = \frac{MC(y)}{SO(y)}$$

which is simply the ratio of the equation of the fitted measured compensated profile data, $MC(y)$, over the equation of the fitted simulated open profile data, $SO(y)$.

Based on this detailed model of the x-ray tube target, filters and collimation, complete simulations of CT x-ray spectra associated with diverse combinations of beam energy, nominal slice thicknesses and open or compensated field were performed (chapter 4). The resulting phase space files were then used in DOSXYZnrc to study the characteristics of the MOSFET dosimeter (chapter 5) and to study the surface dose resulting from various scan conditions of a cylindrical PMMA head phantom (chapter 6).

References


Chapter 4 Results I: Validation of the Monte Carlo Model of the PQ5000 CT Simulator

The validation of the MC model of the PQ5000 CT scanner was made as a two-step process. First, the accuracy of the simulated photon spectra was evaluated through Half-Value Layer (HVL) comparison. Second, the geometrical characteristics of the x-ray beam were evaluated by comparing off-axis in-air radiation profiles in various exposure conditions.

4.1 Ionization Chamber, Electrometer and Radiographic Film

Throughout this work, all the measurements that required the use of an ionization chamber were performed using a Farmer-type Exradin A12 ionization chamber (Exradin Inc., Middleton, WI, USA) with a Keithley electrometer model 6517A (Keithley Instruments Inc., Cleveland, OH, USA). The ion chamber has a 0.65 cm$^3$ collecting volume and the wall, collector and guard material consists of Shonka air-equivalent plastic C552. Kodak X-Omat XV2 radiographic films were also used for relative dose measurements.

4.2 Half-Value Layer

The thickness of material that attenuates a beam to 50% is called the half-value layer or HVL (Johns and Cunningham, 1983). The second HVL corresponds to the thickness of material that attenuates a beam to 25%. It is obtained experimentally by measuring the variation of air-kerma with increasing thickness of attenuator (Ma et al., 2001). High purity aluminium (Al) and copper (Cu) were used as attenuator. The set-up used to measure HVLs is shown in figure 4.1. The ion chamber was placed at the CT isocenter and exposed while keeping the x-ray tube in a stationary position at the bottom of the gantry. The chamber was first exposed without any attenuator in the way of the beam. It was then exposed while
Figure 4.1: Schematic diagram of the set-up used to measure HVLs (side view).

having increasing thickness of Al or Cu placed on the gantry in such a way that all the radiation that reached the chamber sensitive volume had previously passed through the Al or Cu attenuator. The simulated HVLs were determined from photon fluence spectra scored in air at a distance of 62.75 cm from the focal spot, corresponding to the source-to-isocenter distance of the CT where the measurements were performed.

The first step of the validation of the x-ray source model was done by matching the HVLs simulated with those measured for tube potentials ranging from 80 to 140 kVp when only the housing inherent filtration was in the way of the beam. The inherent filtration, corresponding to the housing exit window, was first modeled according to the manufacturer specification by including a slab of 1.5 mm of Al right under the x-ray target. The thickness of this Al slab was then slightly modified until good agreement was obtained between measured and simulated HVLs. It was found that a thickness of 1.27 mm of aluminium yielded discrepancies not larger than 2.2% for the aluminium HVLs over the whole kVp range considered (see Table 4.1).
Table 4.1: Measured and simulated half-value layers in mm of aluminium.

<table>
<thead>
<tr>
<th>Tube Potential</th>
<th>1st HVL mm Al</th>
<th>2nd HVL mm Al</th>
</tr>
</thead>
<tbody>
<tr>
<td>kVp</td>
<td>Measured</td>
<td>Simulated</td>
</tr>
<tr>
<td>80</td>
<td>2.69</td>
<td>2.73</td>
</tr>
<tr>
<td>90</td>
<td>3.10</td>
<td>3.10</td>
</tr>
<tr>
<td>100</td>
<td>3.50</td>
<td>3.51</td>
</tr>
<tr>
<td>110</td>
<td>3.94</td>
<td>3.89</td>
</tr>
<tr>
<td>120</td>
<td>4.31</td>
<td>4.32</td>
</tr>
<tr>
<td>130</td>
<td>4.77</td>
<td>4.74</td>
</tr>
<tr>
<td>140</td>
<td>5.16</td>
<td>5.16</td>
</tr>
</tbody>
</table>

Table 4.2 shows that the copper second HVL simulated do not vary by more than 3.9%. On the other hand, larger discrepancies are observed for first HVLs at the lower energies, reaching a maximum of 12.1% at 80 kVp. This is due to limitation in the thickness of copper slabs available for the measurements. In fact, the thinnest slab of copper available was 0.1 mm, which is already of the order the first HVL. Therefore, the accuracy of the HVL measured at low energy was limited by the small amount of data points that could be collected.

Table 4.2: Measured and simulated half-value layers in mm of copper.

<table>
<thead>
<tr>
<th>Tube Potential</th>
<th>1st HVL mm Cu</th>
<th>2nd HVL mm Cu</th>
</tr>
</thead>
<tbody>
<tr>
<td>kVp</td>
<td>Measured</td>
<td>Simulated</td>
</tr>
<tr>
<td>80</td>
<td>0.11</td>
<td>0.09</td>
</tr>
<tr>
<td>90</td>
<td>0.12</td>
<td>0.11</td>
</tr>
<tr>
<td>100</td>
<td>0.14</td>
<td>0.13</td>
</tr>
<tr>
<td>110</td>
<td>0.15</td>
<td>0.15</td>
</tr>
<tr>
<td>120</td>
<td>0.17</td>
<td>0.17</td>
</tr>
<tr>
<td>130</td>
<td>0.20</td>
<td>0.20</td>
</tr>
<tr>
<td>140</td>
<td>0.22</td>
<td>0.23</td>
</tr>
</tbody>
</table>

4.3 Off-Axis In-Air Radiation Profile

4.3.1 Y-Axis Radiation Profile

The second step of the validation of the PQ5000 x-ray source model was done by comparing in-air off-axis radiation profiles to verify the
modeling of the beam limiter, collimator and bowtie filters, or compensators. Profiles were obtained by fixing the x-ray tube, and by measuring exposure in air with an ion chamber at a distance of 32.75 cm from the source along the fan beam direction (y-axis). A schematic diagram of the set-up used can be found in figure 4.2. The tube was fixed on the side of the gantry and the position of the chamber was shifted by 30 cm away from the isocenter in order to be at the specified distance of 32.75 cm from the source. The scanner couch was then moved up and down in order to cover the whole profile.

![Schematic diagram of the set-up used to measure radiation profiles along the y-axis (front view).](image)

Measurements were done for three different beam configurations: open beam, full compensator and half compensator. All data points were normalized with respect to the value at isocenter. Figure 4.3 shows that the simulated open profile drift away from the one measured as we move away from the central axis of the scanner.
Figure 4.3: Comparison of measured and simulated off-axis in-air radiation profiles along the scanner y-axis for the open beam setting.

Possible causes for these discrepancies at larger distances from the central axis were investigated. Among them, scattered radiation that could be coming from the gantry was studied. A metal ring was included in the model to simulate the CT gantry. Profiles were then simulated with and without the ring being present, but no change in profile shape was noticed, indicating that the number of scattered photons reaching the ion chamber was negligible and could not explain the increased signal measured. The effect of the focal spot size was also studied. Simulations were performed using a focal spot with size of 0.6 x 1.3 mm², the nominal size specified by the manufacturer. Focal spot 50% smaller and 50% larger than the nominal size were also used. The profile shape simulated along the y-axis was not modified when the focal spot size was changed.

Profiles were also measured and simulated with the full compensator, used for body scan, and the half compensator, used for head scan. Results can be found in figure 4.4 and figure 4.5, and show very good agreement between measured and simulated profiles, even at larger distances from the central axis.
4.3.2 X-Axis Radiation Profile

Radiation profiles were also measured and simulated in the x-axis direction, corresponding to the slice thickness profile. To obtain good spatial resolution, films were used to measure profiles for nominal slice thicknesses of 3, 5 and 10 mm at isocenter. All data points were normalized to the isocenter, and corresponding profiles can be found in figure 4.6.
Figure 4.6: Radiation profiles along the CT x-axis. The scanner collimator opening was set to obtain nominal slice thicknesses of 3, 5 and 10 mm at isocenter in an open beam configuration.

The agreement between measured and simulated slice profile is very good for the three different collimator opening studied, except in the penumbra
region where the relative exposure measured is higher than the simulations. As discussed in section 3.2 of chapter 3, this could be explained by a small non-linearity of the film response in the dose range of interest here. In fact, an over-response of the film in the low dose region of the penumbra would cause an artificially high exposure to be measured, as seen in figure 4.6.

The effect of changing the size of the focal spot has great influence on the profile along the x-axis (see Figure 4.7). Even though the full width at half maximum (FWHM) is identical for the three different focal spot sizes used, it was found that using a focal spot with dimensions 50% larger than the nominal dimensions specified by the manufacturer led to a better overall agreement between the measured and simulated slice profile.

Figure 4.7: Effect of changing the focal spot size on the slice radiation profile. Simulations for a nominal slice thickness of 10 mm at isocenter were performed with three different focal spot dimensions: nominal (0.6 x 1.3 mm²), large (0.9 x 1.95 mm²) and small (0.3 x 0.65 mm²).

The spatial definition of the focal spot is dependent on the gaussian spread of the pencil electron beam that is incident on the anode and responsible for the production of x-rays in the tube. Even though in reality
the spatial definition of the focal spot is neither symmetric nor well defined, it is nevertheless specified by manufacturers as being a rectangle with specific dimensions, and modeled as so in MC simulations. Furthermore, when taking into account the degradation of the focal spot spatial definition with time, it is therefore justified to model the focal spot with dimensions larger than the nominal manufacturer specifications.

References

Chapter 5 Results II: Study of the MOSFET Dosimeter

Response

In this chapter, the behaviour of the MOSFET dosimeter is studied. Evaluation is made of the dose-to-dose reproducibility of the MOSFET, the change in response with accumulated dose, the angular response and the energy response.

5.1 The MOSFET Dosimeter System and Phantom

The MOSFET dosimeter system used in this work is the Thomson and Nielsen Patient Dose Verification System model TN-RD-50 (Thomson & Nielsen Electronics Ltd., Nepean, Ontario, Canada). It comprises a reader that allows direct readout in units of mV, rad or cGy, a dual bias supply model TN-RD-22 for high and low sensitivity to which five high sensitivity MOSFET model TN-1002RD are connected. They have less than 0.04 mm$^2$ active detection area that allows pinpoint measurements. Use was also made of solid water slabs (GAMMEX RMI, Middleton, Wisconsin, USA) of various thicknesses.

5.2 MOSFET Measurement Protocol

To obtain optimal results in the context of the low doses delivered in CT diagnostic procedures, measurements involving the MOSFET dosimeters were performed according to a strict experimental protocol:

1. Expose MOSFET dosimeters
2. Two minutes waiting period
3. Read each MOSFET once then zero all dosimeters
4. Three minutes stabilization period
5. Repeat steps 1 to 4 for all subsequent exposures
Reading of a MOSFET is only done once to prevent any artificial signal increase due to the creep-up effect that was described previously in chapter 3. Once MOSFETs have accumulated around 1000 mV in the same experiment, an extra 15 minutes was added to the regular 3 minutes stabilization period of step 4 before more measurements were taken.

5.3 Reproducibility Test

The MOSFET response reproducibility was evaluated to determine the standard deviation, or uncertainty, that would apply to all subsequent measurements performed with the MOSFET dosimeters. With the x-ray tube at a fixed position, five different MOSFETs were placed on a solid water phantom and exposed ten times with the same technique (kVp and mAs). The exposure parameters (140 kVp, 70 mAs) were selected in order to obtain readings in the 100 mV range. The standard deviation (1 sigma) was calculated for each MOSFET based on this sample of ten readings. The experiment was then repeated by modifying the exposure parameters in order to obtain readings in the range of 200 mV (140 kVp, 140 mAs). The whole process was then repeated with five different MOSFETs. The results are shown in table 5.1.

Table 5.1: MOSFET reproducibility results. The uncertainty on all MOSFET readings presented in this work is based on the mean standard deviation obtained from the 20 samples in the table.

<table>
<thead>
<tr>
<th>Standard deviation</th>
<th>Mean Standard Deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>%</td>
<td>%</td>
</tr>
<tr>
<td>3.4</td>
<td>3.2</td>
</tr>
<tr>
<td>3.4</td>
<td>5.0</td>
</tr>
<tr>
<td>4.8</td>
<td>4.3</td>
</tr>
<tr>
<td>4.5</td>
<td>4.7</td>
</tr>
<tr>
<td>4.7</td>
<td>5.0</td>
</tr>
</tbody>
</table>
5.4 Sensitivity Change with Accumulated Dose

The intrinsic physical properties of semi-conductor detectors like MOSFETs are modified or altered every time they are irradiated with ionizing radiation. A change of physical properties of a dosimeter necessarily leads to changes in behaviour. The change in sensitivity of five different MOSFETs was evaluated at three stages throughout their lifespan. The detectors were irradiated with fixed exposure parameters and their response in mV was measured when they had no dose accumulated. The experiment was repeated at half their lifespan, or around 10 000 mV of accumulated dose, and again near the end of their life at around 18 000 mV of accumulated dose. The response was normalized to the initial response, and the resulting sensitivity curves can be seen in figure 5.1.

![Figure 5.1: Normalized sensitivity with accumulated dose of five MOSFETs.](image)

It is clear that MOSFETs become less sensitive as dose accumulates. On average for the 5 dosimeters, the sensitivity at 18 000 mV is 10.4% lower than the sensitivity at the beginning of their life, which exceeds the 5% decrease claimed by the manufacturer, but is still in agreement when taking into account the 6.4% uncertainty on the measurements. This change in sensitivity must be taken into account...
when measuring surface dose: one can recalibrate the MOSFETs more than once during their lifespan or simply apply an ad hoc correction factor assuming a 5% linear decrease in response over the 20 000 mV lifespan.

5.5 Modeling of a MOSFET Dosimeter

To study the MOSFET behaviour, a simple model of the MOSFET dosimeter based on manufacturer specifications was created using the DOSXYZnrc Monte Carlo code (see Figure 5.2).

![Figure 5.2: Schematic diagram of the MOSFET model based on manufacturer specifications used to study their energy and angular response. Dimensions are not drawn to the actual scale due to the extremely small size of the SiO2 sensitive region.](image)

The MOSFET consists of a silicon (Si) chip mounted on a Kapton wire covered by black epoxy. The silicon chip has a volume of 1 mm x 1 mm x 0.15 mm and the silicon-dioxide active volume is only 0.2 mm x 0.2 mm x 0.001 mm. The aluminium wire diameter is 0.0025 mm. The MOSFET components are encapsulated under an ovoid shaped drop of black epoxy. The package of the MOSFET is 8 mm long, 2.5 mm wide and 1.3 mm thick. The actual shape of the black epoxy drop encapsulating the MOSFET varies from detector to detector, thus this region was simply modeled as a rectangular box with dimensions corresponding to the nominal package dimensions available. This choice was also motivated by the work of Roshau and Hintenlang (2003) who showed that the
asymmetric epoxy packaging had almost no effect on the response of the dosimeter.

5.6 Angular Response

The angular response of the MOSFET dosimeter was measured free-in-air. A detector was irradiated at the CT isocenter with the epoxy side facing up where the x-ray tube was fixed. The tube was then rotated around the MOSFET and exposures were taken every 15 degrees at a tube potential of 120 kVp. The angular response was simulated using a parallel x-ray beam spectrum. A simulation using a full phase space file as x-ray source could not be performed due to the very large amount of photons required to obtain good statistical uncertainty in the small sensitive SiO$_2$ volume of the MOSFET, and the relatively limited amount of photons available in the phase space file (around 3 million). In fact, 1.5 billion incident particles per position were simulated to obtain 2% statistical uncertainty on the dose calculated with an x-ray spectrum, which would have required recycling the particles in a simulation using a phase space file about 500 times. Using such a high recycling factor would have led to an underestimation of the dose statistical uncertainty. Simulated and measured angular response are presented in figure 5.3, where the 0 degree position corresponds to the epoxy side of the MOSFET directly facing the beam, and the 180 degrees position corresponds to the Kapton side directly facing the beam. The response is normalized to the average value obtained for all irradiation positions.

The simulated response generally follows the measured response. Significant discrepancies are observed in the 135, 150, 210 and 225 degrees positions. These positions correspond to the beam being incident on the Kapton side of the dosimeter where the gold wires that connect the MOSFET to the bias box are located. Shielding of the sensitive SiO$_2$ region by the gold wires is probably responsible for the under-response measured at these positions.
Since no information was available on these wires, they were not included in the model, which explains the important discrepancies observed between the measured and simulated response at these positions.

The standard deviation on the measured and simulated response is respectively 7.1\%, and 8.5\% from the mean value, which is significantly higher than previous results by Roshau and Hintenlang (2003) who reported nearly isotropic response with a standard deviation of only 2.7\% from the mean value. The fact that they used a slightly different MOSFET (low-sensitivity MOSFET TN-502RDI by Thomson and Nielsen) added to the different tube potential used (70 kVp) might explain the different results obtained.

### 5.7 MOSFET Energy Response

#### 5.7.1 Mono-Energetic Photons

The energy response of the MOSFET was simulated in air with DOSXYZnrc using parallel mono-energetic x-ray beams with energy ranging from 10 – 140 keV. The response is expressed as the MOSFET dose per fluence over the air-kerma per fluence at a given energy. The
number of particles simulated was chosen in order to have less than 2% statistical error on the dose. The MOSFET energy response curve is plotted in figure 5.4 along with the curves of the ratio of the mass energy-absorption coefficient of silicon-to-air and silicon dioxide-to-air.

![MOSFET Response Curve](image)

Figure 5.4: Simulated MOSFET energy response to mono-energetic x-rays. The behaviour of the ratio of the mass energy-absorption coefficient of silicon-to-air and silicon dioxide-to-air is also plotted to show that electrons that deposit energy in the sensitive MOSFET region comes from both the SiO₂ sensitive region and the Si substrate surrounding it.

The mass energy-absorption coefficients were calculated with the $g$ program, part of the EGSnrc Monte Carlo code system, which allows the calculation of air-kerma and related parameters based on photon spectrum fluence. The MOSFET response curve closely follows the Si-Air curve indicating that most, but not all, photon interactions are taking place in the silicon substrate surrounding the sensitive SiO₂ region. At the higher energies, the predominant photon interaction process is the Compton effect, which is typically proportional to the atomic number $Z$ of the absorber ($Z_{(Si)} = 14$). As the energy decreases, the MOSFET response increases due to the predominance of the photoelectric effect, which is typically proportional to the cube of the atomic number $Z^3$ of the absorber.
The response reaches a maximum around 40 keV, after what it decreases rapidly due to attenuation effect in the silicon substrate.

5.7.2 Photon Spectra

The MOSFET energy response was measured in air on the PQ5000 CT simulator for tube potentials ranging from 80 – 140 kVp without added filtration. The normalized response $R$ is expressed as

$$ R = \frac{D/K_{air}}{(D/K_{air})_{80}} $$

(5.1)

where $(D/K_{air})_X$ is the ratio the dose over the air-kerma at tube potential $X$, and $(D/K_{air})_{80}$ is the ratio of the dose over the air-kerma at 80 kVp. The air-kerma was measured with the Exradin A12 ion chamber. The Exradin A12 chamber had been previously cross-calibrated against a reference Farmer chamber model 2505/3 on a kilovoltage therapy x-ray machine. The Exradin chamber exposure calibration factor $N_X$ is plotted in figure 5.5 and tabulated in table 5.2 for various energies. The reference Farmer chamber exposure calibration factor $N_X$ is also plotted in figure 5.5.

![Figure 5.5: Exposure calibration factors $N_X$ as a function of effective photon energy. The Exradin A12 $N_X$ factors were obtained through cross-calibration in the kilovolt range against a reference Farmer chamber model 2505/3.](image)

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The difference in the two $N_x$ curves is essentially due to differences in sensitive volume ($V_{Exradin \ A12} = 0.65 \text{ cm}^3; V_{Farmer \ 2505/03} = 0.69 \text{ cm}^3$) and wall material ($\text{wall}_{Exradin \ A12} = \text{C-552}; \text{wall}_{Farmer \ 2505/03} = \text{graphite}$) between the two chambers.

Table 5.2: Exradin A12 exposure calibration factors given for the energies at which the MOSFETs energy response is evaluated.

<table>
<thead>
<tr>
<th>Tube Potential (kVp)</th>
<th>HVL (mm Al)</th>
<th>Effective Energy (keV)</th>
<th>$N_x$ (R/nC)</th>
</tr>
</thead>
<tbody>
<tr>
<td>80</td>
<td>2.69</td>
<td>32.0</td>
<td>5.06 +/- 0.04</td>
</tr>
<tr>
<td>90</td>
<td>3.10</td>
<td>33.9</td>
<td>5.04 +/- 0.04</td>
</tr>
<tr>
<td>100</td>
<td>3.50</td>
<td>35.9</td>
<td>5.02 +/- 0.04</td>
</tr>
<tr>
<td>110</td>
<td>3.94</td>
<td>37.7</td>
<td>5.02 +/- 0.04</td>
</tr>
<tr>
<td>120</td>
<td>4.31</td>
<td>39.3</td>
<td>5.01 +/- 0.04</td>
</tr>
<tr>
<td>130</td>
<td>4.77</td>
<td>41.3</td>
<td>5.01 +/- 0.04</td>
</tr>
<tr>
<td>140</td>
<td>5.16</td>
<td>42.9</td>
<td>5.01 +/- 0.04</td>
</tr>
</tbody>
</table>

The energy response curve of five different MOSFET is plotted with the simulated response curve in figure 5.6. It is expressed as absorbed dose per unit air-kerma, according to equation 5.1, against tube potential. Even though the response changes slightly from MOSFET to MOSFET, the measured responses agree with the simulated response at all energies. The MOSFET response varies linearly in this energy range as expressed by the curve fit determination coefficient $R^2$ of 0.9724 associated with the simulated results. The response of the MOSFETs when irradiated at 80 kVp is about 10% higher compared to irradiation made at 140 kVp. As discussed earlier, this over-response is due to the increasing predominance of the photoelectric effect over the Compton effect as the photon energy decreases.
Figure 5.6: Comparison of simulated and measured energy response for five different MOSFETs. The simulated response is fitted to a linear curve with a determination coefficient $R^2 = 0.9724$.

Reference
Chapter 6 Results III: Surface Dose Measurements and Simulations

In this chapter, the Monte Carlo simulation calibration as well as the MOSFET calibration procedure based on MC calculations are presented. Surface dose measurements and simulations results obtained for a variety of scanning conditions using a CTDI head phantom are then presented in details.

6.1 Monte Carlo Simulation Calibration

In the previous chapters, all the simulations were presented as relative calculation results, but in this chapter the aim is to compare absolute dose measurements and simulations.

6.1.1 In-Phantom Dose Measurement

As discussed in chapter 3, AAPM protocol TG-61 (Ma et al., 2001) gives guidelines for water dosimetry in kilovolt x-ray beam. Due to space limitations introduced by the bore in a CT x-ray scanner, measuring dose using a water tank is not possible. Thus, dose was measured using a 30 x 30 x 15 cm³ solid water (sw) phantom, which can easily be set up on the PQ5000 scanner, based on TG-61 protocol adapted for solid water dosimetry. The phantom comprises removable solid water inserts at different depth where a chamber can be inserted. The Exradin A12 ion chamber was positioned in the phantom at a depth of 1.5 cm with the surface of the phantom aligned at the machine isocenter (62.75cm). The nominal slice thickness used was 10 mm at isocenter and the chamber was aligned along the x-ray fan beam to insure complete coverage of the sensitive volume by the field. The chamber was exposed in a stationary, open beam configuration (no extra filtration or compensator in place) at tube potentials of 120 and 140 kVp using 300mAs current time product.
The absorbed dose to solid water at 1.5 cm depth was then obtained according to

$$D_{sw, z=1.5 \text{ cm}} = M N_X P_{Q, \text{cham}} P_{\text{sheath}} \left( \frac{\mu_{en}}{\rho} \right)_{\text{air}}^{\text{sw}}$$  \hspace{1cm} (6.1)$$

which is basically the same equation as equation 3.6 described in chapter 3 except for the last term, where use is made of the ratio of the mean mass energy-absorption coefficient of solid water-to-air in solid water instead of the ratio of the mean mass energy-absorption coefficient of water-to-air in water. Since ratios for solid water-to-air are not available in TG-61, they were calculated using the EGSnrc g program for the 2 spectra of interest. The raw chamber measurement only needed to be corrected for pressure and temperature since the polarity and ion recombination correction factor were calculated as unity. The air-kerma calibration factor $N_X$ was calculated from the exposure calibration factor $N_X$ as explained in chapter 3. The Exradin A12 overall chamber correction factors $P_{Q, \text{cham}}$ tabulated in TG-61 for water were assumed to be also valid for solid water. No waterproofing sleeve was necessary so $P_{\text{sheath}}$ was therefore taken as unity. Details of the absorbed dose measurements at 1.5 cm in solid water are presented in table 6.1.

<table>
<thead>
<tr>
<th>Tube Potential kVp</th>
<th>Corrected Tube Reading nC</th>
<th>$N_X$</th>
<th>$P_{Q, \text{cham}}$</th>
<th>$\left( \frac{\mu_{en}}{\rho} \right)_{\text{air}}^{\text{sw}}$</th>
<th>Absorbed Dose cGy</th>
</tr>
</thead>
<tbody>
<tr>
<td>120</td>
<td>1.992</td>
<td>4.392</td>
<td>1.013</td>
<td>1.110</td>
<td>9.835 +/- 0.354</td>
</tr>
<tr>
<td>140</td>
<td>2.535</td>
<td>4.385</td>
<td>1.013</td>
<td>1.112</td>
<td>12.525 +/- 0.451</td>
</tr>
</tbody>
</table>

The uncertainty stated on the dose corresponds to the combined standard uncertainty of 3.6% described in TG-61.
6.1.2 Monte Carlo Simulation Calibration Factor

Dose calculations performed with DOSXYZnrc give results in units of Gray per incident particle on the x-ray target (Gy/particle). In order to have absolute dose calculations in units of Gy/mAs, a MC calibration factor \( F \) is obtained by relating the dose measured in Gy/mAs to the dose calculated in Gy/particle at a specific point in a phantom. The calibration factor is simply expressed as

\[
CF\left(\text{particle/mAs}\right) = \frac{D_{\text{measured}}(\text{Gy/mAs})}{D_{\text{calculated}}(\text{Gy/particle})}
\]  

(6.2)

Every energy used requires a different calibration factor. Most of the clinical CT scan procedures are performed at 120 and 140 kVp and so all the dose measurements and simulations were restricted to these two energies. Therefore, two different calibration factors were calculated and can be found in Table 6.2.

<table>
<thead>
<tr>
<th>Tube Potential</th>
<th>Measured Dose</th>
<th>Simulated Dose</th>
<th>Calibration Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>kVp</td>
<td>(10^{-4}) Gy/mAs</td>
<td>(10^{-20}) Gy/particle</td>
<td>(10^{15}) particles/mAs</td>
</tr>
<tr>
<td>120</td>
<td>3.278 +/- 0.118</td>
<td>6.879 +/- 0.041</td>
<td>4.766 +/- 0.174</td>
</tr>
<tr>
<td>140</td>
<td>4.175 +/- 0.150</td>
<td>8.243 +/- 0.049</td>
<td>5.065 +/- 0.185</td>
</tr>
</tbody>
</table>

As a mean of verification of these calibration factors, dose measurements and calculations were performed at 3 cm depth in the solid water phantom. The chamber was positioned at isocenter and exposed with a current time product of 300 mAs. The measured and simulated doses are in very good agreement as shown by the 0.32% discrepancy at 120 kVp and 0.21% discrepancy at 140 kVp (see Table 6.3).
Table 6.3: Measured and simulated dose at 3 cm depth in a solid water phantom using a stationary, open beam technique (300 mAs).

<table>
<thead>
<tr>
<th>Tube Potential</th>
<th>Measured Dose</th>
<th>Simulated Dose</th>
<th>Percent Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>kVp</td>
<td>cGy</td>
<td>cGy</td>
<td>%</td>
</tr>
<tr>
<td>120</td>
<td>7.218 +/- 0.260</td>
<td>7.195 +/- 0.043</td>
<td>-0.32</td>
</tr>
<tr>
<td>140</td>
<td>9.406 +/- 0.339</td>
<td>9.425 +/- 0.057</td>
<td>0.21</td>
</tr>
</tbody>
</table>

6.2 MOSFET Dosimeter Calibration

To relate the raw MOSFET dosimeter readout in units of mV, corresponding to the threshold voltage required to attain a predefined drain-to-source current, to a measurement of absorbed dose in units of mGy, a precise calibration is necessary. The calibration of the MOSFET is done in 4 steps:

1. Simulate dose at the CT isocenter using a stationary beam technique with DOSXYZnrc at the surface of a 30 x 30 x 15 cm³ solid water phantom in a voxel of 1 mm thickness.

   \[ D_{sw}(z = 0) \] _MC (mGy/mAs)

2. Measure dose with the five MOSFETs in the same conditions described in step 1.

   \[ D_{sw}(z = 0) \] _MOSFET (mV/mAs)

3. Calculate the MOSFET calibration factor in terms of absorbed dose in solid water.

   \[ N_{D,sw} = \frac{D_{sw}(z = 0) \_MC}{D_{sw}(z = 0) \_MOSFET} \] (mGy/mV) \hspace{1cm} (6.3)

4. Calculate the MOSFET calibration factor in terms of absorbed dose in any other medium.

   \[ N_{D,med} = N_{D,sw} \left( \frac{\mu_{en}/\rho}_{sw} \right)_{air} \] (mGy/mV) \hspace{1cm} (6.4)

   where \( \left( \frac{\mu_{en}/\rho}_{sw} \right)_{air} \) is the ratio of the mean mass energy-absorption coefficient of the medium of interest to solid water in air.

   In this work, surface dose measurements are all performed using a Polymethyl Methacrylate (PMMA) phantom. Therefore, ratios of
were calculated using the $g$ program. As discussed in chapter 5, the fact that MOSFET response exhibit non-negligible energy dependence in the kilovolt range, added to the change in response with accumulated dose, specific calibration factors must be determined for every energy at different stages of the detectors life. Calibration factors are typically around 0.26 mGy/mV at 120 kVp and 0.27 mGy/mV at 140 kVp.

6.3 CTDI Phantom

Surface dose measurements are all performed using a cylindrical CTDI head phantom of 15 cm in length and 16 cm in diameter made of PMMA, also called acrylic. It contains five PMMA inserts to be used for CTDI index measurements. The inserts were always left in place since only dose at the surface was of concern in this study.

For dose simulations, a cylindrical mathematical phantom with dimensions and composition corresponding to the CTDI head phantom was generated using a program in MATLAB. The $x$, $y$ and $z$ dimensions of the voxels used are respectively 5 mm x 5mm x 1mm.

6.4 Surface Dose: Stationary Tube Protocol

The centre of the CTDI phantom was placed at the isocenter on a specially designed thin plastic headrest fixed to the CT tabletop. A MOSFET placed on the surface of the phantom was irradiated using a stationary beam technique. The set-up geometry and the angles of irradiation are shown in figure 6.1. The dosimeter was exposed three times at every tube position using energies of 120 and 140 kVp combined to a current time product of 150 mAs and nominal slice thickness of 10 mm at isocenter. Note that the MOSFET and phantom were always kept in the same position. Only the tube was moved in order to successively irradiate the MOSFET under the 6 different angles indicated in figure 6.1. No extra filtration or compensator was used.
Dose calculations in DOSXYZnrc used phase space files previously obtained from full BEAMnrc simulations of the x-ray tube assembly as photon sources. Instead of running a different simulation for every tube position, which would have been time consuming, the cylindrical phantom was exposed to a stationary x-ray fan beam at position 0° only. The dose was then scored in voxels at the surface of the phantom at the positions corresponding to the 6 angles shown in figure 6.1, which is equivalent to the experimental procedure.

The surface dose results for the stationary beam technique can be found in table 6.4. As expected, dose decreases as the angle increases. This can be understood by analysing figure 6.1. In fact, as the angle increases, the distance between the source (tube) and the MOSFET increases, thus leading to a decrease in photon fluence at the dosimeter position (inverse square law effect). The decrease in dose at the MOSFET position is also due to attenuation of the beam by the phantom. The agreement between the measured and simulated doses is very good with
no more than 2.3% difference at 120 kVp. At 140 kVp, the maximum discrepancy between measured and simulated dose is 3.4% and occur at the 100° position.

Table 6.4: Measured and simulated surface dose for a stationary tube technique for 150 mAs current time product, 10 mm nominal slice thickness and no extra filtration or compensator.

<table>
<thead>
<tr>
<th>Tube Potential Position</th>
<th>Measured Dose mGy</th>
<th>Simulated Dose mGy</th>
<th>Percent Difference %</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>59.2 +/- 4.3</td>
<td>59.5 +/- 2.4</td>
<td>0.5</td>
</tr>
<tr>
<td>30</td>
<td>56.7 +/- 4.1</td>
<td>55.7 +/- 2.3</td>
<td>-1.7</td>
</tr>
<tr>
<td>45</td>
<td>54.2 +/- 4.0</td>
<td>52.9 +/- 2.2</td>
<td>-2.2</td>
</tr>
<tr>
<td>60</td>
<td>47.1 +/- 3.4</td>
<td>46.3 +/- 1.9</td>
<td>-1.3</td>
</tr>
<tr>
<td>100</td>
<td>11.2 +/- 0.8</td>
<td>12.6 +/- 0.5</td>
<td>2.3</td>
</tr>
<tr>
<td>130</td>
<td>2.7 +/- 0.2</td>
<td>2.8 +/- 0.1</td>
<td>0.1</td>
</tr>
<tr>
<td>0</td>
<td>75.3 +/- 5.5</td>
<td>76.1 +/- 3.1</td>
<td>1.0</td>
</tr>
<tr>
<td>30</td>
<td>73.1 +/- 5.3</td>
<td>72.8 +/- 3.0</td>
<td>-0.4</td>
</tr>
<tr>
<td>45</td>
<td>68.0 +/- 5.0</td>
<td>68.6 +/- 2.8</td>
<td>0.8</td>
</tr>
<tr>
<td>60</td>
<td>62.0 +/- 4.5</td>
<td>60.6 +/- 2.5</td>
<td>-1.8</td>
</tr>
<tr>
<td>100</td>
<td>15.2 +/- 1.1</td>
<td>17.8 +/- 0.7</td>
<td>3.4</td>
</tr>
<tr>
<td>130</td>
<td>2.5 +/- 0.2</td>
<td>4.1 +/- 0.2</td>
<td>2.2</td>
</tr>
</tbody>
</table>

6.5 Surface Dose: Single Axial Scan

Single axial scans were performed at tube potentials of 120 and 140 kVp, tube current of 175 mA and a rotation speed of 1.5 seconds/revolution, corresponding to typical adult brain scan parameters. The nominal slice thickness was 10 mm at isocenter and no filters or compensators were used. Surface dose was measured by distributing five MOSFETs on the exposed circumference of the CTDI phantom. The simulated dose results are given as an average of the dose in voxels situated all around the surface of the phantom along the exposed slice. Therefore, the measured doses for the five different MOSFETs are compared against a unique simulated dose value. The measured and
simulated results can be found in table 6.5. Measured and simulated results agree within 3.4% at 120 kVp and within 6.6% at 140 kVp.

Table 6.5: Surface dose results for a single axial scan. The dose was measured with five different MOSFETs placed on the phantom. For both 120 and 140 kVp, all the measured doses are compared against the unique simulated dose value indicated.

<table>
<thead>
<tr>
<th>Tube Potential kVp</th>
<th>MOSFET Serial Number</th>
<th>Measured Dose mGy</th>
<th>Simulated Dose mGy</th>
<th>Percent Difference %</th>
</tr>
</thead>
<tbody>
<tr>
<td>120</td>
<td>1601</td>
<td>46.9 +/- 3.4</td>
<td></td>
<td>-2.2</td>
</tr>
<tr>
<td></td>
<td>1602</td>
<td>45.9 +/- 3.4</td>
<td></td>
<td>0.1</td>
</tr>
<tr>
<td></td>
<td>1603</td>
<td>47.4 +/- 3.5</td>
<td>45.9 +/- 1.7</td>
<td>-3.2</td>
</tr>
<tr>
<td></td>
<td>1604</td>
<td>46.2 +/- 3.4</td>
<td></td>
<td>-0.5</td>
</tr>
<tr>
<td></td>
<td>1605</td>
<td>47.5 +/- 3.5</td>
<td></td>
<td>-3.4</td>
</tr>
<tr>
<td>140</td>
<td>1601</td>
<td>58.5 +/- 4.3</td>
<td></td>
<td>4.3</td>
</tr>
<tr>
<td></td>
<td>1602</td>
<td>57.1 +/- 4.2</td>
<td></td>
<td>6.6</td>
</tr>
<tr>
<td></td>
<td>1603</td>
<td>60.3 +/- 4.4</td>
<td>61.1 +/- 2.3</td>
<td>1.4</td>
</tr>
<tr>
<td></td>
<td>1604</td>
<td>62.1 +/- 4.6</td>
<td></td>
<td>-1.7</td>
</tr>
<tr>
<td></td>
<td>1605</td>
<td>60.0 +/- 4.4</td>
<td></td>
<td>1.9</td>
</tr>
</tbody>
</table>

6.6 Surface Dose: Contiguous Axial Scan

6.6.1 Scan Parameters

In this section are shown surface dose results obtained from scans of the CTDI phantom in axial mode using typical clinical adult brain scan parameters. Dose was measured at 120 and 140 kVp using tube current of 175 mA and gantry speed of 1.5 seconds/revolution. For brain scans, a beam limiter is used to narrow the CT fan beam thus limiting the amount of radiation outside the region of interest. To compensate for the difference in tissue thickness the beam goes through at the periphery of the head or phantom compare to the center, use is also made of the half-compensator. To study the effect of collimation on dose, nominal slice thicknesses of 3, 5 and 10 mm at isocenter were used. The scans covered 12 cm of the 15 cm long CTDI phantom. In order to have a complete coverage of the phantom within the 12 cm, 12 contiguous axial slices are required for the 10 mm collimator opening, 24 slices for the 5 mm opening and 40 slices.
for the 3 mm collimator opening. For measurements in axial mode, five MOSFETs were disposed in a linear array fashion on the CTDI phantom surface.

Built-in DOSXYZnrc x-ray sources do not allow simulating contiguous axial scan in one single simulation. On the other hand, one source allows simulation photons coming from multiple directions around a target and was used to simulate the single axial scans of section 6.5. Thus, to simulate contiguous axial scan, multiple single slice simulations were performed using different z position coordinates for each slice, and the corresponding 3ddose files were added together into one file using the combinedose program.

### 6.6.2 Surface Dose Distribution Profile

Figure 6.2 shows the simulated dose distribution pattern at the surface of the phantom at 140 kVp for the three different slice thicknesses. Dose peaks correspond to the center of a slice whereas the dose valleys correspond to positions on the phantom where contiguous slices meet. The difference between adjacent maxima and minima is 17% for the 5 and 10 mm slices but only 10% for the 3 mm slice. This is due to spatial resolution limitation introduced by the voxel width in the z-direction. In fact, dose is averaged over the 1 mm width of the voxels, and because the dose minima and maxima are separated by only 1.5 mm for the 3 mm slices, they cannot be perfectly resolved.
Figure 6.2: Simulated surface dose distribution profile for contiguous axial scan at 140 kVp for 3, 5 and 10 mm slice thicknesses.
6.6.3 Surface Dose Results

Simulated and measured dose results in the axial mode for nominal slice thicknesses of 3, 5 and 10 mm at 120 and 140 kVp can be found in table 6.6. Doses reported correspond to values obtained in the central region of the phantom.

Table 6.6: Measured and simulated (a) minimum and (b) maximum surface doses for contiguous axial scan with three different slice thicknesses.

<table>
<thead>
<tr>
<th>Tube Potential kVp</th>
<th>Slice Thickness mm</th>
<th>Minimum Dose Measured mGy</th>
<th>Simulated mGy</th>
<th>Difference %</th>
</tr>
</thead>
<tbody>
<tr>
<td>10</td>
<td>10</td>
<td>22.9 +/- 1.7</td>
<td>23.8 +/- 0.9</td>
<td>3.4</td>
</tr>
<tr>
<td>120</td>
<td>5</td>
<td>21.9 +/- 1.6</td>
<td>23.6 +/- 0.9</td>
<td>7.5</td>
</tr>
<tr>
<td>140</td>
<td>10</td>
<td>30.8 +/- 2.3</td>
<td>32.5 +/- 1.2</td>
<td>5.2</td>
</tr>
<tr>
<td>10</td>
<td>5</td>
<td>31.3 +/- 2.3</td>
<td>31.9 +/- 1.2</td>
<td>1.8</td>
</tr>
<tr>
<td>140</td>
<td>3</td>
<td>31.6 +/- 2.3</td>
<td>34.2 +/- 1.3</td>
<td>7.8</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Tube Potential kVp</th>
<th>Slice Thickness mm</th>
<th>Maximum Dose Measured mGy</th>
<th>Simulated mGy</th>
<th>Difference %</th>
</tr>
</thead>
<tbody>
<tr>
<td>10</td>
<td>10</td>
<td>27.2 +/- 2.0</td>
<td>28.2 +/- 1.1</td>
<td>3.6</td>
</tr>
<tr>
<td>120</td>
<td>5</td>
<td>28.4 +/- 2.1</td>
<td>28.5 +/- 1.1</td>
<td>0.4</td>
</tr>
<tr>
<td>140</td>
<td>10</td>
<td>36.3 +/- 2.7</td>
<td>39.0 +/- 1.5</td>
<td>6.9</td>
</tr>
<tr>
<td>10</td>
<td>5</td>
<td>38.7 +/- 2.8</td>
<td>38.5 +/- 1.4</td>
<td>-0.7</td>
</tr>
<tr>
<td>140</td>
<td>3</td>
<td>36.9 +/- 2.7</td>
<td>37.9 +/- 1.4</td>
<td>2.4</td>
</tr>
</tbody>
</table>

Large discrepancies are observed for the minimum dose results obtained for the 3 mm slice where the simulated dose is 12.4% higher than the measured dose at 120 kVp and 7.8% higher at 140 kVp. As discussed in section 6.5.2, this is probably partly due to the fact that the dose is averaged over the voxel width, which is of the same order as the distance between dose maxima and minima, thus overestimating the
points of minimum dose. All the other measured and simulated results are all within uncertainty tolerances and discrepancies do not exceed 7.5%. Except for the maximum dose result obtained at 140 kVp and 5 mm slice thickness where the simulated dose is 0.7% lower than the measured dose, simulated surface doses are all slightly larger than the measured doses. One possible explanation for this is that the CTDI phantom plastic holder was not included in the simulations meaning that the simulated x-ray beam is less attenuated than in experimental conditions, thus leading to slightly higher simulated doses.

It is interesting to note that collimation has only a small effect on doses measured or simulated. Table 6.7 shows the dose obtained at 5 mm and 3 mm slice thickness as a percentage of the dose at 10 mm.

Table 6.7: Surface Dose measured and simulated at 3 mm and 5 mm slice thickness as a percentage of the dose at 10 mm. Results for dose minima, \( \left( \frac{D_x}{D_{10 \text{ mm}}} \right)_{\text{min}} \), and dose maxima, \( \left( \frac{D_x}{D_{10 \text{ mm}}} \right)_{\text{max}} \), are tabulated.

<table>
<thead>
<tr>
<th>Tube Potential kVp</th>
<th>Slice Thickness mm</th>
<th>Relative Dose Measured</th>
<th>Relative Dose Simulated</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>( \left( \frac{D_x}{D_{10 \text{ mm}}} \right)_{\text{min}} )</td>
<td>( \left( \frac{D_x}{D_{10 \text{ mm}}} \right)_{\text{max}} )</td>
</tr>
<tr>
<td>120</td>
<td>5</td>
<td>95%</td>
<td>104%</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>97%</td>
<td>102%</td>
</tr>
<tr>
<td>140</td>
<td>5</td>
<td>102%</td>
<td>107%</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>102%</td>
<td>102%</td>
</tr>
</tbody>
</table>

Using a slice thickness of 3, 5 or 10 mm do not change the surface dose by more than 7%. McNitt-Gray et al. (1999) obtained similar results when they measured dose inside a CTDI phantom. Doses measured for 1, 3, 5 and 10 mm slice thicknesses were all within 3%. Therefore the dose for contiguous axial scan depends mostly on the exposure technique used (kVp and mAs).
6.7 Surface Dose: Helical Scan

6.7.1 Scan Parameters

The CTDI head phantom was scanned in helical mode using adult brain scan parameters (120 and 140 kVp, 175 mA, 1.5 seconds/revolution, beam limiter and half compensator in place). The scan spanned 12 cm and combination of pitches 0.5, 1, 1.25 and 2 were used with 3, 5 and 10 mm collimations.

6.7.2 Surface Dose Distribution Profile

The effect of pitch and collimation on dose distributions at the surface of the CTDI phantom was measured by wrapping radiographic films around it. Dose distribution profiles for all pitches and collimations can be found in figure 6.3.

(a)

(b)
Figure 6.3: 120 kVp surface dose distribution profiles measured with radiographic film on the CTDI phantom in helical mode for (a) 3 mm, (b) 5 mm and (c) 10 mm nominal slice thicknesses. Measurements were performed for pitch 0.5, 1, 1.25 and 2.0.

For a given pitch, as collimation increases, the distance between adjacent dose maxima increases, as well as the difference between maximum and minimum dose points (see Table 6.8). Similar effect is observed when collimation is kept constant and pitch increased. This effect is due to the larger gaps between directly exposed parts of the phantom created when higher pitch or larger collimation is used.

Table 6.8: Percent difference between adjacent dose minima and maxima measured with radiographic films.

<table>
<thead>
<tr>
<th>Collimation</th>
<th>3 mm</th>
<th>5 mm</th>
<th>10 mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pitch</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0.5</td>
<td>3%</td>
<td>3%</td>
<td>7%</td>
</tr>
<tr>
<td>1</td>
<td>8%</td>
<td>15%</td>
<td>15%</td>
</tr>
<tr>
<td>1.25</td>
<td>25%</td>
<td>37%</td>
<td>40%</td>
</tr>
<tr>
<td>2</td>
<td>53%</td>
<td>56%</td>
<td>60%</td>
</tr>
</tbody>
</table>

When assessing the risks associated with a scan, being able to determine the maximum dose delivered is of primary concern. Analysis of the surface dose profiles showed that by making a linear array of MOSFETs, at least one of the dosimeter would be on, or very close, to a
point of maximum dose for all collimation and pitch combinations, except for pitch 2 associated with nominal slice thickness of 10 mm where the distance between two dose maxima is larger than the length covered by the five MOSFETs. In the later case, more than five detectors would be required to insure good coverage.

Clearly, pitch has a great impact on the surface dose measured. The relative mean surface dose as measured with radiographic film varies approximately as 1/pitch (see Table 6.9). McNitt-Gray et al. came to same conclusion in a study from 1999 where they measured dose inside a CTDI phantom using TLDs.

<table>
<thead>
<tr>
<th>Pitch</th>
<th>Collimation 3 mm</th>
<th>Collimation 5 mm</th>
<th>Collimation 10 mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.5</td>
<td>1.9</td>
<td>2.0</td>
<td>1.8</td>
</tr>
<tr>
<td>1</td>
<td>1.00</td>
<td>1.00</td>
<td>1.00</td>
</tr>
<tr>
<td>1.25</td>
<td>0.8</td>
<td>0.80</td>
<td>0.8</td>
</tr>
<tr>
<td>2</td>
<td>0.6</td>
<td>0.6</td>
<td>0.6</td>
</tr>
</tbody>
</table>

### 6.7.3 Surface Dose Results

Dose measured at the surface of the CTDI head phantom with MOSFETs for helical scans are summarized in table 6.10, 6.11 and 6.12 for 3, 5 and 10 mm collimation respectively. The tube current was kept constant at 175 mA, a setting normally used for typical adult brain. The time required for the tube to perform 1 full rotation was also constant and set to 1.5 second/revolution. All the scans spanned 12 cm across the PMMA head phantom, as a result of what the scanning times depended on the pitch and the slice thickness used: the smaller the pitch and the slice thickness, the longer the scanning time required to cover the whole 12 cm. The maximum and minimum doses tabulated correspond to average values of the highest and lowest doses measured for three
different scans. Also indicated are the percent differences between the maximum and minimum MOSFET dose measurements along with the expected differences as measured with films (see Table 6.8 sec. 6.6.2).

Table 6.10: Surface dose minima and maxima measured with MOSFET dosimeters for helical scans with 3 mm collimation.

<table>
<thead>
<tr>
<th>Tube Potential</th>
<th>Minimum dose</th>
<th>Maximum dose</th>
<th>Percent Difference</th>
<th>Expected Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>kVp</td>
<td>mGy</td>
<td>mGy</td>
<td>%</td>
<td>%</td>
</tr>
<tr>
<td>0.5</td>
<td>43.7 +/- 3.2</td>
<td>47.1 +/- 3.5</td>
<td>7</td>
<td>3</td>
</tr>
<tr>
<td>1</td>
<td>20.8 +/- 1.5</td>
<td>25.1 +/- 1.8</td>
<td>17</td>
<td>8</td>
</tr>
<tr>
<td>1.25</td>
<td>14.1 +/- 1.0</td>
<td>21.4 +/- 1.6</td>
<td>34</td>
<td>25</td>
</tr>
<tr>
<td>2</td>
<td>6.4 +/- 0.5</td>
<td>17.3 +/- 1.3</td>
<td>63</td>
<td>53</td>
</tr>
</tbody>
</table>

Table 6.11: Surface dose minima and maxima measured with MOSFET dosimeters for helical scans with 5 mm collimation.

<table>
<thead>
<tr>
<th>Tube Potential</th>
<th>Minimum dose</th>
<th>Maximum dose</th>
<th>Percent Difference</th>
<th>Expected Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>kVp</td>
<td>mGy</td>
<td>mGy</td>
<td>%</td>
<td>%</td>
</tr>
<tr>
<td>0.5</td>
<td>43.5 +/- 3.2</td>
<td>48.0 +/- 3.5</td>
<td>10</td>
<td>3</td>
</tr>
<tr>
<td>1</td>
<td>21.4 +/- 1.6</td>
<td>25.8 +/- 1.9</td>
<td>17</td>
<td>15</td>
</tr>
<tr>
<td>1.25</td>
<td>12.0 +/- 0.9</td>
<td>21.7 +/- 1.6</td>
<td>45</td>
<td>37</td>
</tr>
<tr>
<td>2</td>
<td>6.2 +/- 0.5</td>
<td>18.7 +/- 1.4</td>
<td>67</td>
<td>56</td>
</tr>
<tr>
<td>0.5</td>
<td>61.6 +/- 4.5</td>
<td>66.8 +/- 4.9</td>
<td>8</td>
<td>3</td>
</tr>
<tr>
<td>1</td>
<td>28.9 +/- 2.1</td>
<td>34.6 +/- 2.5</td>
<td>17</td>
<td>15</td>
</tr>
<tr>
<td>1.25</td>
<td>19.6 +/- 1.4</td>
<td>31.1 +/- 2.3</td>
<td>37</td>
<td>37</td>
</tr>
<tr>
<td>2</td>
<td>7.9 +/- 0.6</td>
<td>25.9 +/- 1.9</td>
<td>70</td>
<td>56</td>
</tr>
</tbody>
</table>
Table 6.12: Surface dose minima and maxima measured with MOSFET dosimeters for helical scans with 10 mm collimation.

<table>
<thead>
<tr>
<th>Tube Potential kVp</th>
<th>Minimum dose</th>
<th>Maximum dose</th>
<th>Percent Difference</th>
<th>Expected Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>mGy</td>
<td>mGy</td>
<td>%</td>
<td>%</td>
</tr>
<tr>
<td>120</td>
<td>0.5</td>
<td>45.7 +/- 3.4</td>
<td>50.9 +/- 3.7</td>
<td>10</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>23.0 +/- 1.7</td>
<td>27.9 +/- 2.0</td>
<td>17</td>
</tr>
<tr>
<td></td>
<td>1.25</td>
<td>13.2 +/- 1.0</td>
<td>23.5 +/- 1.7</td>
<td>44</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>5.8 +/- 0.4</td>
<td>15.7 +/- 1.2</td>
<td>63</td>
</tr>
<tr>
<td>140</td>
<td>0.5</td>
<td>63.3 +/- 4.6</td>
<td>70.3 +/- 5.2</td>
<td>10</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>30.0 +/- 2.2</td>
<td>37.7 +/- 2.8</td>
<td>20</td>
</tr>
<tr>
<td></td>
<td>1.25</td>
<td>18.4 +/- 1.4</td>
<td>32.3 +/- 2.4</td>
<td>43</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>8.0 +/- 0.6</td>
<td>27.0 +/- 2.0</td>
<td>70</td>
</tr>
</tbody>
</table>

As expected, the highest doses are obtained at the lowest pitch setting (P=0.5), and the highest pitch (P=2) yields the lowest doses. The highest dose maxima measured at 140 kVp and 262.5 mAs per revolution is 71.3 mGy, obtained with 3 mm collimation, and the lowest dose maxima is 24.5 mGy, obtained at 3 mm collimation. At 120 kVp and 262.5 mAs per revolution, the highest dose maxima is 50.9 mGy, whereas the lowest dose maxima is 15.7 mGy, both for measured for 10 mm collimation.

Differences between maximum and minimum doses measured with MOSFETs are generally in good agreement with the differences observed from film measurements. It should be noted that dose from film measurements was determined by assuming a linear response, or linear change in optical density, of the film with accumulated dose. This linear behavior between optical density and deposited dose is only valid for a limited dose range and usually breaks down at low doses where an over-response is observed. This over-response at low doses probably explains the fact that for pitch settings of 1.25 and 2, where the lowest dose are observed, the differences between maximum and minimum doses measured with film are significantly smaller than differences measured with MOSFETs.
As was the case for contiguous axial scan, collimation has only a small impact on the surface doses measured. In fact, for a given pitch, almost all exposures made at 3 and 5 mm collimation yield doses within 9% of the dose at 10 mm. The only exceptions are for the minimum dose at 140 kVp and pitch 2 where the difference in dose between 3 and 10 mm collimation is 18% and at 120 kVp and pitch 2 where the difference in dose between 5 and 10 mm is 19%, but in terms of absolute dose, the 18% and 19% differences correspond to respectively only 3.0 mGy and 1.5 mGy.

Shown in table 6.13 are the mean surface doses measured with MOSFET dosimeters expressed as the ratio of the mean doses at pitch 1. The mean dose is approximately inversely proportional to the pitch, which confirm results obtained with films (see Table 6.9).

Table 6.13: Mean surface dose measured with MOSFET dosimeters as a ratio of the mean dose at pitch 1.

<table>
<thead>
<tr>
<th>Tube Potential</th>
<th>Pitch 0.5</th>
<th>Pitch 1</th>
<th>Pitch 1.25</th>
<th>Pitch 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>kVp 120</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>2.0</td>
<td>1.9</td>
<td>1.9</td>
<td>1.9</td>
<td>1.9</td>
</tr>
<tr>
<td>1.0</td>
<td>1.0</td>
<td>1.0</td>
<td>1.0</td>
<td>1.0</td>
</tr>
<tr>
<td>0.8</td>
<td>0.7</td>
<td>0.8</td>
<td>0.8</td>
<td>0.8</td>
</tr>
<tr>
<td>0.5</td>
<td>0.5</td>
<td>0.5</td>
<td>0.5</td>
<td>0.5</td>
</tr>
<tr>
<td>Tube Potential</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>kVp 140</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>2.0</td>
<td>2.0</td>
<td>2.0</td>
<td>2.0</td>
<td>2.0</td>
</tr>
<tr>
<td>1.0</td>
<td>1.0</td>
<td>1.0</td>
<td>1.0</td>
<td>1.0</td>
</tr>
<tr>
<td>0.8</td>
<td>0.8</td>
<td>0.8</td>
<td>0.8</td>
<td>0.8</td>
</tr>
<tr>
<td>0.5</td>
<td>0.5</td>
<td>0.5</td>
<td>0.5</td>
<td>0.5</td>
</tr>
</tbody>
</table>

References

Chapter 7 Conclusion

MOSFET-type dosimeters were used to measure surface dose delivered in CT examinations. The small size of MOSFETs combined to the ease with which they can be manipulated make them excellent tools for this task, provided their behaviour is well understood so precise measurements can be performed. Monte Carlo (MC) simulations were used to serve as a tool to investigate MOSFET characteristics, to achieve accurate calibration and to validate surface dose measurements done with MOSFETs.

A detailed model of a CT x-ray tube was successfully developed, as shown by the measured and simulated aluminium HVL that agreed to within 2.2% for beam energies ranging from 80 to 140 kVp. The good agreement obtained comparing measured and simulated in-air radiation profiles, along both the x and y-axis of the CT scanner, for a variety of beam configurations completed the validation of the model.

The MOSFET dosimeter response was studied through experimental measurements, and a MOSFET model was built for use in MC dose response simulations. The detectors were found to have a dose-to-dose reproducibility of 4.5%, and exhibited a mean change in sensitivity response of 10.4% with accumulated dose over their lifespan of 20 000 mV. Investigation of angular response showed a slight anisotropy as the standard deviation on the measured and simulated response was respectively 7.1%, and 8.5% from the mean value over 360°. An over-response of around 10% was observed when the energy of the incident radiation was decreased from 140 kVp to 80 kVp.

Different calibration factors are required for each energy used because of the significant energy dependence of the detectors. Recalibration of the MOSFETs is also required at various stages of their life due to changes in response with accumulated dose. At least one recalibration halfway through the MOSFETs lifespan, at around 10 000 mV
of accumulated dose, should be performed. For very precise measurements, three different calibrations would be preferable: once the first time they are used, once around 7000 mV, and then again at around 14000 mV of accumulated dose.

Surface dose on a cylindrical PMMA phantom was assessed in a multitude of scanning conditions, ranging from the simple single exposure with a stationary tube to the more complex helical scans. Simulations were performed for a stationary tube, single axial scans and contiguous axial scans. The simulated results agreed with measurements to within 3.4% for the fixed tube technique and 6.6% for the single axial scans. Minimum and maximum surface dose simulated for contiguous axial scans were also in good agreement with the MOSFETs measurements, with generally less than 7.5% discrepancies.

The wide variation in surface dose distribution profiles introduced by the different pitch-slice thickness combinations used in helical scans was measured with film. Analysis of the different profiles showed that by making a linear array of five MOSFETs, it was possible to effectively measure at least one dose minimum and one dose maximum, except for the combination of pitch 2 with nominal slice thickness of 1 cm. In the latter situation, more than five MOSFETs would be required. The mean surface dose was found to be largely dependent on the pitch setting, and varied approximately as 1/pitch. The minimum and maximum surface doses measured with the MOSFET also depended on pitch setting. As expected, the smallest pitch ($P=0.5$) yielded the highest dose maxima, 50.9 mGy at 120 kVp and 71.3 mGy at 140 kVp, whereas the largest pitch ($P=2.0$) yielded the lowest dose maxima, 15.7 mGy at 120 kVp and 24.5 mGy at 140 kVp. On the other hand, collimation only had a small impact on the surface dose measured. In fact, for a given pitch, the dose measured using 3 and 5 mm nominal slice thicknesses were almost all within 9% of the dose measured using 10 mm nominal slice thickness.
Recommendations for Clinical Use of MOSFETs

The main concern for the proper use of MOSFET dosimeters is the determination of calibration factors. The accuracy of the calibration mostly depends on the ability of the user to precisely evaluate the dose at the surface of a phantom in predetermined conditions. Although this work presented a method using MC simulations that effectively allowed calculating dose deposited within a very short distance from a phantom surface, 1 mm depth in this case, implementation of such MC based calibration methods is usually not possible in a clinical environment because of either time constraints, or the simple absence of MC simulations infrastructure. Nevertheless, in the context of monitoring patient dose in diagnostic procedures, 10% uncertainty on the measured dose is usually considered acceptable, so even though the precision level attainable with MC simulations might represent the gold standard, it is probably not essential. Intrinsic limitations in the reproducibility of the MOSFET response (evaluated as +/- 4.5% in this work) in the dose range observed in CT also motivates this statement.

Calibration factors should be determined for all CT beam energies the MOSFETs are to be used with. Ideally, the detectors should be calibrated the very first time they are used, then again at around 7 000 mV of accumulated dose, and finally once they have reached around 13 000 mV of accumulated dose. Care should be taken to always read out the MOSFETs at constant time intervals within 15 minutes following irradiation for both calibration and actual surface dose measurements. When these precautions were taken, MOSFETs were found to be able to measure surface dose on phantom well within a 10% uncertainty and should therefore be appropriate for monitoring surface dose on patients during clinical CT procedures.

A calibration protocol for clinical use of MOSFET is proposed in the appendix. It is based on the in-air method described in TG-61 for determination of absorbed dose to water at the phantom surface.
In the future, a model could be developed to perform surface dose calculations for helical CT scan procedures. Surface dose measurement in CT exams using MOSFETs could include other phantoms such as pelvis and extremities and therefore allowing the assessment of the surface dose delivered for most of the clinical scanning procedures. The use of MOSFET dosimeters could also be extended to other types of high dose diagnostic procedures such as fluoroscopy, MVCT and portal imaging.

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APPENDIX MOSFET Calibration Protocol for Clinical Use in CT Examinations

I. Absorbed Dose at the Surface: The In-Air Method with an Ion Chamber

The first step of the calibration requires to determine the absorbed dose to water at the phantom surface ($z_{ref}=0$) at a given beam energy based on the in-air method described in TG-61. It is recommended in TG-61 to use a Farmer-type cylindrical chamber calibrated free-in-air. The CT collimator opening should be set wide enough in order to completely cover the sensitive chamber volume. The absorbed dose to water at the phantom surface can be determined according to:

$$D_{w,z=0} = M \cdot N_K \cdot B_w \cdot P_{stem,air} \left( \frac{\mu_{en}}{\rho} \right)_{w,air}^{w}$$

(A.1)

where $M$ is the free-in-air corrected chamber reading, with the center of the sensitive air cavity of the ionization chamber placed at the point of measurement ($z_{ref}$); $N_K$ the air-kerma calibration factor for the given beam quality; $B_w$ the water backscatter factor to account for phantom scatter; $P_{stem,air}$ the chamber stem correction factor accounting for the change in photon scatter from the chamber stem between the calibration and measurement and $\left( \frac{\mu_{en}}{\rho} \right)_{w,air}^{w}$ the ratio for water-to-air of the mean mass energy-absorption coefficients averaged over the incident photon spectrum.

For Farmer-type cylindrical chambers, $P_{stem,air}$ is usually well within 1% (Ma et al., 2001) and can therefore be considered as unity. Values for $B_w$ and $\left( \frac{\mu_{en}}{\rho} \right)_{w,air}^{w}$ can be found in TG-61 for various beam energies and field sizes.
II. Absorbed Dose to the MOSFET at the Surface

The second step of the calibration requires measuring the dose deposited in the MOSFET at the surface of a water phantom ($Z_{\text{ref}}=0$), $D_{\text{MOSFET,}z=0}^w$, in units of mV. The backscattering material should normally be water, but due to space limitations introduced by the CT gantry and other practical purposes, using a water tank for this measurement is not recommended. It is recommended to use a solid water phantom instead, and it will be shown that both measurements are equivalent. The MOSFET should be fixed at the surface of the phantom, and exposed in the same geometrical and exposure settings used to determine the absorbed dose to water described in section I. The measurement obtained is the absorbed dose to the MOSFET at the surface of a solid water phantom, $D_{\text{MOSFET,}z=0}^s$, in units of mV, which can be related to the dose deposited in the MOSFET at the surface of a water phantom $D_{\text{MOSFET,}z=0}^w$ as:

$$D_{\text{MOSFET,}z=0}^w (mV) = B_w \frac{D_{\text{MOSFET,}z=0}^s (mV)}{B_{sw}} \quad (A.2)$$

where $B_w$ and $B_{sw}$ are respectively the water and solid water backscatter factors for the field size and beam energy of interest.

Monte Carlo calculations for beam energies of 120 kVp (4.31 mm Al HVL) and 140 kVp (5.16 mm Al HVL) have shown that $B_w$ and $B_{sw}$ did not differ by more than 0.1% for the field size of $1 \times 30 \text{ cm}^2$ that was used in this work for MOSFET calibration. Thus, the ratio of the water-to-solid water backscatter factor can be considered as unity, and equation A.2 simply becomes:

$$D_{\text{MOSFET,}z=0}^w (mV) = D_{\text{MOSFET,}z=0}^w (mV) \quad (A.3)$$
III. MOSFET Calibration Factor

The MOSFET water calibration factor, $N_{D,w}$, for a given beam energy can be determined as:

$$N_{D,w} \text{(mGy/mV)} = \frac{D_{w,z=0} \text{(mGy)}}{D_{MOSFET,z=0} \text{(mV)}} \quad (A.4)$$

where $D_{w,z=0}$ is the absorbed dose to water at the phantom surface in units of mGy measured with the ion chamber, and $D_{MOSFET,z=0}$ is the dose deposited in the MOSFET at the surface of a water phantom in units of mV. $N_{D,w}$ can then be used to determine the dose at the surface of a water phantom in any scanning conditions.

In order to use MOSFET dosimeters to measure dose at the surface of a patient skin, a MOSFET skin calibration factor, $N_{D,\text{skin}}$, is required, and can be obtained as:

$$N_{D,\text{skin}} \text{(mGy/mV)} = \frac{\left(\bar{\mu}_\text{en} / \rho\right)_\text{water, skin} \times N_{D,w} \text{(mGy/mV)}} {\left(\bar{\mu}_\text{en} / \rho\right)_\text{water, air}} \quad (A.5)$$

where $\left(\bar{\mu}_\text{en} / \rho\right)_\text{water, skin}$ is the ratio for skin-to-water of the mean mass energy-absorption coefficients averaged over the incident photon spectrum in air, for which values are tabulated for ICRP skin tissue for various beam quality in TG-61, and $N_{D,w}$ is the MOSFET water calibration factor obtained from equation A.4.

IV. Absorbed Dose at the Surface of the Patient Skin

The absorbed dose at the surface of a patient skin resulting from a CT examination at a given beam energy can be determined in any scanning conditions as:

$$D_{\text{skin,z=0}} \text{(mGy)} = D_{\text{MOSFET,z=0}}^{\text{skin}} \text{(mV)} \times N_{D,\text{skin}} \text{(mGy/mV)} \quad (A.6)$$

where $D_{\text{MOSFET,z=0}}^{\text{skin}}$ is the dose deposited in the MOSFET at the surface of the patient skin in units of mV, and $N_{D,\text{skin}}$ is the MOSFET skin calibration factor in units of mGy/mV as described in equation A.5.
Reference