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MOSFET annealing and interface dosimetry in contemporary radiation therapy

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CERTIFICATION

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ABSTRACT

[Background and Purpose] In-vivo, skin, and interface dosimetry in modern radiation therapy is an important issue that demands a reliable radiation dosimeter capable of measuring skin doses. The Metal Oxide Semiconductor Field Effect Transistor “MOSFET” dosimeter became popular as a radiation dosimeter in radiation therapy and one of its recently designed versions for skin dosimetry, called MOSkin, has become very popular. Unfortunately MOSFET has a limited lifespan due to a saturated build up of positive charges in its sensitive dosimetric volume - gate oxide, which is mainly located near the Silicon- Silicon dioxide (Si-SiO$_2$) interface. This built up charge reduces its sensitivity, linearity, and stability. The aims of this thesis are an investigation of MOSkin™, with the possibility of annealing the MOSFET dosimeter designed by the Centre for Medical Radiation Physics, and being able to re-use it again after multiple cycles of irradiation and annealing. Virgin and annealed MOSkin detectors have been reused in addition to an Attix chamber, a Monte Carlo simulation, and EBT2 films to verify a newly developed radiotherapy treatment planning algorithm (TPA), the Acuros® XB (Varian, Palo Alto, CA). This verification covered dose predictions in the areas with strong electronic disequilibrium such as the build-up region and interfaces inside the phantom such as tissue – air, tissue – lung, and tissue – steel. The current clinical treatment planning algorithm (TPA), the Anisotropic Analytical Algorithm AAA (Varian, Palo Alto, CA) was verified experimentally in comparison to the Acuros® XB.

[Materials and Methods] Groups of MOSkin detectors were irradiated with different doses on a 6 MV X-ray beam from a medical linear accelerator (LINAC)
and then annealed using different annealing methods. Ultraviolet (UV) light with 4.92 eV mono-energetic photons from a filtered Mercury vapour lamp was used for UV annealing, while isothermal annealing was carried out in the furnace heated to a temperature of 150°C. Direct electric current (DC) was also used to anneal MOSkin with a current ranging from 5 mA to 15 mA, and pulsed electric current was used for current annealing with a current ranging 50-300mA per pulse. The annealing time required to recover the threshold voltage ($V_{th}$) varied for each type of annealing, so the parameters influent on this were studied. The annealed MOSkin detectors were irradiated again to check their sensitivities, linearity, and signal stability. The cycles of annealing and irradiation for each group of MOSkin were repeated many times to investigate the effectiveness of each annealing method. To verify and benchmark the treatment planning algorithms, MOSkin and Attix ionising chambers were used to verify the dose calculated by the Acuros® XB, and the AAA at the build-up region for 6 MV and 18 MV X-ray photon beams with field sizes of 4 cm x 4 cm, 10 cm x 10 cm, and 40 cm x 40 cm, under a normal incidence X-ray beam from linear accelerator (LINAC). To check the performance of treatment planning systems (TPSs) for dose calculations under an oblique beam, experiments with a 45˚ beam incidence and 10 cm x 10 cm field size with 6 MV and 18 MV X-ray photons beams were carried out.

For interface dosimetry, MOSkin and EBT2 films were used to measure doses near the air, steel, and lung interfaces, as well as using Monte Carlo simulations with 6 MV and 18 MV photon beams and 10 cm x 10 cm field sizes, for all but the lung cases, with a small 3 cm x 3 cm field size.
[Results] The annealing methods that were investigated revealed different clinical outcomes. UV light was able to recover the threshold voltage but not the sensitivity with essential MOSkin $V_{th}$ instability (within ±28 mV). Isothermal annealing was a good alternative in that it fully recovered the sensitivity and threshold voltage with an acceptable signal instability (±4 mV). DC annealing was shown to be much better than UV and isothermal annealing provided full recovery of sensitivity, threshold voltage linearity dose response, and excellent stability within ±1 mV. Pulsed current annealing was the best annealing method of all, with full recovery of the sensitivity, threshold voltage dose response linearity, and excellent stability ±1 mV.

Dosimetry in the build-up region resulted in an excellent agreement between MOSkin and Attix (within ±2%), although Acuros® XB was much better than AAA in all the setups used. For interface dosimetry, Acuros® XB performed better than AAA near the interfaces, although it had some slight shift in depth dose distribution within 2 mm of the proximal and distal interfaces, for all cases except the lung. With the lung case, Acuros® XB and AAA performed satisfactorily and agreed with the Monte Carlo simulated doses within ±5%.

[Conclusions] Direct current annealing (DC) and pulsed current annealing are the best annealing techniques for a MOSkin dosimeter. This will have potential benefits for MOSFET applications in medicine and space dosimetry, especially for departments with limited budgets, and in developing countries.

The verification of Acuros® XB showed that this TPA performed much better than previous TPA and AAA, and it was very close to the MC simulation and experimental measurements. As an application for the annealing procedure, pulsed current annealing was used with build-up measurements. The total doses during the
experiments for build-up and interfaces dosimetry were 230 and 185 Gy, respectively. It has been shown that two periodically annealed MOSkins detectors were enough for all the build-up measurements, while for doses on interface dosimetry measurements, the 11 virgin MOSkins that were used clearly showed the benefits of the annealing technique developed for MOSkin dosimeters for clinical applications.
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PUBLICATIONS AND PRESENTATIONS


4. Eyad Alhakeem, **Sami AlShaikh**, A.B. Rosenfeld, S. Zavgorodni, “Validation of Acuros® XB against Monte Carlo calculations as well as MOSkin and EBT2 measurements at interfaces with high and low density media” the World Congress on Medical Physics and Biomedical Engineering in Beijing, China, May 26 - 31, 2012. [Accepted and presented as oral presentation]

5. Eyad Alhakeem, **Sami AlShaikh**, A.B. Rosenfeld, S. Zavgorodni, “Experimental and Monte Carlo verification of Acuros XB calculations near low and high density heterogeneities”, the 58th Annual Scientific Meeting of the Canadian organization of Medical Physicists and the Canadian College of Physicists in Medicine, July 11-14, 2012, Halifax, Canada, [Accepted and presented as oral presentation]

CHAPTER 1

1. Introduction

1.1 Radiation Therapy Dosimetry

Advancement in radiation therapy (RT) technology has led to improved treatment for cancer. Many techniques have been introduced over the past decade, conformal RT, intensity modulated RT (IMRT), image guided RT (IGRT), and stereotactic radiosurgery (SRS), to name a few, all of which have been improved by advancements in technology that improved the saving or of organs at risk. Current treatment planning calculations and dose delivery process is now much more complicated than historical radiation therapy, for instance the accuracy of dose delivery recommended by the International Commission on Radiation Units and Measurements (ICRU) should be within ±5%, which is broadly based on the biological response data for the tumour control probability (TCP), and the normal tissue complication probability (NTCP)[1]. However, to satisfy this percentage of precision in radiation dose delivery complex RT treatment technique requires accurate and reliable quality assurance procedures to verify the dose to target and the organs at risk. Local measurements \textit{(in-vivo)}, including skin dose or interface dose in some complicated treatment scenarios, where treatment planning may overestimate or underestimate skin dose [2, 3], add to the arsenal of dose QA assessment.
Moreover, RT modalities such as IMRT or SRS that produce high dose gradient regions require small detectors with high spatial resolution which can accurately measure the penumbral dose. Some detectors which can satisfy these requirements are solid state detectors such as a thermoluminescent detector (TLD), a semiconductor diode, and the Metal Oxide Semiconductor Field Effect Transistor (MOSFET) [4]. However, a TLD procedure requires a retrospective readout to get the measured dose, and the sources of uncertainties are high [4, 5], so getting precise and quick results online is not viable for TLDs. Hence for online readings semiconductor detectors are attractive for radiation dosimetry [6-10]. MOSFET detectors have been used successfully to measure surface, skin, and target dose [9, 11-14]. In addition, semi-conductor diodes are also widely used in radiation therapy for in-vivo dosimetry, however, commercial systems include a build up for measuring incident dose so they cannot be used for surface or skin dosimetry. They also have an energy response due to silicon that sometimes requires backscatter shielding compensation due to strong energy dependence at low energy radiation [6, 15].

To assess patient dose there are various options. In-vivo dosimetry can be performed on the patient’s skin to measure entrance dose, and skin dose. In phantom measurements can also be performed to verify incident dose or dose at air cavity interface regions, or to verify the skin dose for complex treatments. The detector selected for each experiment may vary so to measure a patient’s incident dose requires a detector with specific encapsulation or a build-up cab that can be used to measure the entrance dose relative to a specific depth and/or exit dose, or both of them can be used to determine the target dose [4, 16]. Contemporary RT like IMRT that are able to deliver a conformal dose to the target by multiple radiation fields, or
by increasing the number of monitor units (MUs), can increase the probability of error. This means that verifying complex treatment delivered to a phantom before delivery to a patient is of paramount importance. Thus, a single, multiple, or linear diode array detector can be used for point dosimetry to verify the planned target dose [17-19], while two dimensional detectors like films and electronic portal imaging devices (EPID), and more complex devices such as DELTA 4 [20] or ArcCHECK™ [21] for 3D dosimetry are also used. The measure of correctness of treatment delivery according to TPS is a special metric called gamma analysis [22-24].

1.2 Project Aims

Aim 1: In this thesis a MOSFET designed at the Centre for Medical Radiation Physics (CMRP), the MOSkin™ will be used because like other MOSFETs, it has a finite radiation lifespan. The linearity of the MOSkin response (threshold voltage, \( V_{th} \), vs absorbed dose) is degraded by an accumulated dose, typically 0-100 Gy, so the possibility of reusing MOSkin by recovering its initial parameters is investigated in this thesis.

An irradiated MOSFET is recovered using UV light, thermal annealing (see Chapter 3), and current annealing techniques, all of which are discussed (see Chapter 4 and Chapter 5). The purpose of these studies is to extend the radiation lifespan of the MOSFETs by recovering the initial parameters of the MOSkin after a certain dose increment that results in the same sensitivity as a virgin MOSkin. Multiple cycling of this process (irradiation-recovery-irradiation) and its limitations is investigated.
**Aim 2:** Predicting doses in a strong electronic disequilibrium such as the surface of a patient or interfaces inside of body cavities are complicated. This is reflected by the fact that commercially available TPS are not very good at predicting doses in those situations, and why a TPS based on Boltzmann transport equations (Varian®, Palo Alto, CA) and called Acuros® XB, has recently been developed to address this issue. The second aim of this study was to compare two TPAs Acuros® XB and the currently used AAA (Varian®, Palo Alto, CA) for dose prediction on the surface of a phantom and build in different interfaces (muscle–air, lung-muscle and steel-muscle). The predicted doses are compared to the doses on interfaces simulated by Monte Carlo and measured experimentally with MOSkin™ detectors designed for surface dosimetry (see Chapter 6).
2. LITERATURE REVIEW

2.1 Skin dosimetry

Skin is very sensitive to radiation, and the most sensitive part of the skin is the epidermis. The most sensitive part of the epidermis is the layer of basal cells which directly react with radiation to cause skin toxicity. Its depth varies, but it’s about 20-60µm deep. The effects of ionising radiation on the skin are a function of dose and time, although it can be summarised chronologically as: transient erythema within ~24 – 48 hrs, main erythematous reaction within 3 – 6 weeks; mainly due to basal cells damage, dermal ischaemia or necrosis within 8 – 16 weeks, dermal atrophy within 24 weeks, and telangiectasia and necrosis within 52 weeks [25, 26].

Many studies conducted with clinical cases in radiation therapy have shown that the skin dose is affected by the treatment technique, and the accessories and immobilisation used during radiotherapy, like thermoplastic masks, wedges, blocks and trays [2, 3, 27-33].

Fiorino et al, 1992, found that mask material increased skin dose for 40% to 44% in cases of head and neck radiation therapy, while tray and blocks can contribute another 3% to 4%. In addition, a decrease in the source to surface distance (SSD), also increases the skin dose [27]. Kim et al, 1998, investigated skin dose by changing the main clinical parameters, energy, field size (FS), SSD, blocks, physical wedges, dynamic wedges, and beam shaping with a multi-leaf collimator (MLC). They concluded that the skin dose increases for large field sizes of 40 x 40 cm² for energies of 8 MV and 18 MV, from 6% to 38% and from 5% to 44%, respectively. In
addition, an acrylic block tray increased these values for large FS by ~19% for both
genergies. Moreover, a skin dose that increases with a decrease in SSD also increases
with shaping blocks, wedges with large FS, and shaping MLC. However, an increase
in the wedge angle also increases skin dose because dynamic wedges have been
shown to increase in field skin dose compared to physical wedges, and MLC
decreases the skin dose compared to shaping blocks [28]. Lee et al, 2002,
investigated skin toxicity as a result of an increase in skin dose from multiple fields
extended intensity modulated radiation therapy (Ex-IMRT) for head and neck (H&N)
cancer, and concluded that this reaction resulted from causes such as the
immobilisation masks used during irradiation, or during the inverse treatment
planning system. An 18% reduction of the skin dose has been observed if the skin is
declared as sensitive area during inverse planning [29]. Butson et al, agreed with
previous authors that Perspex blocking trays increased the skin dose of a 6 MV X-
ray by about 22%, while an increase that occurred by shortening the SSD was much
lower [30]. These results were obtained by Chuang et al. 2002 from Ex-IMRT. This
lead some authors to investigate the difference between open and IMRT incident
fields for 0 and oblique angles because the IMRT beams which were used for Ex-
IMRT were mainly oblique. Dogan et al, found no difference between open field
conventional RT and IMRT for an oblique angle. However, any increase in the
surface dose is a result of a shift in the build-up region caused by an oblique angle
[34]. Price et al, investigated the skin dose for 6 MV IMRT and open field
conventional RT and concluded that there is no significant difference between IMRT
and open filed conventional RT [32]. Roland et al, evaluated the skin dose from three
RT techniques, Helical Tomotherapy, Serial Tomotherapy, and IMRT, for prostate
and H&N cancers, and found that Helical Tomotherapy deposits the highest skin
dose in prostate and H&N, followed by Serial Tomotherapy, and IMRT was the lowest [3].

Hence an innovative approach to find a suitable radiation detector for skin dose is necessary in order to obtain accurate and reliable results.

Many authors investigated several radiation detectors to measure the build-up region, skin, and surface dose for different radiation modalities [2, 3, 11, 27, 34-45].

The detectors used by those authors are one or more of the following: a parallel-plate ionisation chamber, TLD, film, and MOSFET. The Attix chamber is regarded as a reference detector for build-up and surface dose measurements, whereas TLD, film, and MOSFET are relative detectors that were calibrated against an ionisation chamber for this purpose. On the other hand, the Attix chamber cannot be used for skin or in-vivo dosimetry with a patient due to its size and high operational voltage, so other detectors are more convenient with patients for this aim [4]. Each one of those relative detectors has its own advantages and disadvantages that may make it more favourable than the others in various applications.

For instance, TLD is widely used for in-vivo dosimetry due to its tissue equivalency and its lower cost, and it can also be used for skin dosimetry by using an extrapolation method [35]. However, it is not suitable for on-line measurements because they take a long time to analyse and get results. Moreover so many sources of errors may arise from annealing and reading procedures that can reduce its reliability as a detector [4].
Radiochromic film is also being used for skin dosimetry because its effective depth is deeper than basal cells and it can be used with correction against an ionisation chamber for accurate dose estimation [39], or without correction, to get an approximate dose near basal cells [26]. However, it cannot be used online and radiochromic film calibration and readout procedures may introduce some uncertainties especially if the calibration and irradiated films were scanned after different period of time [46]. Handling, storage, or environmental factors such as temperature, humidity, light exposure, or scanning factors such as film orientation, resolution, scanner light source, can introduce large discrepancies in its readings [46], so a great deal of care is needed to obtain reliable results with radiochromic film. However, film shows excellent results for cavity measurements, but it under-estimates the dose for distal cavity (by 5 - 6%) as a result of self-build-up when the film is parallel to the beam direction [47]. Film dosimeter can give two dimensional dose map with good resolution (~80µm WED as for EBT2) [41, 48].

Rosenfeld et al, 1995, Butson et al, 1996 [11, 49], promoted MOSFET as a surface and skin dosimeter in radiation therapy. They found that bare MOSFET in excellent agreement with an Attix ionisation chamber on the build-up region, which makes it suitable for commissioning the build-up region and for measuring the skin dose with patient. It has many advantages, such as its small size, simple reading circuit, low power operation, ability to store the dose history, and it can be used for online readout. The sources of uncertainties in MOSFET readings is different than the film, it is mainly associated to temperature, creep-up effect/instability and angular dependence. It is not affected by the wire like ionisation chamber, because
it’s an integrator based on charge build-up on the gate, it can work and store signal even if there is no wire connected [49, 50].

2.2 The Metal Oxide Semiconductor Field Effect Transistor (MOSFET) as a radiation detector

The metal oxide semiconductor field effect transistor (MOSFET) is a four terminal semi-conductor device that is used to control the current in applications such as computer processors and memories. The MOSFET illustrated in Figure 1 has two identical terminals called source and drain that are made from highly doped silicon (Si) n or p-type, oxide gate layer between the source and drain, with all of them being placed on the Si substrate, as shown in Figure 1. The doping material of the substrate depends on the source and drain, if the source and drain are p-type Si, the substrate will be n-type Si and the MOSFET is called a p-channel MOSFET, while the opposite is called an n-channel MOSFET. When a positive potential is applied onto the gate of an n-channel MOSFET, a channel is formed between the source and drain with electrons as charge carriers, while the same effect is formed if a negative potential is applied onto a p-channel MOSFET, but with holes as charge carriers. There are two MOSFET designs which are based on their operational gate voltage (Vg); when Vg is zero, they are a depletion and enhancement mode MOSFET. A depletion mode MOSFET is on when Vg is zero, while an enhancement mode MOSFET is off when Vg is zero [51]. So, if a channel is created and there is an applied potential between the source and drain, an electric current (I_{ds}) will pass through. The current control is based on a change in the oxide charges or the gate voltage which affects the electric field on the oxide and the channel leading to a change in the resistivity of the channel.
An enhancement mode MOSFET is commonly used as a radiation detector, with or without bias applied onto the gate. The theory of this operation is based on the space-charge principle [7, 10] which applies when ionising radiation passes through the device; a charge will be created on the insulator, the positive charge will immigrate to the Si-SiO₂ interface where they will accumulate, which affects the resistivity of the conducting channel. This effect can be measured as a shift in the applied potential, or simply, the creation of electron-hole pairs on the gate oxide (SiO₂) created by ionising radiation leads to an accumulative effect of positive charges on the SiO₂ and Si-SiO₂ interface. This causes a change in the resistivity of the channel and makes a permanent shift on the threshold voltage ($\Delta V_{\text{th}}$) when a constant electric current passes through the channel, as shown in Figure 1. This shift in the threshold voltage is proportional to the absorbed dose.

![Diagram of MOSFET](image)

**Figure 2-1:** Creation of electron-hole pairs on a MOSFET structure.

Thereafter, a MOSFET detector has been proposed by [52] as a radiation detector for space dosimetry. The amount of $\Delta V_{\text{th}}$ depends on parameters such as the oxide thickness and bias applied during irradiation. It has been found that $\Delta V_{\text{th}}$ for passive
mode (zero bias operation) P-channel MOSFET follows a sub-linear relationship [53]:

\[ \Delta V_{\text{th}} \approx 0.0022D^{0.4}t_{\text{ox}}^2 \]

While in active mode, it almost follows linear behaviour and follows this equation:

\[ \Delta V_{\text{th}} = 0.04Df_t_{\text{ox}}^2 \]

Where \( D \) is the dose, \( f \) is the probability of escaping from a recombination, and \( t_{\text{ox}} \) is the oxide thickness.

It is clear from these equations that \( \Delta V_{\text{th}} \) will be much higher for thick oxide and/or when MOSFET is operated in an active mode with an applied voltage, especially when \( f = 1 \) at high gate bias (\( V_g \)). It has been found that the linearity range may reach 50Gy and the sensitivity of the MOSFET increases with increasing \( V_g \), and although no saturation was detected up to 100V near the breakdown for the tested samples, the sensitivity was about \(~50\text{mV/cGy at 100V operating bias for doses ranging from 5cGy up to 10Gy} \) [54]. This low detection limit makes it a suitable dosimeter for low dose dosimetry like high dose rate (HDR) brachytherapy, build-up region, and surface or skin dosimetry.

The MOSFET has two main issues, a decrease of its sensitivity and an increase in its instability with dose history. This decrease in its sensitivity makes recalibrating MOSFET a requirement after certain accumulated doses, while the
increasing instability increases the error on the MOSFET signal, which makes it difficult to measure low doses [55-57]. Many workers investigated the physics behind these issues; the decrease in sensitivity is based on the trapping ability of the oxide, so the more doses MOSFET receives, the more positive charges are accumulated, so the pre-existing traps on the oxide decrease, leading to a decrease in sensitivity. However, this is based on the main dosimetric parameter and charge build up that causes $\Delta V_{th}$. Furthermore, the increase in instability is based on an increase in the interface traps created by ionising radiation at Si-SiO$_2$ with dose [56, 58]. The charge distribution on the MOS device after irradiation can be seen in Figure 2. The interface traps concentrated on the Silicon side of the Si-SiO$_2$ interface could be broken chemical bonds due to differences in the crystalline material on the Si-SiO$_2$ interface, or impurity atoms created when the device was being manufactured, or radiation induced interface traps [59].

Figure 2-2: Distribution of charges on MOS device after irradiation, adapted from [60].
When the MOS device is operated and the conducting channel is present, those traps can be donor-like or acceptor-like traps. The donor-like traps are positively charged, and if present, create negative shift on $V_{th}$, while the acceptor-like traps are negatively charged, and if present, create a positive shift on $V_{th}$. However, radiation induced interface traps reduce the transconductance of the device, which is the gain or slope of the current voltage characteristics (I-V) curve of the device, Figure 3, where the more radiation on the device, the more interface traps, while the I-V slope continues to decrease [57, 59]. However, the MOSFET dosimeter has many features compared to other detectors, such as its very small size, direct reading, dose rate independent, and integrating dosimeter that can keep a dose history during its life with minimal fading, a simple non-destructive operating circuit with low voltage and minimal power consumption [11, 50, 57]. It is suggested by many authors that MOSFET can be a good alternative to a thermoluminescent detector, TLD [12, 50].
Figure 2-3: The typical effect of interface traps on MOSFET I-V characteristics, adapted from [57].

Thus, for clinical purposes, MOSFET should be calibrated and characterised in parallel with an ionisation chamber for different clinical situations. Calibration must be performed under reference conditions at depth of dose maximum ($d_{max}$) and 10 cm x 10 cm field size, using source to surface distance (SSD) or source to axis distance (SAD) techniques [13, 38, 61, 62]. In addition, the characterisation should cover any parameter that may affect MOSFET response, such as energy, modality (electron or photon), SSD, beam angle, field size (FS), temperature, dose history, build up caps, and any accessories being used within the treatment (i.e., block, tray, wedge, etc) [12, 13, 38, 49, 62, 63].
It is very important to check the linearity range and reproducibility before the initial clinical use [64]. Soubra et al., investigated MOSFET is linear for wide dose range, Thomson Nelson (TN) type TN-502RD MOSFET and TN-502RDM MicroMOSFET show good reproducibility within 1% and good linearity within 0.3% up to 25 Gy [13]. However, the lifespan for these dosimeters as suggested by the manufacturer is about 200Gy or 20,000 mV, or as estimated by others, about 18,500 mV. By the end of its life span, the sensitivity will decrease and the instability will increase, so the dosimeter should be disposed [12]. An angular response is another essential characteristic that must be checked. Ramani et al, [12] observed 15% overestimations on TN MOSFET’s for angles between 140º to 220º. Kwan et al [40] noticed 15% angular responses for RadFET MOSFET, and 2% for CMRP MOSkin. They concluded that this difference came from the build up layer over the RadFET MOSFET sensitive area.

Temperature dependence is also one of the main factors that may influence MOSFET readings because variations in the temperature of MOSFET will cause a shift to the $V_{th}$, especially if the MOSFET has been pre-irradiated. [65-68]. There are many techniques to overcome this problem; one suggested by Buehler et al 1993 was to operate the MOSFET at its thermostable current. This point of operational current has proven to remain unchanged for a temperature ranging from -50C to 125C [66], however, this thermostability point disappears after irradiation [68]. Soubra et al 1994 suggested another solution that is independent of the thermostable point. They used two identical MOSFETs on the same substrate but with different operational irradiation bias’, they read both of them and then subtracted the threshold voltage shifts ($\Delta V_{th1} - \Delta V_{th2}$), so the net shift was independent of temperature [64]. Cheung et
al 2004, suggested calibrating MOSFET and make the reading on the same
temperature either by placing the MOSFET on patient’s skin for 60 s (if calibrated at
patient skin temperature ~31-35 °C), or by making the reading after removing the
MOSFET from the patient’s skin for 120 s (if calibrated at room temperature) [65].

Angular dependence is another issue that should be taken into account when
working with MOSFET dosimeter. It comes from asymmetrical construction of the
detector; this problem can affect MOSFET reading especially for surface, skin, and
interface dosimetry. So, to overtake this problem, a smart design with face to face
dual MOSFET detector has been used by CMRP MOSkin™ to eliminate this problem
with almost zero angular dependence [69]. Other approach to eliminate angular
dependence has been applied by Thomson Nelson™ (currently Best Medical™)
MOSFET by adding epoxy bubble to make its response symmetrical within 360
degree. However, this design cannot be used for skin or build-up region because its
WED has been shifted by the epoxy bubble, and also it was not really eliminating the
angular dependence as expected, due to the non-uniformity of the epoxy [70].

2.3 MOSFET Annealing

The radiation induced effects on the MOSFET structure, as discussed above,
i.e., a positive charge build-up and interface traps, can be annealed as suggested by
some authors using Ultraviolet light, thermal annealing, or electrical annealing [7,
57, 71-76].
2.3.1 Ultraviolet Annealing

The minimum energy required to shift an electron from the Silicon (Si) valance band to the silicon oxide (SiO\(_2\)) conduction band and to recombine with a silicon oxide hole is well known and experimentally estimated to be 4.25 eV [Williams, 1965], which lies within the Ultraviolet range. Williams concluded that this energy is the same for n-type and p-type silicon substrate [77]. Early investigations of the effect of Ultraviolet (UV) light on MOS structures have been carried out by Holmes-Siedle et al, in 1975 using wide energy range 3-14 eV UV light sources [73]. They found that UV light may cause a recombination within the Si-SiO\(_2\) interface, while the oxide traps were not affected at all, this is a result of the short main free path of the photo-injected electrons in the oxide thickness. In recent studies it has been concluded that UV annealing for MOSFET detectors can anneal \(V_{th}\) which is the main clinical parameter used to prolong its use in clinical dosimetry [71]. However, he found that UV light can create switching traps, which include the creation of an interface charge as well, so in this case the instability of the MOSFET signal was expected because the interface traps are the main reason for instability in MOSFET \(V_{th}\). Thus, the feasibility of a UV annealed MOSFET should be investigated in terms of linearity, and reproducibility and instability because all these specific characteristics are very important in radiation dosimetry.

2.3.2 Thermal annealing

MOSFET can be annealed isothermally or isochronally. Isothermal annealing of MOSFET devices means annealing the MOSFET at a certain temperature for a period of time, until the recovery of \(V_{th}\) while isochronal annealing is performed by
fixing the time and changing the temperature until the threshold voltage is recovered. Many authors investigated isochronal and isothermal annealing to repair the device from radiation damage [72-75]. In 1968 Danchenko et al [72], investigated isothermal and isochronal annealing of a p-channel MOSFET transistor, and concluded that isothermal annealing at 200 °C followed by 300 °C for 2hrs, completely annealed the MOSFET to their original $V_{th}$, although it was also noted that some samples annealed below its original $V_{th}$ produced an increase in its transconductance. They also found that the required thermal activation energy to initiate the annealing process was about 1 eV. Kelleher et al 1994, investigated isothermal annealing on a MOSFET radiation detector (RadFET), and found the RadFET can be annealed at 150 °C and re-used again [74]. Yilmaz et al 2008, used thermal annealing at 200 °C for 30 minutes after irradiation to 60Gy; these irradiation/annealing cycles were repeated for up to nine cycles. They concluded that the MOSFET can be annealed and recovered from damage induced by radiation [75].

2.3.3 Direct and pulsed current annealing

The principle of current annealing for medical radiation dosimetry has not been investigated, and generally was investigated less by other authors. Rosenfeld et al [78], applied a pulsed current to a p-i-n diode for fast neutron dosimetry to anneal radiation defects, and concluded that this method caused localised heating and ionisation by dissipating high thermal energy inside the sensitive area. However, in MOSFET devices, when electrons or holes gained kinetic energies greater than the Si-SiO$_2$ energy barrier, they are called hot charges or hot carriers [51, 79]. Those hot carriers can reach this energy if they are affected by a high electric field. This field
could be applied between the source and the drain, or through the p-n junction between the substrate and the source, or between the substrate and the drain. The effect of the injection of hot carriers in p-channel and n-channel MOSFET has been investigated by other authors [79-81]. It has been concluded by Kwok Ng et al. 1983 [79], that for n-channel MOSFET, the injection of hot carriers (electrons), increases the resistivity of the channel leading to a decrease in the drain current, and the creation of interface traps. While in a p-channel MOSFET, a decrease in channel resistivity leading to an increase in the source-drain current [79] was noted, it was also noted that when the current passes through the channel there are no hot holes in a p-channel MOSFET, but there are hot electrons coming from the substrate, and moreover, during the injection of hot carriers, no interface traps were generated on a p-channel MOSFET [79]. This method had been used to anneal and recover the electronic parameters of the Complementary Metal Oxide Field Effect Transistor (CMOS) circuits from radiation defects [76]. However, to measure the radiation and anneal its effect on CMOS or MOSFET devices, a separate study is required to investigate the main dosimetric parameters such as, linearity and sensitivity. Thus, as Kwok Ng et al. 1983 concluded, hot carrier injection traps electrons on the insulator which increases the current on the channel (i.e., reduces channel resistivity) without producing any interface states for p-channel MOSFET, which means there are benefits for using p-channel MOSFET as a radiation dosimeter. The injection of hot electrons to the insulator (SiO$_2$) material will certainly cause a recombination with the radiation induced positive charges on the oxide, causing a decrease on the $V_{th}$ of the MOSFET.
2.4 Dosimetry around interfaces

The planning and monitoring of dose distribution near the interface regions is one of the most challenging aspects of radiotherapy treatment planning systems (TPS). The human body consists of a large number of interfaces between bone, muscle, organs, and air cavities that can affect dose distribution; in addition, foreign materials in the form of prosthetics, metallic implants, and temporary surgical aids, such as tubes or pins, may also be introduced. The lack of electronic equilibrium as the beam crosses heterogeneous regions with different atomic numbers and densities perturbs the dose near the interfaces which makes accurate dose computation a challenge to most current TPS.

2.4.1 The effect of air cavities on dose distribution

One of the early studies involving dose measurement at the air-tissue interface was conducted by Klein et al. 1993. Using 4MV and 15 MV X-Ray beams, which were commonly used for RT, the authors performed their measurements using a parallel plate chamber with an effective volume of 0.5cm$^3$, in order to study the effect of air-water interface heterogeneity. Different air cavities shaped in the form of layers, channels, cubes and triangles were selected according to cavities found within tissues such as the larynx. The data obtained were compared with those from a Monte Carlo (MC) simulation process. In addition, the authors used a semi-conductor diode without a protective cover (bare) to measure the effects of air cavities on the lateral sides around the cavities. The authors observed that while most algorithms could accurately estimate dose distributions in homogenous media, such estimation around a cavity was more difficult, and it could differ by as much as 30% for both the 4MV and 15 MV photons beams, depending on size, depth, and location of the cavity.
However, an underestimation of the dose on the volume beyond the cavity, with a 15 MV photons beam, was larger and reached deeper down, which is why a lower energy RT is recommended for optimum clinical outcomes when used for Larynx radiotherapy [82].

### 2.4.2 The effect of metallic inserts on dose distribution

Das and Kahn (1989) [83], studied the problem of electron backscatter at high atomic number (Z) interfaces, such as metal inserts, and its effect on dose calculation, for beam energies up to 24MV. The authors developed a theoretical formula to take into account the Backscattered Dose Factor (BSDF), which they proposed to apply as a correction factor for the corresponding dose in a homogenous medium. It was proposed that a BSDF would be calculated from a range of parameters such as the equivalent dose in a homogenous tissue phantom, beam energy, field size (FS) at the measurement point, atomic number, width and thickness of the encountered inhomogeneity, and the thickness of the tissue prior to the interface. The authors conducted experiments with beam energies from 4 to 24MV, using a polystyrene phantom, and high Z materials such as: bone, aluminium, stainless steel, copper, tin, Cerrobend, and lead. They plotted changes in BSDF as a function of Z at various beam energies, and noted that the direct proportionality was possibly due to higher secondary electron scattering and fluorescence at higher Z. The range of the backscatter effect was also found to be limited to only a few millimetres, while its magnitude remained almost unchanged at all beam energies. Based on their experiments, the authors developed empirical equations for calculating the BSDF factor for different parameters; materials, beam energies, thicknesses, depth, and field sizes. Later on, Niroomand-Rad et al. 1996 [84], investigated the effects of inhomogeneity at tissue-metal interfaces at 6 MV and 10 MV photon beams, for
head and neck patients. They focused their efforts on patients with titanium based alloys implanted in dental patients, measuring both the dose reduction and effects of enhancement at the interface using radiographic and radiochromic films, and a parallel plate ion chamber. They used titanium plates with two thicknesses to simulate clinical implants, and placed these in a water phantom. They observed results similar to those obtained earlier by Das and Kahn 1989 [83], that dose characteristics are a function of the beam energy, thickness, and Z of the inhomogeneity. In addition, they also observed dose enhancement within 22.5% - 23% for 6 MV photon beam at the proximal interface, although there was dose underestimation on the distal interface, and a slightly lower dose enhancement of ~20% for 10 MV photon beam. The American Association for Physicists in Medicine Task Group 63 (AAPM TG63) report #81 (2003) [85], investigated the dose distribution in the area near the interface of high atomic number hip prostheses in some detail. Beam energies at 6 MV and 18 MV were used with a water phantom and a Co-Cr-Mo slab that was 3cm thick, placed at depths of 5-8cm. The effects of scattered photons and electron disequilibrium were accounted for using approximate methods, such as effective TMR ratios. The report stated that dose perturbations occur near the interface, and extend to a distance that is either greater or lower than the equivalent homogenous medium dose distribution. This perturbation depends on factors such as beam energy, difference in the photon energy transfer coefficient, the atomic number, and the thickness and density of the implant. A dose correction factor CF was defined, made from two components: the BSDF and the Forward Dose Perturbation Factor (FDPF), to account for the exit side of the implant. The report also mentioned various models of hip implants to be used and their physical properties, including alloy composition. The report gives some guidelines and steps
for reducing the effect of dose perturbation, especially at high dose rates, and advised that the maximum dose as calculated by the TPS should be kept below 55Gy within a 1cm thick tissue surrounding the implant. Special beam arrangements such as the “box” and diamond” plans were also considered, and then the results of the Dose-Volume-Histogram (DVH) for these were compared. Several recommendations were made, such as: determining the limitations for heterogeneity correction for each TPS manufacturer and verifying them with phantom measurements, and considering beam arrangements, in conjunction with the physicist, that avoid irradiating the prosthesis, and in-vivo measurements with diode, film, or TLD after first fraction of the treatment, especially for verifying the exit dose. The report mentioned that enhancing the dose at the bone-metallic interface may cause necrosis of the bone, the probability of occurrence increases with each dose, and complications appear for doses of 50 - 65Gy.

Ravikumar et al 2004 [86], investigated the effect of metallic inhomogeneities such as aluminium, mild steel, copper, cadmium and lead for Cobalt-60 gamma rays and X-ray beam energies of 6 MV and 18 MV. They reported that the BSDF value attained its saturation point a few millimetres away from the inhomogeneity, and the atomic number of the metal determined the thickness, at which value is reached, with lesser thickness being reported for higher Z values. Further, the thickness at which the saturation value is attained also depended on the photon energy, and the quantity of backscattered photons increased at lower energy compared to higher energy photons. The authors also observed that a higher atomic number of the metal resulted in a higher BSDF, but not in its significant variation. On the other hand, they observed that effects of dose perturbation only persisted for a few millimetres before
the metallic interface, and hence concluded that the range of backscattered electrons was quite limited.

As a result of these and many other studies, the AAPM Report No. 85 (2004) [87] discussed in detail the need to correct tissue inhomogeneity, for treatment planning algorithms to be involved, and correction factors for different situations such as air cavity, lung, bone, and high-Z materials. The report recommended a dose distribution accuracy of between 1% and 2%, with the tissue inhomogeneity correction component contributing an uncertainty of 1.4% at most. The report observed that air cavities presented the highest dose perturbation, which is of clinical concern, especially for tumours extending to the surface. The report suggests several recommendations, such as, in the vicinity of soft tissues and air cavities, superposition/convolution algorithms, or MC simulation based algorithms can be used.

2.5 Convolution – Superposition Treatment Planning Algorithms

2.5.1 The Pencil Beam Convolution (PBC), Collapsed Cone (CC) and the Anisotropic Analytical Algorithms (AAA)

Over the last decade, from the mid 1990s to 2000s, in addition to measurement based algorithms, three model based TPAs were commercially available, the Pencil Beam Convolution (PBC) algorithm, the Collapsed Cone algorithm, and the Anisotropic Analytical Algorithm (AAA). The PBC was within acceptable limits for measurements based on the water phantom, although it was inaccurate if some heterogeneity was present. It corrected for one dimensional radiation transport, taking into account the density along with the depth, while neglecting the spread of lateral energy and backscattered radiation [88-91]. The dose calculation of the PBC
The algorithm is based on the convolution of total energy released per unit mass (Terma) with the energy deposition kernel (K).

\[
D(x, y, z) = \int \int \Psi(x', y') \frac{K_w(x - x', y - y', z)}{\rho} dx' dy'
\]

Where \(D(x, y, z)\) is the dose at points \(x, y, z\), \(\rho\) is the medium density, \(z\) is the depth where the dose was deposited, \(\Psi(x', y')\) is the energy fluence, and \(K_w\) is the tabulated energy deposition kernel [90].

The collapsed cone calculations algorithm takes into account the variations in density in three dimensions at the calculation voxel in water located at point \(r\), with its energy deposition kernel \(h(r - r')\), the dose can be calculated as the convolution between the terma \((T)\) and the kernel \(h(r - r')\), as in the equation below [89]:

\[
D(r) = \int \int \int T(r) \frac{h}{\rho} (r - r') dr'
\]

In 2005, AAA TPA was developed which is mainly based on the convolution-superposition algorithm that combines a Monte Carlo based energy spread kernel data and the measured data as well, to create a model that calculates the absorbed dose. The corrections for inhomogeneities taken into account by calculating the scaled scattered energy kernel \((K_\beta)\) factors of the electrons and photons anisotropically, which include many components that affect the energy fluence in the beamlet, such as primary photons, contaminant electrons coming from the flattening filters, ion chambers, jaws and other accessories [92, 93].
As shown in Figure 2-4, the dose $D$ at point $X$, $Y$, $Z$, calculated by AAA [94] is:

$$D_{\beta}(X,Y,Z) = \Phi_{\beta} \times I_{\beta}(Z,\rho) \times \int \int K_{\beta}(x',y',x,y,Z;\rho) \, dx' \, dy'$$

Where $\Phi_{\beta}$ is the energy fluence, $I_{\beta}$ is the density function, $K_{\beta}$ is the scattered energy kernel.
2.6 Deterministic dose planning Algorithm

2.6.1 Acuros XB TPA

Recently, the linear Boltzmann transport equation (LBTE) solver dose calculation based algorithm Acuros XB, has been introduced onto the Eclipse TPS. It is based on a deterministic solution of the LBTE that governs radiation transport in a medium. The deterministic solution to the time independent linear Boltzmann transport equation (LBTE) was first used by a group of researchers at Los Alamos National Laboratories to simulate radiation transport within matter; it was based on a general software code called the Attila LBTE grid solver [95]. The Attila software was used later to simulate radiation transport on radiation therapy by Gifford et al, 2006 and Vassiliev et al, 2008 [96, 97]. The coupled time independent linear Boltzmann transport equations are:

\[ \hat{\Omega} \cdot \nabla \Psi^\gamma + \sigma_t^\gamma \Psi^\gamma = q^{\gamma \gamma} + q^\gamma \]

\[ \hat{\Omega} \cdot \nabla \Psi^e + \sigma_t^e \Psi^e - \frac{\partial}{\partial E} S_R \Psi^e = q^{ee} + q^{re} + q^e \]

Where:

- \( \Psi^\gamma \) and \( \Psi^e \) are angular photon and electron fluence respectively,

- \( q^{\gamma \gamma} \), \( q^{ee} \) and \( q^{re} \) are photon to photon, electron to electron, and photon to electron scattering,

- \( q^e \) and \( q^\gamma \) are external electron source, and external photon source,

- \( S_R \) is the restricted collisional and radiative stoping power,

- \( \sigma_t^\gamma \) is the macroscopic photon total cross section,
\( \sigma_{t}^{e} \) is the total cross section of a macroscopic electron.

Once the above two equations for electrons fluence \( (\Psi^{e}) \) are solved using the finite elements method (Wareing 1999), the dose \( D(\vec{r}) \) can be calculated as:

\[
D(\vec{r}) = \int_{0}^{\infty} dE \int_{4\pi} d\Omega \frac{\sigma_{dep}^{e}(\vec{r},E)}{\rho(\vec{r})} \Psi^{e}(\vec{r},E,\hat{\Omega})
\]

Where \( \sigma_{dep}^{e} \) is the cross section of the macroscopic energy deposition and \( \rho \) is the density of the voxel being calculated.

Acuros® XB is an optimised LBTE grid solver for the application of external radiation therapy, based on the Attila code. It gives an exact and direct solution to the LBTE as a function of parameters such as, energy, position, and angle. This numerical solution may have some systematic errors as a result of parameter discretization and the treatment of cross section data with particles. The Monte Carlo (MC) based algorithm solves the LBTE stochastically with uncertainties on their number of interactions (histories). However, MC simulation takes a long time to reach acceptable uncertainties while Acuros® XB with its explicit solution is much faster at satisfying the clinical requirements which demand time and accuracy management [93].

However, Monte Carlo based TPAs are the standard and most accurate method for treatment planning, but they are slow in comparison to the current commercial convolution-superposition TPAs [98]. There has been some progress in speed using Macro Monte Carlo methods [46, 70].
2.6.2 Dosimetric benchmark of TPA

Monte Carlo simulation provides a stochastic calculation of radiation transport in matter. It can be claimed to be the standard for treatment planning algorithms as many authors used MC simulation or ion chamber measurement to benchmark TPAs [90, 98-104].

Bragg and Conway (2006) investigated the accuracy of the Anisotropic Analytical Algorithm (AAA) for calculating dose distribution in both homogenous and heterogeneous media. The authors used water and a variety of complex geometries to measure the homogenous dose distribution, while for the inhomogeneous part they used bone, lung, and low density slabs. For photon beams of 6 MV and 10 MV, the authors found an agreement between calculated and measured values to within 1.5% beyond \(d_{\text{max}}\), while for rectangular fields, isodoses agreed to within 2mm, but for inhomogeneities, the agreement was within 2.5% for bone equivalent material. As a result, the authors concluded that AAA achieved better accuracy than the pencil beam convolution models being used at the time, especially when used for correcting low density inhomogeneity [99].

To benchmark AAA, Gagne and Zavgorodni (2007) [90] compared AAA and Pencil Beam convolution (PBC) algorithms in dose calculation using a water-lung phantom, and photon beams of 6 MV and 18 MV at different configurations. They used the Eclipse TPS with AAA algorithm and also the Pencil beam convolution algorithm from Varian Medical Systems, and the benchmark chosen was a 2D dose distribution with MC simulation. The authors observed that while both AAA and PBC algorithms correctly represent dose distributions in water within 2mm of the penumbral dose, AAA displays better penumbral dose modelling as calculated by
MC. PBC shows greater discrepancy in modelling across the entire lung block, particularly for an 18 MV photon beam and 4 x 4 cm$^2$ field size combination. As for inhomogeneity, the agreement for AAA was within 4% for all cases, while for PBC it ranged from 3.5% to 11.2%. In general, AAA was much better than PBC for both the vertical and oblique interfaces modelled with minimal deviation from the acceptable limit. They concluded that AAA models the penumbral region in water, lung, and the interface regions more accurately than PBC.

Some authors used *in-vivo* measurements based comparison with small and thin detectors like TLD and film to verify the TPA. Davidson et al. (2008) compared five IMRT TPSs, three of them using AAA and two using the PBC algorithms. The beam energy used was 6 MV, while the phantom was water filled with synthetic heart, lung, and spinal cord materials to simulate tissue inhomogeneity. A nylon target, representing a tumour, was placed inside the left lung in the medial-anterior direction, and thermoluminescent detectors and radiochromic films were used as dosimeters. The authors reported that AAA modelled the dose distribution more accurately than PBC in low density regions such as the lung, it was within %5 /3-mm, whereas the other PBC TPSs overestimated the dose in the target and underestimated the penumbral dose as a result of their heterogeneity correction methods[105]. Ono et al. (2010) studied the accuracy of AAA in dose calculations using 6 MV and 10 MV photon beams with two phantoms that include different interfaces – water-lung, and water-lung-bone. By comparing the Percentage Depth Dose (PDD) curves simulated by AAA and MC methods with those obtained through ion chamber measurements, the authors observed that these agreed well for both field sizes of 4 x 4 cm$^2$ and 10 x 10 cm$^2$, to within 3% of observed values for the water-
lung phantom. However, with the 10 x 10 cm² field, AAA was within 5% deviation for 6 MV and 4.1% for 10 MV photon beams respectively, while the MC calculations were in uniform agreement for all cases [100].

Acuros® XB, a grid based Boltzmann equation solver, was tested for inhomogeneity dose predictions at 6 MV and 18 MV photon beams by Vassiliev et al. (2010). The authors compared Acuros with MC results using a phantom constructed from tissue, bone, and lung, equivalent materials as well as a breast treatment plan. The authors observed that when the dose exceeded 10% of the maximum dose, Acuros® XB predicted local dose levels to within 2% or 1mm distance-to-agreement of MC simulations, under similar conditions for breast treatment the prediction was within 2% or 2mm, but in all other cases, the prediction was within 1% of the maximum dose. However, according to the authors, the advantage of Acuros was its fast calculation time; in their case it was 5 minutes on a fast computer, which makes it suitable for clinical conditions as well as arc therapy treatment where a large number of beams are used [101].

Fogliata et al. (2011) carried out further validation of the Acuros® XB algorithm, this time against AAA, using four photon beams – 6 MV and 15 MV from a Varian LINAC 2100 iX and 6 MV and 10 MV from a TrueBeam LINAC. All the measurements were carried out in water using a PTW-MP3 phantom. They reported that Acuros® XB agrees closely with calculations from AAA as well as actual measurements, within 1% for simple, homogenous test cases, and 2% for wedged cases [103].
Bush et al. (2011) benchmarked the accuracy of dose calculation for Acuros® XB with MC and compared their results to the AAA in homogenous and heterogeneous geometries using photon beams of 6 MV and 18 MV. The phantoms used for this purpose were homogeneous and heterogeneous with lung, bone, and air inhomogeneities. The authors noted good agreement between the two methods for homogenous regions, within ±1.9%, however, in the case of inhomogeneity they observed an agreement with MC for lung within ±2% and ±2.9% for normal and low density lung, respectively, and for air cavities the agreement was within ±1.5% to ±4.5%. They concluded that one of the most significant advantages of using Acuros® XB for arc therapy is where it can become significantly faster than convolution based algorithms, and with better accuracy [102].

Han, et al, (2011) used homogenous water based as well as multi-layer phantoms made from materials equivalent to tissue, bone, and lung, to compare the performance of Acuros® XB with MC, as well as two popular clinical algorithms – AAA in Eclipse and the Collapsed Cone Convolution (CCC) in Pinnacle treatment planning system (Philips Radiation Oncology Systems, Fitchburg, WI). They obtained depth dose and lateral dose profiles for both phantoms at photon beams of 6 MV and 18 MV, and followed by the AAPM TG53 guidelines to perform 3D gamma index analyses in the slab regions. They found that the dose calculations by Acuros® XB and MC were closer than either AAA or CCC, with particular improvements in the bone and lung equivalent materials. Acuros® XB also performed better than AAA and CCC for predicting electron backscatter at the interfaces, and hence the Acuros® XB performs better dose prediction calculations than either of these two
algorithms in heterogeneous body regions such as the lung, or the head and neck [104].

2.7 Detectors for interface dosimetry

Many detectors have been used for interface measurements [41, 45, 69, 82-84, 86], but all of them must have a thin, sensitive area with very thin, or no build-up material. This is to ensure that the effective point of measurements is close to the interface, and to avoid averaging the detector sensitive volume or to avoid any build-up/build-down (dose gradient) on the detector cap. Examples of these detectors on skin dosimetry, as mentioned earlier are: parallel plate ionisation chamber, radiochromic film, thermoluminescent detector (TLD), and MOSFET.

2.7.1 Parallel plate ionisation chamber

One of the parallel plate ionisation chamber is Attix, which is an air filled parallel plate ionisation chamber with a flat plane towards the beam and an effective point at the surface of the air cavity. It is usually made from materials that are equivalent to tissue, and has a very thin window to allow the dose at the interface region to be measured. However, it does has some limitations, that it cannot be used with patient skin dose measurements because it has a polarising voltage. It is also very large (6cm diameter for Gammex RMI 449), so it cannot be used for fields that are smaller than its diameter (see Figure 2-5), and its finite thickness and large size would also increase the skin dose to the region, if placed on a patient. Another parallel plate chamber is the extrapolation chamber, which is almost similar to Attix chamber but with variable volume (see Figure 2-6), which can be adjusted/extrapolated to zero volume. So, it is perfect for surface dose and build-up region dosimetry, but, it is
quite complicated to get a measurement with it for clinical purposes, so it is good for research [106].

Figure 2-5: Gammex RMI 449 Attix parallel plate Chamber.

Figure 2-6: PTW extrapolation chamber, adapted from ptw.de.
Many authors used this type of chambers to measure the dose on the build-up region and around the interfaces, or to benchmark the treatment planning systems calculation on these regions [27, 30, 82, 100, 106-108].

2.7.2 Radiochromic Film

Radiochromic films have been used by several researchers for build-up, skin, and interface dosimetry [3, 32, 39, 41, 44, 46, 109]. As it is almost tissue equivalent ($Z \sim 7$), with almost flat energy response over wide therapeutic photon beams, especially for the newly developed types Gafchromic EBT and EBT2 [110], although its thickness shifts the effective point of measurement several microns from the surface or the interface [3, 46, 111]. Thus, an overestimation of the dose at the interfaces is expected and should be taken into account [83, 105]. The use of Gafchromic EBT2 film around air, steel, and lung interfaces will be discussed in more detail in Chapter 6.

2.7.3 Thermoluminescent detector

Several investigators used TLD for in-vivo or interface dosimetry to verify their TPS or study the effect of an inhomogeneous medium on dose measurements [105, 107, 112, 113]. However, its size prevents it from getting accurate measurements, which means they are getting an extrapolation or average dose value.

2.7.4 MOSFET detector

A MOSFET dosimeter is an interface dosimeter that produces good spatial resolution as point detector, it can be used as face up or face down or as edge-on detector for
high resolution measurements (~1μm) [114, 115]. The MOSFET energy dependence is negligible for therapeutic dose range [49, 116]. Several investigators have used it for surface and build-up dosimetry [11, 41, 69, 117]. The use of a MOSFET dosimeter for TPS verification for interface dosimetry with different types of inhomogeneities will be conducted during this thesis. A detailed study is given in Chapter 6.
3. DOSIMETRIC PERFORMANCE OF MOSKIN DOSIMETERS ANNEALED WITH ULTRAVIOLET PHOTONS AND HEAT FOR CLINICAL REAPPLICATION

3.1 INTRODUCTION

Modern advanced radiation therapy techniques demand precise and reproducible dose measurement, which in turn, requires a very reliable and stable radiation dosimeter. While an ionisation chamber is well recognised and accepted as the gold standard in radiation dosimetry, the MOSkin dosimeter, a Metal Oxide Semiconductor Field Effect Transistor (MOSFET) based dosimeter [7, 52] has found applications in clinical practice as an alternative point dosimeter for precise measurement of the skin dose [11, 118]. The main advantage of the MOSkin dosimeter is that it minimally perturbs the radiation treatment field, so in principle it makes an ideal in vivo quality assurance tool for common forms of radiotherapy (e.g. Intensity Modulate Radiotherapy (IMRT)). Such a routine application is cost prohibitive because it would require several MOSkins positioned around the patient to measure the skin dose at each IMRT treatment field angle. Moreover, the MOSkins are usually disposable, single use devices, for reasons discussed below.

The MOSkin operates as a radiation dose integrator, and is very small in size and has a real time readout capability. For dose increments in the order of a single radiation
treatment fraction in conformal radiotherapy, the MOSkin’s response is linear. The MOSkin has a well characterised effective radiation lifetime for larger total dose irradiation that depend on the specific design of the MOSkin [69]. Typically, as the total absorbed dose increases, the response of the MOSkin becomes non-linear, with the non-linearity presenting itself as a reduction in radiation sensitivity in the order of 5% per 1500 cGy. This non-linearity is well known and is due to a limit in the positive charge build up at the Si-SiO$_2$ interface of the gate contact [52, 55]. The MOSkin therefore requires a dose recalibration to account for this change in sensitivity if it is to be reused clinically, otherwise it must be disposed off and a new MOSkin calibrated and used. Such a practice requires many MOSkins which ultimately limits the uptake of these dosimeters in routine clinical practice due to cost limitations.

The average energy required to move an electron from the Si valance band to the SiO$_2$ conduction band, which can be recombined with an oxide hole, has been experimentally estimated as 4.25 eV [Williams et al. 1965] which lies within the UV range. Williams concluded that this energy is independent of the Si type, either n-type or p-type [77]. UV annealing of MOS capacitors and transistors has been investigated using a UV light with a wide energy range 3-14 eV [73]. Furthermore, some investigators have used the same principle to erase Floating gate MOSFETs (FAMOS) because it has some similar structures such as a metallic gate, sensitive oxide, and Si-substrate [119-121]. Recently, it has been shown that UV annealing for MOS dosimeters is a promising method for annealing the radiation defects in these detectors [71].
Thermal annealing at specific temperature (isothermal), has been used and suggested by some authors as a method of repairing radiation defects on MOSFET detectors [7, 57, 71, 73-75]. They have shown that isothermal annealing at a specific temperature $150^0C$ or $300^0C$ can recover the sensitivity of MOSFET perfectly, [57, 73, 74]. Danchenko et al., [72] investigated isothermal annealing of MOSFET devices and found that the thermal activation energy required to initiate the annealing process is about 1 eV.

The newly developed packaging of a radiation dosimetry probe based on MOSFET and called MOSkin™ is a P-channel MOSFET designed at the Centre for Medical Radiation Physics, University of Wollongong, such that it allows UV light to reach the gate oxide. It has been achieved by a CMRP drop in design where the MOSFET chip is embedded in a KAPTON pigtail using a polyamide film carrier which simultaneously provides electrical connections to the MOSFET chip and protection from the environment. The polyamide carrier used here is thin and transperable for UV light. Two different layers of polyimide were used [118]. The MOSkin™ with layer of polyimide can also be annealed at a raised temperature of $150^0C$ without damaging the detector. Figure 3-1 shows the design of the MOSFET packaging (MOSkin)
Figure 3-1: Schematic diagram showing the MOSFET chip (blue) packaging design (MOSkin) embedded in a KAPTON pigtail (green) and covered by a polyamide carrier (red), and a) A dual thick polyamide over layer, b) A single thin polyamide over layer.

The MOSkin™ has a 0.55 µm thick gate oxide and a polyimide over layer of a thickness that results in the water equivalent depth (WED) of dose measurement of 0.07 mm. Dose measurements with the MOSkin™ at the surface or on the skin are therefore easily achievable and reliable. This chapter one of the thesis focuses on the UV and isothermal annealing of CMRP MOSFET (MOSkin™) which was investigated to examine the possibility of extending the use of MOSkin dosimeters in routine in-vivo quality assurance of radiation therapy techniques such as IMRT. In addition, the recovery of the threshold voltage ($V_{th}$), the effect of different over layer encapsulation materials in Figures 3-1 A) and B), the UV annealing behaviour, linearity, and the sensitivity of MOSkin™ is investigated for clinical use.
3.2 Materials and methods

For the UV annealing related studies, six MOSkin™ detectors were used with different materials covering the gate oxide to investigate the role of the over layer when the MOSkin dosimeter is being annealed. Over layers included 20 µm thick polyimide carrier material (MOSkin Type#1 and #6), 3M clear plastic tape, 35 µm thick (MOSkin Type #2), 3M black plastic, 30 µm thick (MOSkin Type #3), and Kapton polyimide, 60 µm thick, (MOSkin Type #4 & MOSkin Type #5), Figure 3-2. This particular design allows us to compare the relative absorption of UV light within the various encapsulations during UV annealing, the effects of any fading at room temperature, and the effect of UV light on the un-irradiated MOSkin detectors. The MOSkins were irradiated to a total dose of 15 Gy, (3Gy per fraction) using a 6 MV X-ray field from a Linear accelerator (Varian 2300EX) at the Illawarra Cancer Centre, Wollongong. The MOSkins were positioned at a depth of 15 mm (d_{max} for this energy) on the central axis of a 30x30x30 cm³ solid water phantom. The source to surface distance (SSD) was 100 cm and the radiation field was 10 x 10 cm². The threshold voltage readings were measured after each fraction of irradiation. The V_{th} was measured in a consistent manner by reading out 30s after each irradiation fraction, and then waiting ~1.5 minutes between each dose fraction. It should be noted that such a strict procedure is not normally necessary if the MOSkins are only used for a single radiation lifetime. Our preliminary data indicated that the above readout protocol provided consistency between measurements of the annealed MOSkins and prevented uncertainties arising from short time fading effects, so it was used for all the measurements taken during this study. The instability of V_{th} for all detectors before irradiation was ±1 mV.
Figure 3-2: Four MOSkin detectors with different encapsulation materials under UV light.

The UV annealing source was a Philips G8 T15 low pressure Mercury vapour UV-C filtered lamp with a dominant UV photon energy of 4.92 eV. The detectors were kept at room temperature and monitored for 36 hours before annealing for any fading. The annealing process was repeated with $V_{th}$ monitored until the original threshold voltage of each detector was reached. However, the detectors with encapsulation material that absorbs UV light, (MOSkin#3 and MOSkin#4), were kept under the same conditions, under UV light, to check the change in sensitivity relative to other detectors. Detector MOSkin#5, was kept away from the UV light after irradiation, to compare its response and sensitivity to other UV annealed detectors. The instability of the threshold voltage of the MOSkin detectors was estimated after each irradiation by taking the difference between two consecutive readings after 30s, three times. This cycle of irradiation and annealing was repeated consecutively, three times.

Three MOSkin™ detectors were used for the thermal annealing studies, with “Kapton” polyimide tape covering the gate oxide. All the detectors were irradiated
under the same conditions as the UV annealed MOSkins mentioned above. The MOSkin detectors were also kept at room temperature for 36 hours after irradiation. The isothermal annealing technique is described in reference [74] and was used in conjunction with a Carbolite MODEL, (Sheffield, England) temperature controlled furnace and operated at 150°C. The detectors were in direct contact with a ceramic sheet to make a good thermal contact with the hot surface of the furnace. The threshold voltage of each detector was monitored by taking a reading after the detector cooled down to room temperature. This cycle was repeated many times until the threshold voltage of the detectors reached their initial threshold voltage, or until the next irradiation experiment. The dosimetric parameters associated with the annealed MOSkins, including dose linearity and sensitivity, were investigated in each irradiation-annealing cycle.

3.3 Results and discussion

3.3.1 UV annealing

For the first cycle of irradiation and annealing, all the MOSkin™ detectors, except for MOSkin#6, were irradiated to a total absorbed dose of 15 Gy. Each detector had a linear response with a dose up to 15 Gy, within ± 3%, and their average sensitivities after 15Gy dose were 2.54 mV/cGy. Figure 3-3 shows the 1st UV annealing cycle results for the five MOSkins (#1 – #5). The recovery of MOSkin is demonstrated by a reduction in the threshold voltage with UV light exposure time. Figure 3-3 also shows that MOSkin #1 and MOSkin #2 appeared to be annealed at different rates. This is due to a variation in the intensity of the UV light penetrating the respective polyimide over layer of the MOSkin. MOSkin #1 and MOSkin #2 were fully annealed in 13 hrs and 45 hrs respectively. Detector #5, which was not exposed to UV light, shows a similar room temperature annealing curve compared to UV
irradiated detectors #3 and #4, which indicates there is a significant amount of UV absorption within the over layer materials in these MOSkins. The UV annealing process for MOSkin#3 and MOSkin#4 was therefore considered too long for any clinically practical purposes. The un-irradiated control MOSkin #6 was kept for 72 hrs under UV light, and no change was observed in the threshold voltage ($V_{th}$) of this MOSkin.

Figure 3-3: UV Annealing results for the five MOSkins in the 1st annealing cycle showing a reduction in the threshold voltage over time. The error bars are too small, within ±0.003V (±3mV).

The second irradiation and annealing cycle revealed that the response of UV annealed MOSkins’ (#1, #2) were still linear with dose when irradiated to a further 15 Gy, but with a 6.5% reduction in sensitivity compared to their initial values (as shown in Figure 3-4, Figure 3-5) after each cycle. The $V_{th}$ signal instability however,
increased significantly to ±11 mV. MOSkin #3 and MOSkin #4, which were not affected by UV light, had almost similar sensitivities compared to MOSkin #1 and MOSkin #2, and yet their average \( V_{th} \) signal instability increased to ±6 mV. The subsequent annealing times for MOSkin #1 and MOSkin #2 were 10 hrs and 37 hrs respectively, which was shorter than the 1st annealing cycle.

The third irradiation and annealing cycle indicated that the response of each detector was still linear, with the dose as shown in Figure 3-4 for MOSkin #1. A similar annealing result was obtained for MOSkin #2, Figure 3-5. MOSkin #1 and #2 showed a 10% decrease in sensitivity compared to their initial sensitivities, while the average instability in the \( V_{th} \) increased from ±1 mV before irradiation to ±26 mV after three cycles of irradiation and annealing. MOSkins #4 and #5 showed a similar increase in their average \( V_{th} \) instability of ±8 mV after three cycles of irradiation and annealing, and a decrease in their sensitivities which were never recovered with UV because the over layer suffered from poor UV transparency. The annealing time required for the \( V_{th} \) to recover its original value for MOSkin #1 and MOSkin #2 was 8.5 hr and 58.75 hr respectively, after the 3\(^{rd} \) cycle, as shown in Figure 3-6 and Figure 3-7.

The reason for the increased instability in the \( V_{th} \) measurement after UV annealing could be the result of defects in the radiation leading to the creation of interface traps. It is well known that MOSFET transconductance decreases with an increase in the interface traps [57, 59], so to obtain an indirect approximate estimation of the creation of the interface traps, the I-V characteristics of MOSkins #1 and #2 were recorded for the two UV annealed samples to calculate the transconductance from the graph, which is equal to the slope of the I-V curves. The change of
transconductance with X-ray irradiation and UV annealing in every cycle for MOSkin #1, can be shown in Figure 3-8.
Figure 3-4: The dose response for 3 annealing cycles for MOSkin #1.

Figure 3-5: The dose response for 3 annealing cycles for MOSkin #2.
Figure 3-6: UV annealing rate for MOSkin #1 in the three cycles. The error bars are too small and cannot be seen in the graph.
Figure 3-7: UV annealing rate for MOSkin #2 in the three cycles. The error bars are too small and cannot be seen in the graph.
Figure 3-8: Transconductance of MOSkin #1 after consecutive cycles of X-ray irradiation and UV annealing.

It is very clear from the graph (Figure 3-8) that transconductance decreases, which mean that the interface traps, increased in each X-ray irradiation or UV annealing, until it almost reached saturation at the last UV annealing.

At first glance, Fig. 5 presenting the kinetics of the UV annealing of MOSkin #1, indicates that the total annealing time appears to decrease for the 2\textsuperscript{nd} annealing cycle compared to the 1\textsuperscript{st} and 3\textsuperscript{rd} cycle, compared to the 2\textsuperscript{nd}.

This again supports our model that showed the UV recovery of $V_{th}$ is not really associated with annealed (remove) holes trapped in border traps in SiO\textsubscript{2} during irradiation, rather it is supported by the fact that the MOSkin decreased in sensitivity
with each cycle. This decrease in sensitivity is due to an accumulation of holes
trapped on the border traps leading to a sub-linear dose response by the MOSFET.

Irradiation with UV leads to the creation of defects that act like electron traps in
SiO$_2$ and an increase in the interface traps created by X-ray radiation. Electron traps
with a negative charge on them compensate for the effect of a positive charge on the
border traps and a reduction of $V_{th}$ that appeared as a pseudo recovery of $V_{th}$, i.e. UV
annealing.

A decrease in the UV irradiation time for $V_{th}$ recovery after each cycle of 15 Gy
irradiation can be explained by a reduction in the increment of a trapped positive
charge after each irradiation observed in the sub-linear MOSkin dose response. It
requires a reduced increment of negative charge (electrons traps) to compensate for
the increment of positive charge, i.e. the recovered $V_{th}$. Figure 3-6 confirms this
possibility where the time for a “pseudo” recovery after each cycle of X-ray
irradiation with 15 Gy dose has been reduced. Moreover, a reduced sensitivity to
radiation by the MOSkin after each cycle (Figure 3-4 and Figure 3-5), is additional
confirmation of a “pseudo” recovery of the MOSkin after UV annealing. This model
is supported by [71, 73, 122] where the UV irradiation does not affect the trapped
positive charge.

The essential increase in instability by MOSkin #1 and #2 with X-ray irradiation
and the UV annealing cycle is associated with increasing interface traps introduced
by UV irradiation. This result is supported by a decreasing transconductivity of the
MOSkin after each irradiation and annealing, as measured from the I-V characteristics and presented in Figure 3-8.

The energy of these traps is different compared to the interface traps induced by
X-ray radiation. This can be explained by the ease of recharging under room
temperature that increased the instability compared to the UV one that was not irradiated (MOSkin #5), or MOSkin #3 & #4 with UV protected over layers.

Figure 3-9: Annealing time with cycle number for MOSkin #1.
Figure 3-10: Variation in the MOSkin sensitivities in three cycles for the irradiated and UV annealed detectors.

An increase in the annealing time of MOSkin #2 compared to MOSkin #1, is due to the degradation of the clear plastic tape by UV irradiation, shown by the tape becoming discoloured as the exposure to UV increased. The average sensitivity of MOSkin #1 and MOSkin #2 decreased with each cycle by about 10% after receiving 45 Gy in the three cycles. The response of MOSkin #1 and MOSkin #2 can be seen in Figure 3-4 and Figure 3-5 and the sensitivities for all the irradiated detectors in all cycles are shown in Figure 3-10.

3.3.2 Thermal Annealing

Thermal annealing of three MOSkins by irradiating them with a dose of 15Gy and annealing them under constant temperature was investigated.
In the 1st cycle of thermal annealing, all the virgin detectors, MOSkin #1a, MOSkin #2a, and MOSkin #3a had a linear response up to 15 Gy, where their initial sensitivities at 300cGy were 2.59 mV/cGy, 2.69 mV/cGy, and 2.69 mV/cGy respectively, as shown in Figure 3-11. Their average $V_{th}$ instability was within ±3mV at the end of the three cycles of irradiation. Isothermal annealing was performed by keeping the detectors at room temperature for about 36 hrs, to check annealing (fading) at room temperature. The temporary kinetics of isothermal annealing at $T=150^\circ C$ for MOSkin#1a, MOSkin#2a, and MOSkin#3a can be seen in Figure 3-12. Two clear components of the recovery of the $V_{th}$ with essentially different time constants for each MOSkin were observed. The first component is fast and led to a substantial recovery (drop) of the $V_{th}$ for all the MOSkins in the first hour of thermal annealing. The second component is much slower so the thermal annealing recovery was more logarithmic. The total time for thermal annealing leading to 95% recovery of the $V_{th}$ was 75hrs at $T=150^\circ C$. 

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Figure 3-11: The dose response for the three virgin MOSkin detectors in the 1st cycle.

The error bars are too small and cannot be seen in the graph (±1mV).
Figure 3-12: The isothermal annealing behaviour of the three MOSkin detectors in the 1st cycle (T=150°C). The error bars are too small and cannot be seen in the graph (±1mV).

The response of all three detectors in the 2nd Cycle was still linear and their sensitivities were very close, as in the 1st cycle, they are 2.597 mV/cGy, 2.627 mV/cGy and 2.603 mV/cGy respectively, with about +0.27%, -2.34%, and -3.13% deviation from the 1st cycle, see Figure 3-13. Those annealed detectors were very stable, with about ±4mV instability in measured $V_{th}$. During isothermal annealing the threshold voltage ($V_{th}$) for MOSkin#1a, MOSkin#2a, and MOSkin#3a almost recovered after about ~100hr of annealing.
Figure 3-13: The dose response in the 2nd cycle for the three MOSkin detectors. The error bars are too small and cannot be seen in the graph (±2mV).

The annealing process for the 2nd cycle also has two timing components, fast and slow, just like the 1st cycle, and is shown in Figure 3-14. While all MOSkins had similar temporal behaviour of $V_{th}$ recovery, there was some variation in the trend of $V_{th}$ recovery between each MOSkin. These variations could be related to a non-uniform temperature gradient inside of the furnace. MOSkin#1 took about 63 hr, and the others more than 100 hr, however full $V_{th}$ recovery was not achieved for MOSkin#2 and MOSkin#3 after 100hr of annealing.
Figure 3-14: The annealing process for the three detectors in the 2\textsuperscript{nd} cycle. The error bars are too small and cannot be seen in the graph (±2mV).

The 3\textsuperscript{rd} thermal cycle showed that MOSkin\#1, MOSkin\#2, and MOSkin\#3 were still linear in their response even after receiving a total of 45 Gy of radiation over the three cycles, as shown in Figure 3-15. The sensitivities for these detectors were 2.57 mV/cGy, 2.66 mV/cGy, and 2.69 mV/cGy respectively, and the percentage of recovery of sensitivity and $V_{\text{th}}$ for each cycle of the isothermal annealing compared to their initials values is shown in Figure 3-16 and Figure 3-17 (a similar graph should be presented for $V_{\text{th}}$ recovery in each cycle). The instability in the $V_{\text{th}}$ measurement for each MOSkin detector increased from ±1mV to ±4mV (three standard deviations), at the end of the 3\textsuperscript{rd} cycle. Taken together, these significant features make the MOSkin\textsuperscript{TM} a good candidate for reuse in radiotherapy because we have shown they are easy to anneal without damaging the detector, which is better than other commercial MOSFET detectors that were investigated previously [74].
Figure 3-15: The dose response for three MOSkin detectors during the 3rd cycle. The error bars are too small and cannot be seen in the graph (±4mV).

The third annealing performed to check the behaviour of the detectors after 45 Gy is shown in Figure 3-18. The annealing of $V_{th}$ for the three detectors took longer than the 1st and 2nd cycles. This could be related to an increase in the number of radiation induced traps in the SiO$_2$ and on Si-SiO$_2$ interface, with a total radiation absorbed dose [71, 74, 75].
Figure 3-16: The change in threshold voltage $V_{th}$ for the three MOSkin detectors for three cycles of irradiation annealing.

Figure 3-17: The change in sensitivities for the three MOSkin detectors for three cycles of irradiation annealing.
3.4 Conclusion

The life span of MOSFET dosimeters is determined by a saturation of the trapped positive charged in SiO$_2$ on a border traps which limits their wide use in clinical dosimetry because they should be disposed of after a certain dose. The possibility of recovering MOSkin detectors designed for real time skin dosimetry in radiation therapy has been investigated.

Two types of annealing, UV and isothermal at $T=150^0C$, were investigated on MOSkin detectors by multiple cycles of radiation and annealing. UV annealing allowed for recovery of the $V_{th}$ but a detailed investigation revealed this to be a
pseudo recovery associated with creation of electron traps in SiO₂ and interface which is a compensating radiation induced accumulated positive charge in the gate of the oxide.

After UV annealing the MOSkin detector demonstrated a sub-linear dose response usually observed for MOSFET with an accumulated dose. Additionally, the UV effectively increased the concentration of traps on the Si-SiO₂ interface, leading to a ±26mV instability of the MOSkin response after the 3rd annealing, and a total accumulated dose of 45Gy, compared to ±2-3 mV after a total dose 45Gy without UV annealing; effectively it reduced the low dose limit to 10cGy. All these facts suggested that UV annealing did not recover the MOSkin detectors and is not a suitable option for clinical applications.

In the case of isothermal annealing at 150°C for 100h, we demonstrated the possibility of complete recovering the MOSkin detectors. In this case the MOSkin detector showed constant sensitivity after recovery of the V_th, and an acceptable level of instability (±4mV) for clinical reuse in contrast to after UV annealing. It is a much longer annealing process than pseudo UV annealing, but it has a much better dosimetric outcome. Higher temperature annealing may eventually reduce the annealing time, but this will require further investigation because the soldering material and encapsulation material of the detector may ultimately limit this option.

To conclude, thermal annealing of MOSkins appears to be the most promising technique for clinical reuse of MOSkins detectors. The sensitivity was effectively recovered to within 99% of its original value so the MOSkin’s dose calibration factor was maintained. Such an annealing procedure may find a practical clinical use in
radiotherapy centres in developing countries where financial resources are more limited than in developed countries.
CHAPTER 4

4. DIRECT CURRENT ANNEALING OF MOSFET DOSIMETER

4.1 Introduction

As mentioned in the previous chapter (Chapter 3), annealing of the MOSFET dosimeters can potentially be used to achieve MOSFET dosimetry with an unlimited linear dynamic range and ability which is currently not possible. MOSFET dosimeters developed at the CMRP have a linear range for doses of more than 50-60 Gy that can be used for approximately 25-30 fractions of dose delivery on medical LINACs, for most radiotherapy modalities. After these accumulated doses the MOSFET response is slightly sub-linear because of an accumulated positive charge of a border traps in SiO$_2$ and on the interface due to creation of traps with concentrated $N_{it}$ in Si on an Si-SiO$_2$ interface, that has a negative charge in the case of p-MOSFET [55, 58].

The possibility of recovering MOS structures by thermally annealing the positive charges in SiO$_2$ and on the interface has been well studied [71, 72, 74, 75], and has been demonstrated when applied to MOSFET dosimeters [74] using the isochronous and isothermal method of annealing. It was shown that p-MOSFET dosimeters can be fully annealed under $T = 150^\circ C$, and zero bias conditions for 100h after doses of 50Gy and 400Gy.

However, thermal annealing reduces its sensitivity by approximately 20% during the second round of irradiation under the same electric field in oxide 0.125 MV/cm (it is similar to the operating conditions of the CMRP MOSFET detector $+5V$ and
$t_{ox}=0.6\mu m$, i.e. electric field 0.1MV/cm). The reasons for this is based on existing investigations associated with an introduction of neutral electron traps in the gate oxide which cannot be annealed under 150 $^0$C or with additional border traps (or their restructuring in the oxide) further away from the interface, and which are not as effective at changing the hole concentrations in the channel [58]. The positive and negative border traps and interface traps of an energetic MOS structure are shown in Figure 2-2 on Chapter 2.

Thermal annealing is quite a good option for recovering of the MOSFET dosimeters but it is not suitable for a busy clinical practice due to requiring 100h annealing cycles. Reducing the sensitivity with consecutive irradiation is also less desirable, but it is not essential because the calibration of a MOSFET dosimeter after each annealing run is always required in clinical practice.

Another approach to the recovery of irradiated MOS structures which has not been investigated as much is current annealing. This type of annealing was initially proposed to improve the radiation hardness of the Complementary Metal Oxide Semiconductor (CMOS) electronics during irradiation [59], and was investigated in CMOS electronics in case of alternating current annealing with frequency 50-200 Hz [76].

Experiments were carried out on CMOS integral circuits (IC) irradiated with Co-60 gamma photons and 4MeV electrons up to doses of 5x10$^6$ Rad. Current injection was fed through power terminals that switched the parasitic n-p-n-p transistors in a conducting mode to provide essential current through the source – substrate(sub), drain-substrate, and protective diode of multiple transistors, followed by a fast heating of the total integral circuit up to a temperature $T=200-240\ C^0$ additionally to
electron-hole (e-h) plasma injection. Electron plasma injection produces a strong Coulomb field close to the traps which reduces their energetic barriers and releases the captured charge in border traps easier under a lower temperature and similarly reduced charge on the interface traps.

Additional enhancement of the thermal heat effect was observed for alternating current because the motion of electron plasma to the lattice dissipates the energy more effectively and leads to “shacking” and the ordering of broken bonds of a dislocated ion nature, i.e., it minimises the lattice’ potential energy and annealing of the radiation defects [123]. It was demonstrated that alternating current annealing in CMOS IC irradiated with Co-60 doses up to $10^5$ Gy under a current 350-700mA during 30-60 s can provide 90% recovery in an individual p-MOSFET switchers and less than 60-70% in n-MOSFET switchers [76].

Another synergetic effect related to current annealing is the hot electron effect. In a strong electric field an electron (called a hot electron) can gain a kinetic energy that is far above the equilibrium thermal energy of electrons and lattice. [51]. The effect produced by hot electrons and holes in a channel of the MOSFET is described in [79-81], and presented in a Figure 4-1.
Figure 4-1: Injection of hot electrons through the Si-SiO\(_2\) interface (adapted from ref.[51]).

Hot electrons injected through the Si-SiO\(_2\) barrier to the gate oxide that are recombined with trapped holes or trapped by electron traps, are compensating positive charge on the boarder traps [79].

This effect is realised as the drain current in n-MOSFET decreases and interface traps are created, and because the electrons form the majority of charge carriers in the n-channel, they can break the Si-Hydrogen bonds on the interface to create the interface traps [81, 124], while in p-MOSFET an increase in the drain current and a decrease of \(V_{th}\) were observed. [79, 125, 126]. In the case of p-MOSFET when the current in an inverse channel is due to holes, the hot electrons are injected to the gate oxide from the substrate (sub) from the source –substrate or drain –substrate currents. In addition, the probability of hot carriers being injected reaches its maximum when \(V_g = V_d\) [51]. During the injection of hot carriers there are no interface traps generated on a p- MOSFET, because there are no hot electrons passing through the p-channel. This reason make the p-MOSFETs more suitable than n-MOSFETs under...
high stress conditions [79, 125, 127], which is also confirmed through the almost full recovery of p-MOSFETs in contrast to n-MOSFETs during the current annealing [76]. The effect of current annealing depends on the operating conditions of the MOSFET, whether annealing under irradiation or after irradiation, the bias on the gate during annealing, and other factors. All these effects associated with current annealing have contributed to the recovery of p and n type MOSFETs in CMOS IC [76]. The efficiency of current annealing is determined by the specific dissipated power in a MOSFET sensitive volume \( P = IV/A \), where \( I \) and \( V \) are the current and voltage drop on a p-n junction(s) and \( A \) is the cross sectional area through which the injected current is flowing. Based on this approach it is expected that the annealing current for a single MOSFET should be less than that for IC with multiple transistors. Multiple cycling (irradiation, then current annealing) in CMOS IC was reproducible and was shown to keep IC operational [76].

The application of current annealing for recovering radiation sensors was proposed by Rosenfeld et al. [78] for P-I-N diodes, and has been used for fast neutron dosimetry. Single pulsed currents with a duration of 1-2 seconds in a forward bias of a P-I-N diode, with a current density 0.1A/mm\(^2\) allowed for 60% recovery of the forward voltage drop of the diode due to a synergy between the heating effect and injection of e-h plasma. Thermal only annealing of Radiation detectors due to displacement clusters produced by fast neutrons after irradiation required a temperature of \( T=600 \, C^0 \) for 1h for the same recovery [128].
Previous studies for the current annealing of a single p-MOSFET dosimeter have been encouraging, but this method has yet to be performed for the application of a MOSFET in clinical use.

4.2 Materials and methods

As part of this study, the annealing of a CMRP p-channel MOSFET was investigated based on this phenomenon. In the initial experiments an attempt was made to pass a high direct electric current (DC) between the source and the drain to heat up the p-MOSFET dosimeter using the circuitry presented in Figure 4-2, but this current was limited by the transconductance of the p-MOSFET. To inject high specific power into a p-MOSFET, the two forward biased source-substrate and drain-substrate p-n junctions of the MOSFET can be used while the gate is left floating. The use of the two p-n junctions to deliver high specific power will be discussed in detail in Chapter 5, while DC annealing through a source-drain current will be investigated in this chapter.

Four p-MOSFETs, the MOSkin™, designed at the Centre for Medical Radiation Physics, University of Wollongong, were used in this experiment. Three were pre-irradiated to ~20Gy dose ($V_{th}$~14,600 mV- 11,600 mV) and one was un-irradiated $V_{th}$~7,000 mV. The MOSFETs were connected to an electronic circuit similar to the one that is widely used for MOSFET readout, as shown in Figure 4-2. In this circuit the gate and the drain are connected together ($V_g = V_d$), allowing the probability of injecting hot carriers to be maximised [51]. For the MOSFET readout, the current passing through the p-channel was set to 160µA. Current annealing was carried out with current in a range of 5mA to 15mA, which corresponded to voltage drops on the p-MOSFET in a range of 40V to 55V respectively. This was performed as each sample has a different channel resistivity due to different initial pre-irradiation and
the DC current required for heating may vary. The main criterion in this setup was to maintain a constant temperature on the MOSFET gate without melting the contacts soldered to the flexible carrier. To investigate the optimal annealing temperature, the temperature was varied from 60°C to 105°C and was monitored using a thermocouple sensor placed on top of the flexible polyamide carrier above the gate. This provided the possibility of adjusting the current to avoid damaging the contacts. In a process of annealing, of the positive border charge, the $I_{ds}$ current in the p-MOSFET was increased under a constant voltage drop across the p-MOSFET that will increase the dissipation of power and cause an uncontrollable increase in the temperature. In such a mode it will be difficult to understand the role of the current magnitude in annealing the p-MOSFET.

In this case a constant current source was used so that the temperature is controlled by adjusting the current to maintain an optimal temperature during the annealing process. Therefore, under a constant current, if the channel resistivity decreases due to annealing, the magnitude of the voltage drop on the p-MOSFET (voltage on the gate, $V_g$) will decrease.
The MOSkins were irradiated on a medical LINAC with total doses of 31.5, 31, and 30.5 Gy in three consecutive cycles, by placing the detectors under a full build-up condition in a solid water phantom at $d_{\text{max}}$. The field size (FS) was 10x10 cm$^2$ and the SSD was 100 cm. An initial radiation of the p-MOSFETs was performed to check their linearity and sensitivity, and after irradiation the MOSFETs were kept at room temperature for ~24 hours before annealing. The cycles of annealing and irradiation were repeated three times in three consecutive weeks, to avoid any long term annealing at room temperature.

The aim of this experiment was to irradiate the MOSFETs with an essential dose in one session that corresponds to multiple typical fractions, and to investigate the main experimental dosimetric parameters such as the reproducibility of annealing, linearity, and sensitivity.

The p-MOSFET sensitivity for each dose was calculated as:
\[ S_v = \frac{\Delta V_{th}}{Dose} \text{ (mV/cGy)} \]  

Equation 4.1

and the average sensitivity as:

\[ S_{v,av} = \frac{\Delta V_{th,Total}}{Dose_{Total}} \]  

Equation 4.2

The average sensitivity is equal to the slope of the accumulated threshold voltage \((\Delta V_{th})\) with an accumulated dose, and in case of a linear response, is equal to the initial sensitivity of the MOSFET. It can be used to indicate whether to re-calibrate the MOSkin with an accumulated dose or to continue dosimetry.

4.3 Results

4.3.1 DC annealing

4.3.1.1 First cycle of annealing and irradiation

In this experiment the four MOSkins (sample.1: virgin), and three pre-irradiated MOSkins (sample.2, sample.3, and sample.4) were used. Firstly, a pre-irradiated test sample (sample.4) was annealed by heating the MOSkin to 60°C with current \((I = 5.5 \text{ mA})\) with thermocouple attached for one minute, but there was no visible effect on its \(V_{th}\) change. This was repeated at 70°C and 75°C \((I = 8 – 9 \text{ mA})\), and no change in the \(V_{th}\) was observed within 1 min of current annealing. However, when the
temperature reached 80°C - 82°C (I = 11 – 12.4 mA), there was a decrease in $V_{th}$ after 1 min of annealing. This experiment showed that the annealing is correlated to a current magnitude, where increasing the current leads to faster annealing. When the temperature reached approximately 105°C (I = 15.2 mA), the annealing was very fast but the soldering material for this sample melted, so the annealing temperatures were kept between 90°C - 102°C, for the rest of the samples. This is equivalent to a current “I” ranging from 11 to 15.7 mA, which was easy to control by current through the p-MOSFET. The annealing parameters: temperature (T) and time (t), in addition to the irradiation dose, and MOSFET $V_{th}$ during the initial (1st) annealing are presented in Table 4-1.

Table 4-1. The annealing parameters during initial (first) annealing, temperature (T) and time (t), the irradiation dose, initial voltage ($V_{init}$), post-irradiation voltage ($V_{irr}$), post-annealing voltage ($V_{ann}$), for three MOSkin samples.

<table>
<thead>
<tr>
<th>Dose(Gy)</th>
<th>1st annealing parameters</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Sample.1</td>
</tr>
<tr>
<td>Dose(Gy)</td>
<td>0 (Virgin)</td>
</tr>
<tr>
<td>$V_{init}$</td>
<td>12.650</td>
</tr>
<tr>
<td>$V_{irr}$</td>
<td>12.650</td>
</tr>
<tr>
<td>$V_{ann}$</td>
<td>2.830</td>
</tr>
<tr>
<td>$dV_{irr} - V_{init}$</td>
<td>0</td>
</tr>
<tr>
<td>$dV_{irr} - V_{ann}$</td>
<td>9.820</td>
</tr>
<tr>
<td>T( C)</td>
<td>90</td>
</tr>
<tr>
<td>Annealing time (min)</td>
<td>30</td>
</tr>
</tbody>
</table>
I-V characteristics of the MOSkin sample 1, before and after applying the current annealing procedure, and when it was not irradiated and then after being irradiating with a dose of 31.5 Gy, are shown on Figure 4-3.

Figure 4-3: I-V characteristics for sample 1: Virgin, annealed without preliminary irradiation, irradiated, and post-irradiation-annealing.

It has been noticed that annealed MOSkins need at least 2.5 Gy of pre-irradiation before they became stable. All the following data and graphs are after pre-irradiating the MOSkins to 2.5 Gy.

The linearity, and sensitivity of the MOSkins for doses up to 31.5 Gy, can be seen on Figure 4-4, and Figure 4-5 for the three samples. All the samples have shown very good linearity with an accumulated dose within ±2%, and with slightly different average sensitivities, 2.53 ± 0.06 mV/cGy, 2.43 ± 0.04 mV/cGy, and 2.50 ± 0.05 mV/cGy for Samples 1, 2, and 3 respectively. The instability in Vth measurements in
each point was within ±1mV, so the error bars cannot be seen in the graph. In addition, the reproducibility of the change in Vth during consecutive measurements was stable to within ±1 mV at all the dose points. The automatic reader was accurate to ±1 mV.

Figure 4-4: The response of the three MOSkins with a dose accumulated during irradiation up to 31.5 Gy on the 1st irradiation. The error bars are too small and cannot be seen in the graph (±1mV).
Figure 4-5: Variation in sensitivity with an accumulated dose for the three samples up to 31.5Gy on the 1\textsuperscript{st} irradiation.

4.3.1.2 Second annealing-irradiation

The parameters that were used for the second round of annealing are shown in Table 4-2. The MOSkin detectors were kept at room temperature for 24 hr to let the signal stabilise from any short term fading after current annealing. The I-V characteristics of all the MOSkin detectors were without any changes in comparison to the original shape of the I-V curves. It is important to note that the time of annealing was different for all 3 MOSkin detectors.
Table 4-2 shows the annealing parameters during the second cycle, temperature (T) and time (t), the irradiation dose, initial voltage ($V_{init}$), post-irradiation voltage ($V_{irr}$), post-annealing voltage ($V_{ann}$), for three MOSkin samples.

<table>
<thead>
<tr>
<th></th>
<th>Sample.1</th>
<th>Sample.2</th>
<th>Sample.3</th>
</tr>
</thead>
<tbody>
<tr>
<td>$V_{init}$</td>
<td>2.830</td>
<td>4.020</td>
<td>2.618</td>
</tr>
<tr>
<td>$V_{irr}$</td>
<td>11.374</td>
<td>11.997</td>
<td>10.997</td>
</tr>
<tr>
<td>$V_{ann}$</td>
<td>3.160</td>
<td>2.720</td>
<td>2.450</td>
</tr>
<tr>
<td>$dV( V_{irr} - V_{init} )$</td>
<td>8.544</td>
<td>7.977</td>
<td>8.379</td>
</tr>
<tr>
<td>$dV( V_{irr} - V_{ann} )$</td>
<td>8.214</td>
<td>9.277</td>
<td>8.547</td>
</tr>
<tr>
<td>Dose(Gy)</td>
<td>31.5</td>
<td>31.5</td>
<td>31.5</td>
</tr>
<tr>
<td>$T(\degree C)$</td>
<td>93</td>
<td>94</td>
<td>102</td>
</tr>
<tr>
<td>Annealing time (min)</td>
<td>45</td>
<td>7</td>
<td>3</td>
</tr>
</tbody>
</table>

The annealed MOSkins on the second irradiation show excellent linearity within ±1% and their average sensitivities are 2.5±0.05 mV/cGy, 2.5±0.04 mV/cGy, 2.52±0.04 mV/cGy for Samples 1, 2, and 3, respectively, as shown in Figures 4-6 and 4-7.
Figure 4-6: Variation in sensitivity for the three samples on the 2\textsuperscript{nd} irradiation.

![Graph showing sensitivity variation for three samples]

Figure 4-7: Response of the three MOSkin detectors on the 2\textsuperscript{nd} irradiation. The error bars are too small and cannot be seen in the graph (±1mV).

![Graph showing response of three MOSkin detectors]
4.3.1.3 Third annealing-irradiation

After receiving 31 Gy in the previous cycle, the MOSkins were annealed according to the parameters shown in Table 4-3. The threshold voltage after annealing ($V_{\text{ann}}$) can be seen on the same table, which also shows that $V_{\text{ann}}$ for sample 2 was very low, so an illustration of its I-V characteristics is shown in Figure 4-8.

Table 4-3 The annealing parameters during the third cycle, temperature (T) and time (t), the irradiation dose, initial voltage ($V_{\text{init}}$), post-irradiation voltage ($V_{\text{irr}}$), post-annealing voltage ($V_{\text{ann}}$), for three MOSkin samples.

<table>
<thead>
<tr>
<th>3rd annealing parameters</th>
<th>Sample.1</th>
<th>Sample.2</th>
<th>Sample.3</th>
</tr>
</thead>
<tbody>
<tr>
<td>$V_{\text{init}}$</td>
<td>3.159</td>
<td>2.504</td>
<td>2.286</td>
</tr>
<tr>
<td>$V_{\text{irr}}$</td>
<td>10.631</td>
<td>9.962</td>
<td>9.705</td>
</tr>
<tr>
<td>$V_{\text{ann}}$</td>
<td>2.250</td>
<td>1.290</td>
<td>2.088</td>
</tr>
<tr>
<td>dV($V_{\text{irr}}$ - $V_{\text{init}}$)</td>
<td>7.472</td>
<td>7.458</td>
<td>7.419</td>
</tr>
<tr>
<td>dV($V_{\text{irr}}$ - $V_{\text{ann}}$)</td>
<td>8.381</td>
<td>8.672</td>
<td>7.617</td>
</tr>
<tr>
<td>Dose(Gy)</td>
<td>31</td>
<td>31</td>
<td>31</td>
</tr>
<tr>
<td>T(°C)</td>
<td>96</td>
<td>96</td>
<td>96</td>
</tr>
<tr>
<td>Annealing time (min)</td>
<td>15</td>
<td>10</td>
<td>6</td>
</tr>
</tbody>
</table>
Figure 4-8: I-V characteristics for typical p-MOSFET and in comparison with annealed sample 2 after 3rd annealing.

The response of all the MOSkins on the third irradiation was linear, and their sensitivities were very close, except for sample 2 with the non-typical I-V characteristics, as shown in Figures 4-9 and 4-10. The average sensitivities were 2.42±0.13 mV/cGy, 2.22±0.1 mV/cGy, and 2.53±0.1 mV/cGy for sample 1, 2, and 3 respectively, as shown in Figure 4-7 and Figure 4-8.
Figure 4-9: Variation in sensitivity with dose on the 3\textsuperscript{rd} irradiation.

After the third irradiation two out of three samples were damaged due to extensive connection and disconnection during these experiments, which rendered a fourth annealing impossible.
Figure 4-10: Response of the three MOSkin detectors on the 3\textsuperscript{rd} irradiation.

4.4 Discussion:

4.4.1 DC annealing

Direct current (DC) proved to be efficient for recovering the p-MOSFET $V_{th}$, and the optimal annealing temperature was found to be 100ºC. A higher temperature may be better for annealing, but it risks melting the soldering materials, which limits this option. The solder melts at 183ºC, so the 105ºC ambient melting point measured by the thermocouple may represent approximately 183ºC at the contacts. The thermal conductivity between the gate of the MOSkin and the thermocouple do not allow for accurate measurements of temperature for many reasons, such as the size of the MOSkin gate, the contact point, and the surrounding air temperature. It also indicates that the internal annealing temperature is higher than the temperature that can be measured. The main reasons for this annealing are mainly from the dissipation of
localised thermal energy, the injection of hot carriers from the drain, and the effect of plasma [78, 79].

4.4.1.1 First annealing-irradiation
The first annealing showed that the MOSkins recovered to voltages that were less than their initial values. In addition, we demonstrated that un-irradiated MOSkin (sample 1), can be annealed before irradiating to lower $V_{th}$ values than after the process of manufacturing. This means that pre-annealing can be performed on un-irradiated MOSkins to set a common threshold voltage for the whole batch and to extend their useful range. It can be seen from Table 4-1 that a slight increase in the current or the annealing temperature from 90ºC to 96ºC, as measured by the thermocouple, leads to a rapid decrease in the annealing time (from 30 min to 5 min). This can be explained by the electric power law where the power is proportional to $I^2$. The annealed MOSkin proved linear with excellent stability ± 1mV under irradiation. The average sensitivities were very close, 2.53±0.06 mV/cGy, 2.43±0.04 mV/cGy, and 2.50±0.05 indicating that the DC method has effectively annealed both the interface and SiO$_2$ border traps.

4.4.1.2 Second annealing-irradiation:
As with the 1st cycle the MOSkins sensitivities, linearity, and $V_{th}$ were recovered. The total recovery of sensitivity and threshold voltage indicates that all SiO$_2$ and interface traps were annealed. The excellent stability of the threshold voltage (± 1mV) suggests that the radiation created interface traps had been totally dissipated [59], and the response of all three MOSkins were almost identical after 1st cycle of irradiation and annealing. This can be explained because some unstable defects associated with mechanical stress produced during fabrication had become annealed.
4.4.1.3 Third annealing-irradiation:
The inconsistent behaviour of sample 2, as seen from the I-V characteristics in Figure 4-8, indicate that repetitive cycles affect sensitivity, as shown in Figures 4-9 and 4-10. This is could be a result of the MOSkin degrading from electrical stress which affects the charge carrier concentration and adjusts the length of the channel [51].

4.5 Conclusion
The annealing of MOSFET detectors, i.e. specifically the CMRP MOSkin, has proved to be possible using the direct current method. This method provided full recovery of the sensitivity and threshold voltage. In addition, the instability induced from the radiation that created interface traps was totally eliminated. Furthermore, annealing takes effect for temperatures of 80ºC and above, and within one minute.
The temperature presented is not necessary equal to the temperature at the gate of the MOSkin; however DC annealing demonstrated the ability of p-MOSFET to recover under lower temperatures and in a much shorter time than isothermal annealing [4]. An increase in the current/or temperature leads to a rapid increase in power and faster annealing. This should be taken into account to prevent excess heat damaging the MOSkin when the proposed recovery method is used in clinical practice.
The $V_{th}$ adjustment of un-irradiated MOSkins is possible and can be used to extend the linear range or set the identical initial $V_{th}$ for the whole batch of MOSkin.
Annealing the MOSkin to voltages less than 2 V affected its I-V characteristics and response. Pre-irradiation of the MOSkins to an X-ray dose of (~2.5Gy) followed each annealing cycle improved its response and linearity. The mechanism behind this annealing is a combination of factors such as, plasma annealing, localised heating, and the injection of hot electrons to the SiO₂.
5. PULSED CURRENT ANNEALING OF P-MOSFET BASED DOSIMETER

5.1 Introduction

DC current annealing and the physics behind it have been covered in the previous chapter. However, there are some problems and limitations with DC current annealing regarding the linearity, time of annealing, reproducibility of sensitivity, and possible deterioration of electrical contact between MOSkin and the flexible carrier. In this chapter, the use of pulsed high current (50mA~240mA) annealing to recover irradiated MOSkin dosimeters for clinical use will be investigated. The idea behind pulsed annealing is localised fast heating of the gate area of the MOSkin without essentially heating the contact pads. This will cover the current magnitude, pulse duration, number of pulses, and the annealing time. This will also cover the dosimetric characteristics, linearity, sensitivity, recovery of $V_{th}$, and the recovery of sensitivity of the p-MOSFET before and after annealing, in multiple consecutive cycles of irradiation and annealing.

5.2 Materials and Methods

A CMRP p-MOSFET called “the MOSkin™” was used in this study. As shown in the previous chapters, this type of MOSFET is reusable; it can be annealed by UV, and thermal or Direct Current (DC). Two MOSkin detectors were used in this study. Some of them were pre-irradiated and had reached their recommended lifespan so they were used to investigate the optimum annealing parameters. The remainder were four virgin MOSkin detectors. The virgin ones were irradiated to a dose of 80 Gy at $d_{max}$, on a 6 MV X-ray photon beam from the LINAC 23EX at ICCC,
Wollongong. The irradiation was carried out in five consecutive weeks. The detectors were kept at room temperature after each irradiation for 24 hrs then annealed by pulsed current.

The annealing was carried out using an adjustable pulsed current source up to 300mA. High current pulsed annealing was performed with different regiments as follows: with water drop above the MOSFET chip, or totally immersed in water, while some were in air. The electronic circuit used in these experiments is shown in Figure 5-1, where the two P-N junctions, source-substrate, and drain-substrate, operated with forward bias to allow high current to pass through the two junctions. Such a bias leads to an affective injection of the hot carriers [129-131] and to heat the sensitive area (SiO$_2$) locally. The width of the pulse (duration) and current magnitude were maintained using a pulse generator as the power supply for the current source, and was monitored by oscilloscope, see Figure 5-2. Many tests were
carried out with pre-irradiated MOSkin detectors (up to its recommended lifespan) to identify the optimum pulsed current annealing parameters without burning the MOSkins. These parameters include the current magnitude pulse duration and the time between the consecutive current pulses. The current magnitudes used were 240 mA, 200 mA, 150 mA, 100 mA, and 50 mA. The duration of each pulse was determined experimentally to avoid melting the soldering materials while the number of current pulses was determined so that each sample could achieve full recovery of the threshold voltage $V_{th}$.

Figure 5-2: Pulsed current annealing setup.

5.3 Results and Discussion

Preliminary tests:

5.3.1 Annealing with 240mA current

The first test with pre-irradiated MOSkin was with 240mA current and a pulse duration of 2 seconds. It was observed under the microscope that the MOSkin
soldering pads melted very quickly, within less than 2 seconds, because of the high power delivered locally to a small volume (~1.7 Watt). Thus, to avoid melting the detector pad a cooling medium for a heat sink from the MOSFET was investigated.

It was observed during the experiment on several MOSkin samples that 12 - 14 seconds was enough to evaporate the water drop placed above the flexible carrier in a MOSFET, so 12 seconds became the duration of a single pulse. A virgin MOSkin (Sample.1) was irradiated to the maximum dose that can be read by the reader, in this case 82 Gy, or 28,000 mV limits in reading the $V_{th}$, and annealed with 240mA, for 12 seconds in presence of a drop of water above the surface of the MOSkin. This MOSkin recovered 82% of its initial $V_{th}$, and the average sensitivity was checked by irradiating the MOSkin to 80Gy again. As shown in Figure 3, the dose response curves for the virgin and annealed sample almost overlapped and have the same slope or average sensitivity (~2.4 mV/cGy) up to 66Gy on the second irradiation, as shown in Figure 5-3. That part of the curve after 66Gy was not included in the graph because the $V_{th}$ for the used regiment of annealing had not fully recovered and the reader had reached the maximum voltage limit after 66Gy irradiation. The I-V characteristic curves for Sample1 are shown in Figure 5-4.
Figure 5-3: MOSkin dose response for sample1: virgin MOSkin- blue diamond; annealed with 240mA for 12 s with water drop above the gate-red square.
Figure 5-4. I-V characteristic curves for sample1, including a virgin MOSkin irradiated to a dose of 82Gy, and then annealed and irradiated again to a dose of 82Gy.

While the second annealing process takes place with a water drop, the soldered contact still melted. It seems that controlling the amount of water is not easy so a suggestion was made to totally immerse the MOSkin under water at a temperature of about 18°C to prevent it from melting.

5.3.2 1st Cycle: Annealing with 200mA and 150mA, MOSkin immersed in water.

Three virgin MOSkin detectors were used in this experiment (sample.2, sample.3, sample.4), all of them were irradiated to 80 Gy and annealed as follows: sample.2,
three pulses of 200 mA, each pulse of 20s duration, sample.3 and sample.4, three pulses of 150 mA, each pulse of 20s duration, all of them were immersed in water. They were then irradiated again to about 80 Gy. It was noted for samples 2, 3, and 4 respectively that their threshold voltages ($V_{th}$) had recovered to 94%, 93.6%, and 90.8% from their initial $V_{th}$ and their average sensitivities had recovered to 80%, 89%, and 84% respectively, compared to their initial average sensitivities. The samples on this cycle behaved in almost the same way in terms of linearity and sensitivity. To illustrate this, sample.2 and sample.3 are presented in Figure 5-5.

![Figure 5-5: The MOSkin dose response for virgin and annealed samples 2 and 3 with pulsed currents of 200mA for 20s, and 150mA for 20s, respectively, and all were immersed in water: Sample 2 Virgin (black diamond) and annealed (light blue); Sample 3 Virgin (red square) and annealed (star violet),](image-url)
5.3.3 2nd Cycle: Annealing with 200mA and 150mA, MOSkin in air.

These three samples were annealed again with different regimes. Sample 2 was annealed with 19 square pulses of 200mA, and the duration of each current pulse was 1s and pitch 4s, to give time for cooling in air. Sample 3 was annealed with 20 square current pulses of 150mA, the duration was 3s and the pitch was 4s. Sample 4, with 22 current pulses of 150mA, the duration was 3s and the pitch was 4s. The dose response for the virgin Sample 4, after the first pulsed current annealing immersed in water, and after the second pulsed annealing in air, is presented in Figure 5-6.

Figure 5-6: Dose response for the virgin Sample 4, annealed while immersed in water and pulsed current annealing in air.
Figure 5-6 shows, as an example for sample 4, that the response – dose curve is slightly better than the previous annealing cycle (with water). The recoveries of $V_{th}$ for these samples compared to their initial $V_{th}$ were 79%, 83%, and 88% for sample 1, sample 2, and sample 3, respectively. The recovery of $V_{th}$ in this annealing cycle was slightly better, ~2%-10% for all samples, than when previously annealed in water. In addition, the recovery of the average sensitivities for samples 2, 3, and 4 compared to their initial average sensitivities were 82%, 92.3%, and 87%, respectively. This represents an increase in their sensitivity compared to the previous cycle (in water) by 3%.

5.3.4 3rd Cycle: Annealing with 200mA + 50mA, MOSkin in air.

In the third cycle, the annealing procedure was changed to investigate the possibility of recovering the sensitivity better. It was observed from previous results that annealing with a higher 200mA current in air, recovered the baseline $V_{th}$ very quickly, i.e., the majority of annealed traps located near the Si-SiO$_2$ interface, which are mainly interface traps and SiO$_2$ border traps [58, 132], whereas at a low 150mA current it takes longer to recover the $V_{th}$. On the other hand, a long annealing time produces a lot of thermal energy which can recover the sensitivities (anneal bulk oxide traps) much better than a short annealing time, as happened with sample1. For this reason a combination of them was used in the next cycles; high current annealing (200mA) in the beginning for a short duration (2s - 22s), followed by low current annealing (100mA-50mA) for a longer time, ~60s.

Sample.2 and Sample.4 were annealed with 10 current pulses at an amplitude of 200mA, a 1s pulse duration and 3 s pitch, to recover $V_{th}$, followed by 50 pulses at a
current amplitude of 50mA, a pulse duration of 1s and pitch of 3s to recover the MOSFET sensitivities.

The threshold voltage $V_{th}$ in this cycle was recovered by 188% for sample 2, and 168% for sample 4, compared to their initial $V_{th}$ when they were virgin. It is clear in this cycle that high current annealing can recover the $V_{th}$ very quickly, not only to its original value, but also to a value lesser by 88% and 68% of their original values respectively. This will extend the useful dynamic dose range of the detectors. The average sensitivities were checked by irradiating them to 80Gy again, where they were 95.4% and 97% of their original average sensitivities respectively. The dose response linearity compared to the previous annealing cycles is shown for sample 2 and sample 4, in Figure 5-7 and Figure 5-8.
Figure 5-7: Dose response relationship for sample 2, showing a comparison of the annealing cycles.
Figure 5-8: Dose response relationship for sample.4, showing a comparison of the annealing cycles.

5.3.5 4th Cycle: Annealing with 200mA + 100mA, MOSkin on air.

Furthermore, sample.2 and sample.4 were annealed again, with one pulse of 200mA as done previously, followed by 50 pulses at a current amplitude of 100mA, a 1s pulse duration and 4s pitch. In this annealing cycle (4th cycle) sample 2 was annealed with one 1s duration pulse and recovered 31% below its pre-irradiated $V_{\text{th}}$ while sample 4 required 12 pulses of 1s each, to recover 34% below its pre-irradiated $V_{\text{th}}$. Their average sensitivities were checked again by irradiating them to 80Gy, sample 2 recovered 98.2% and sample.4 98% of their initial average sensitivities, respectively. Their dose response curves are shown in Figure 5-9 and Figure 5-10, and are very close to each other.
Figure 5-9: Dose response relationship for sample.4, showing a comparison of the annealing cycles.
Figure 5-10: Dose response relationship for sample 2, showing a comparison of the annealing cycles.

As the above figures show, those MOSkin detectors have received a total 400Gy each and are still demonstrating 98.2% and 98% of their original average sensitivities, while their threshold voltages were less than their original values, i.e. the proposed pulsed current annealing regime enables the MOSkin detector to recover its sensitivity and extend its dynamic range. Compared to the previous annealing, it could be concluded that the pulsed current annealing regime of 200mA+100mA is slightly better than a 200mA+50mA regime because recovery was almost full, as shown in Figures 5-9 and 5-10. A summary of the average sensitivities and threshold voltages ($V_{th}$) recoveries for sample.2 and sample.4 during all cycles can be seen in Figure 5-11 and Figure 5-12. The transconductance of the annealed MOSkin
detectors after the 5\textsuperscript{th} cycle were checked for sample 2, and sample 4, they were about $\sim99\%$ of the virgin ones.

Figure 5-11. Recovery of sensitivity by sample.2 and sample.4 during the four cycles of pulsed current annealing with the regimes described above.
Figure 5-12: Vth recovery of sample.2 and sample.4 during the four cycles of pulsed current annealing with the regimes described above

5.4 Conclusion:

In conclusion, MOSkin seems to be promising in term of reusability for a longer lifetime. This is a potential addition to radiation dosimetry for clinical and space applications. As investigated in this study with pulsed current, it is concluded that pulsed current annealing should be divided into two parts to recover the original threshold voltage $V_{th}$ and the sensitivity. The procedure should start with high current annealing for a short time (~1s-20s) to anneal the interface and border traps at Si-SiO$_2$, followed by low current annealing for a longer time (~50s) to anneal the SiO$_2$. It is also concluded that a high current with a short pulse totally eliminates the instability on the $V_{th}$ created by radiation, and recovers it to the original uncertainty.
of the reader ±1mV. It seems that the dose history does not affect the reusability of the MOSkin up to the range of 400Gy. In addition, it seems that the MOSkin became more reliable with consecutive annealing, and decreased the annealing time when the dose history increased. Higher doses may need further investigation, but this is out of the scope of our clinical investigations. Application and investigation in details of current annealing for recovery of clinically used MOSkin detector was done firstly and can’t be compared with other similar work.
CHAPTER 6

6. VERIFICATION OF THE ACUROS® XB DOSE CALCULATION ALGORITHM IN THE BUILD UP REGION AND AROUND DIFFERENT INTERFACES

6.1 Introduction

Calculating the dose around interfaces where electronic disequilibrium exists is challenging for most of the current commercial treatment planning systems (TPS). In addition, measuring the dose at interfaces is also a challenge for most current radiation detectors. Dosimetry may be desired around interfaces in the human body such as the build-up region, the skin, air-tissue, air-tissue-air (e.g. larynx, rectum), tissue-lung-tissue, tissue-metal-tissue (e.g. hip prosthesis), tissue-bone-tissue, and tissue-metal-bone-tissue. At any of these interfaces TPS may over or underestimate the dose [41, 87-90, 105, 133-135].

Monte Carlo (MC) calculations can estimate the dose at interfaces between inhomogeneities with very accurately, and they can also be used to benchmark the treatment planning algorithm as well as ionisation chamber comparisons. One disadvantage of MC calculations is the time they take to achieve a reasonable accuracy [92, 98], which is why using real time radiation detectors is a good adjunct to MC calculations to verify the TPS [12, 13, 18, 41, 42, 45, 50, 63, 68, 136-138].

However, as mentioned in a chapter 2, not all detectors are suitable for interface dosimetry where there is a high dose gradient because interface dosimetry requires dosimeters with a very small sensitive volume to avoid volumetric effects and
minimal perturbation of the radiation field. Solid state detectors are more likely to be used in this case due to their small size [50].

As previously mentioned in chapter 2, Acuros® XB is a new treatment planning algorithm (TPA) available from Varian Medical Systems as part of their dose calculation software, Eclipse™. It is based on the deterministic solution of the Linear Boltzmann Transport Equations (LBTEs) for radiation transport in matter [97, 101]. Evaluating Acuros® XB in the build-up region and interfaces is important for confidence in any prediction of superficial skin doses on air-tissue interfaces, and in the build-up region under strong electronic disequilibrium conditions. In addition, doses around interfaces of varying media, such as lung tissue and metal tissue are important for calculating the target dose. In this chapter the dose predicted by the Acuros® XB TPS will be verified using an Attix ionisation chamber as a gold standard, and a CMRP developed MOSFET detector for interface dosimetry (MOSkin™) will also be used.

6. 2 Materials and method

6.2.1 Verification of the Build-up region

The measurements were carried out at the British Columbia Cancer Agency, BCCA, in Vancouver Island Centre, Canada. Dosimetry in the build-up region, as predicted by Acuros® XB, was benchmarked against an Attix Plane-Parallel ion chamber, (RMI Gammex 449), the MOSkin™, and a current TPS algorithm known as an Analytical Anisotropic Algorithm (AAA). The MOSkin™ is a semi-conductor MOSFET detector especially designed for superficial and skin dosimetry by CMRP. The depth dose profile at the build-up region was calculated for both algorithms at
the central axis of the beam. The phantoms were 30x30x30 cm$^3$ and 50x50x30 cm$^3$
RMI solid water phantoms which were used to cover all the field sizes. The radiation
field sizes (FS) were 4 cm x 4 cm, 10 cm x 10 cm, and 40 cm x 40 cm with a $0^\circ$
Gantry angle, as well as 10 cm x 10 cm with a $45^\circ$ Gantry. The beam parameters
were 6 and 18 MV X-rays generated from a Varian CLINAC 21EX, and the distance
from source to surface (SSD) was 100 cm. The settings of the TPS software
Eclipse™ for Acuros® XB and AAA settings were:

- Acuros® XB version 11.0.02, AAA version 10.0.28.
- Virtual Solid water Phantom dimension 50x50x30 cm$^3$
- 1 mm Grid size (1 mm$^3$ Voxel) with heterogeneity correction
- 6 MV and 18 MV X-rays from a Varian CLINAC 21EX

The MOSFET clinical dosimetry system was provided by the Centre for Medical
Radiation Physics (CMRP) at the University of Wollongong. Because this MOSFET
dosimeter (the MOSkin™, Figure 6-1) is unique, it has many advantages over other
commercial MOSFETs such as a reliable 0.07 or 0.02 mm WED of dose
measurements, real time dosimetry, and a probe thickness of 0.35mm. It can be used
for surface, skin, & In-vivo dosimetry. The MOSkin used for build-up dosimetry had
a 20µm build up layer, as shown in Figure 3-1.

Figure 6-1: The CMRP MOSFET dosimeter - MOSkin™.
Only two MOSkins were used for dosimetry in the build-up region. Both of them were annealed after each experiment or after reaching $V_{th} = 16$ Volts, which represents approximately 23 Gy of accumulated dose. Annealed MOSkins have demonstrated a good linearity and reproducible sensitivity. The current annealing procedure was described in detail in the previous chapter. The total dose in all the build-up measurements was about 230 Gy, which would have required at least 10 - 12 MOSkins to maintain a fixed sensitivity and stable signal ($\pm 1$ mV), however, in our case only two MOSkins were used and annealed when necessary. In a similar manner to MOSFET dosimetry, an Attix Parallel Plate ionisation chamber (IC) was used for all measurements apart from the 4x4 cm$^2$ field size because the Attix IC is larger than the field size. In this case MOSkin was used to verify Acuros® XB and AAA. A polarity correction factor was also applied to the Attix chamber in all measurements, as was recommended [139], albeit the correction for the guard ring was very small (less than 1%), and indeed it was omitted because the guard ring was sufficiently large for the Attix chamber[140, 141].

Figure 6-2: Attix ionisation chamber on the solid water phantom.
The build up measured with the Attix IC and MOSkin™ from the surface to the \( d_{\text{max}} \), on the central axis of the beam for all the photon energies that were used (See Figure 6-2, same setup was used with MOSkin™ Figure 6-5). Thin plastic sheets 150 µm, 200 µm and 400 µm thick were used to verify the dose in the build-up region. All the measurements and calculated depth doses were normalised to the dose at \( d_{\text{max}} \). In addition, the percentage difference (\( \% \Delta \)) between the Attix reading and the measured/calculated doses from each detector was plotted. The percentage deviation was also included in the graph to clarify the magnitude of deviation. With the 4 cm x 4 cm FS the MOSkin™ reading was used as the gold standard.

### 6.2.2 Verification of Acuros® XB around high and low density materials

The Attix ionisation chamber, MOSkin™ and GAFCHROMIC® EBT2 film dosimeters were used, as well as Monte Carlo simulations using EGSnrc with the LINAC modelling package BEAMnrc and the dose calculation package DOSXYZnrc. Doses near the interfaces, including air gaps, steel inserts, and lung were measured with MOSkin™ and EBT2 film, and were then compared to the doses calculated with Monte Carlo, Acuros XB, and AAA TPAs on the central axis of the beam. The measurements were started from the upper direction toward the heterogeneities for all experiments with MOSkin™ and EBT2. They were then moved vertically using the solid water slabs and the thin micro plastic sheets. MOSkin™ and EBT2 film were placed with the front face towards the beam. For this reason the substrate of both detectors should affect the measurements at the proximal interface, while the distal interface should not be affected because the sensitive volume of the detector was located on the interface.
6.2.2.1 GAFCHROMIC® EBT2 film

The active layer of GAFCHROMIC® EBT2 film lies under approximately 80 µm of WED. The configuration of EBT2 film is shown on Figure 6-3. The combined depth of the upper layers of film was approximately 80 µm while the substrate of the film below the active layer was approximately 75 µm deep. The EBT2 film was calibrated according to the procedure described by Ferreira et al, 2009 [142]. This method gives a high resolution depth dose reading with less than 2% over all uncertainties. The irradiated films were scanned using an EPSON Expression 10000 XL PHOTO flatbed scanner.

Figure 6-3: The GAFCHROMIC® EBT2 film layers thicknesses
6.2.2.2 Monte Carlo simulation

The LINAC radiation source was modelled using BEAMnrc to simulate the radiation output from the Varian linear accelerator 21EX located at BCCA. This model was designed and tuned for 6 and 18 MV photon beams [102]. The materials and density were assigned from the virtual CT images created by Eclipse™. Voxel sizes of 1.0 x 1.0 x 1.0 mm$^3$ were used to calculate the doses.

6.2.2.3 Phantoms and LINAC set up

The phantoms used for the steel insert and air cavity had similar 30 cm x 30 cm x 30 cm dimensions, while the steel insert and air cavity were 2.35 cm x 2.35 cm x 30 cm, as shown in Figure 6-4, Figure 6-5, and Figure 6-6. These heterogeneities were placed in phantom to simulate real clinical cases, with metallic objects (e.g. hip prosthesis) and air gaps larynx [82, 85, 143]. To simulate a tumour inside the lung, two Gammex RMI455 lung slabs 5 cm x 30 cm x 30 cm were sandwiched between solid water slabs with a 1 cm x 30 cm x 30 cm tissue slab in between to simulate a tumour. The first phantom lung was covered by a 5 cm x 30 cm x 30 cm tissue slab as build up, as shown in Figure 6-7. An isocentric radiation technique was used for all cases. To obtain a high resolution reading inside the target (1 cm slab thick) with MOSkin, at least 10 reading points (depths) were measured using thin plastic sheet. For EBT2 dosimetry, the film was perpendicular to the beam in order to read so many points, as described by Ferreira et al 2009 [142]. The photon beams used for all the heterogeneous experiments were 6 and 18 MV, with 10 cm x 10 cm field sizes. The field size for the lung case was 3 cm x 3 cm. In every case with these heterogeneities the treatment planning algorithm was pushed to conditions of extreme calculations.
Figure 6-4: Steel insert setup under LINAC.

Figure 6-5: Top view of the MOSkin and its cable position in a phantom at the depth of the steel insert/Air cavity.
Figure 6-6: Phantom design for air and steel heterogeneities.

Figure 6-7: Phantom design for Lung with tumour.
6.3 Results and discussion

6.3.1 Verification of Build-up region

6.3.1.1 6 MV X-ray build up region

The measured and calculated doses within the build-up region for the field size of 10 cm x 10 cm and 6 MV photon beam can be seen in Figure 6-8. The graph shows that the MOSkin™ and Attix chamber doses were in excellent agreement for all depths, with less than 2% deviation in every region except the surface where there was a 5% deviation. For the TPA calculations, AAA overestimated the dose by 80% at the surface and Acuros® XB overestimated the dose by 18%. This represents a better agreement between Acuros® XB and the Attix chamber than the AAA.
Figure 6-8: Doses in the build up for 10 cm x 10 cm field size, 6 MV photon beam, and 0° Gantry angle. Attix data points not visible as they are overlayed by MOSkin data points.

With the 40 cm x 40 cm field size and 6 MV photon beam, the MOSkin and Attix chamber doses show excellent agreement, with less than 2% discrepancy at the surface and less than 1% for all points. With the TPA calculations, AAA overestimated the dose by 25% while Acuros® XB underestimated the dose by 17%. It can be seen in Figure 6-9 that both codes converged to the Attix chamber and MOSkin measurements at a depth of 2mm.
Figure 6-9: Doses in the build-up region for 40 cm x 40 cm field size, 6 MV photon beam, and 0° Gantry angle. Attix data points not visible as they are overlayed by MOSkin data points.

Due to the large diameter of the Attix IC compared to the 4 cm x 4 cm field size, the surface dose was measured with the MOSkin™ as the gold standard reference. AAA and Acuros® XB overestimated the surface dose by 90% and 18% respectively. Both methods of calculation changed from overestimating to underestimating with depth, and converged to MOSkin™ at a depth of 2mm. Acuros® XB is in a good
agreement with the MOSkin measurements for depths greater than 0.5 mm, as shown in Figure 6-10.

Figure 6-10: Build-up region for 4 cm x 4 cm FS, 6 MV X-ray, 0° Gantry angle.

An oblique beam with a 45° delivery angle was used to check the feasibility of the TPAs in this situation because angled beam incidences are common in clinical cases. The Attix IC and MOSkin were in excellent agreement, being within 4% at the surface. The AAA algorithm underestimated the dose at the surface by 100%,
whilst Acuros XB performed much better than the AAA algorithm, with only a 10% overestimation at the surface. Acuros XB was in good agreement with the MOSkin™ for depths greater than 1 mm, as shown in Figure 6-11.

Figure 6-11: Build-up region for 10 cm x 10 cm FS, 6 MV X-ray, and 45° Gantry angle. Attix data points not visible as they are overlayed by MOSkin data points.
The Acuros® XB generally performed better than the AAA algorithm for all field sizes in regions of electronic disequilibrium, such as at the surface of the phantom. A summary of the findings and discrepancies of the surface doses predicted by both TPS’s is presented in Figure 6-12. Using Acuros® XB will potentially improve the clinical outcomes by predicting the surface doses more accurately than the current commercial treatment algorithms. This is important in the case of breast, head, and neck IMRT treatments where inducing low skin doses is important in retaining acceptable skin reactions. [48, 138, 144, 145]

Figure 6-12: Variation of the surface dose with field size for both algorithms, AAA and Acuros XB, and for 6 MV X-ray beams.

6.3.1.2 18 MV X-ray build-up region

For 18 MV X-ray beams, 10 cm x 10 cm FS, and 0° gantry angle, the MOSkin™ and Attix chamber measurements agreed to within 1%, as shown in Figure 6-13. Acuros® XB was in good agreement with the Attix chamber measurements for depths greater
than 3 mm and within 18% for depths less than 3 mm, while the AAA algorithm overestimated the dose at the surface by 200% converging to the Attix chamber measurements for depths greater than 5 mm.

Figure 6-13: Build-up region for 10 cm x 10 cm FS, 18 MV X-ray and 0° Gantry angle.
For large field sizes of 40 cm x 40 cm, 18 MV, and 0º Gantry angle, the MOSkin™ and Attix chamber measurements were in excellent agreement (within 2%) as expected, whilst the AAA algorithm and Acuros® XB overestimated the surface dose by 60%. Acuros® XB underestimated the surface dose by 38%. Acuros® XB data converged to agree with the Attix data within 2mm of the surface. The AAA takes 4 mm to reach a similar agreement, as shown in Figure 6-14.

![Graph](image)

Figure 6-14: Build-up region for 40 cm x 40 cm FS, 18 MV, and 0º Gantry angle.
For small 4 cm x 4 cm size fields, the MOSkin™ was used to benchmark the TPAs in the build-up region. The surface dose was overestimated by 80% for both TPAs. Acuros® XB measurements converged to the MOSkin™ measurements faster than the AAA algorithm, as shown in Figure 6-15. Acuros® XB was much better at calculating the surface dose for 6 MV X-ray beams under the same conditions.

Figure 6-15: Build-up region for 4 cm x 4 cm FS, 18 MV X-ray, and 0° Gantry angle.
With an oblique angle of 45°, 10 cm x 10 cm field size and 18 MV photon beam, the MOSkin™ measurements were in excellent agreement with the Attix chamber measurements, to within 2%, as shown in Figure 6-16. Both the AAA and Acuros® XB algorithms overestimated the surface dose by 80% and 15% respectively. With every size field in the 18 MV X-ray fields, Acuros® XB predicted the doses in electronic disequilibrium conditions on the surface far better than the AAA algorithm, as shown in Figure 6-17.

Figure 6-16: Build-up region for 10 cm x 10 cm FS, 18 MV, and 45° Gantry angle.
6.3.2 Verification of Acuros® XB around high and low density materials

6.3.2.1 Air Gap heterogeneity

A 6 MV X-ray beam was used extensively for these experiments, although some also included an 18 MV beam to highlight the greater impact the extended electron range at 18 MV had on doses in the lung. The 18 MV photon beam was not reported for air cavity.

The MOSkin™ and EBT2 were in excellent agreement with MC (within ±2%) at the proximal and distal interfaces (tissue-air and air-tissue respectively), as shown in Figure 6-18. At the tissue-air interface the difference between MC and Acuros® XB was +5.2% and -2.3% between MC and AAA. At the air-tissue interface the
difference between MC and both Acuros® XB and AAA were -3.2% and +13.5% respectively.

Figure 6-18: %DD for air gap heterogeneity inside solid water phantom for 6 MV X-ray.
As shown in Figure 6-19, the MOSkin™ and EBT dosimetry results agree closely with the results of the MC simulation, whereas 2 mm before the first interface, Acuros® XB overestimated the dose on the proximal interface by 4% and underestimated the dose 2 mm beyond the distal interface by 9.5%. AAA calculations appear to apply little or no correction for dose perturbation from the air gap, although both algorithms matched the MC predicted doses before 2 mm from the proximal interface and after 4 mm beyond the distal interface. This is likely to be due to these regions being a sufficient distance from each interface for electron equilibrium to be re-established.
6.3.2.2 Steel insert heterogeneity

When a 6 MV X-ray beam with a steel insert was used on the phantom, a peak can be seen at the tissue-steel interface [85], as shown in Figures 6-20 and 6-21. This dose enhancement peak resulted from backscattered photons, backscattered secondary electrons coming from the tissue-metal interface, and backscattered secondary electrons coming from the metallic insert [83, 85]. This peak was measured with the MOSkin™ and EBT2 film, and calculated using MC, Acuros® XB, and AAA. The AAA algorithm underestimated the peak height by 22% from MC (122%dd), and there was also a 3 mm shift of the peak from its expected position toward the steel insert. The EBT2 film and MC calculations were within ±1% agreement for peak height, while the MOSkin™ predicted a 5% less dose than the MC when measuring the backscattered peak. This was expected because there is a 0.35 mm Si substrate underneath the sensitive area of MOSFET shifting the sensitive area away from the surface. Acuros® XB underestimated the peak height by 10% and this peak was shifted 2 mm shallower from the expected peak position. All algorithms converged within 3.5 mm from the proximal interface, within ±2% agreement with MC, MOSkin™, and EBT2 film. However, 3 mm beyond distal interface, only Acuros® XB converged with MC and the experimental measurements, while AAA overestimated the dose and shifted from MC by 5.7%, while the AAA and MC dose calculations started to converge at depths ~8 mm beyond the distal interface. In terms of accuracy with the steel insert, Acuros® XB agreed with MC and MOSkin™ within 2 mm-3 mm before and beyond the metallic insert, while AAA underestimated the dose by 20% and 5.7% before the proximal interface and beyond the distal interface respectively. Acuros® XB shows reduced calculation errors as a result of the metallic medium inside the patient, making of the estimated backscattered peak
dose more accurate than AAA, potentially preventing normal tissue radiation toxicity side effects (necrosis) from backscattered radiation, especially if there are bones close to metal [85], e.g. a pin holding a hip prosthesis glued into bone.

Figure 6-20: %DD for steel insert heterogeneity inside solid water phantom for 6 MV X-ray, 10 cm x 10 cm FS.
With the 18 MV X-ray beam, the backscattered peak was higher than the 6 MV photon beam by approximately 130 %DD. The Acuros® XB, MOSkin, and EBT2 film were within ±2% agreement with the MC simulations before the proximal and distal interfaces, as shown in Figures 6-22 and 6-23. The Acuros® XB predicted peak was shifted 1 mm back from the expected proximal position. The AAA algorithm underestimated the dose before the proximal interface by 30% and did not predict the backscattered peak at all. The AAA algorithm also overestimated the dose beyond the distal interface by approximately 6%, from a depth of 6 mm and beyond. The Acuros® XB with the 18 MV photon beam calculated much better than the AAA algorithm at all depths.
Figure 6-22: %DD for steel insert heterogeneity inside the solid water phantom for 18 MV X-ray, 10 cm x 10 cm FS.

Figure 6-23: %DD for steel insert heterogeneity inside the solid water phantom for 18 MV X-ray, 10 cm x 10 cm FS (Zoomed).
6.3.2.3 Lung slabs heterogeneities

As this case contains different heterogeneities with different thicknesses, it represents a calculation challenge for the TPA, especially to target a tumour inside the lung. With the 6 MV X-ray beam both calculation algorithms were in good agreement with MC and MOSkin\textsuperscript{TM}, being ±2% inside the target volume, before and after the lung slabs, as shown in Figures 6-24 and 6-25. The maximum overall uncertainty for EBT2 film was within ±1.7%, but this error is not presented on the graphs to prevent overlapping with other error bars and markers. Figure 6-24 shows the 6 MV photon beam case where, if the MC dose is taken as a comparison, the AAA slightly underestimated the dose in the lung while the Acuros\textsuperscript{®} XB slightly overestimated the dose.

Figure 6-26 shows 18 MV energy where the AAA gives a slightly higher dose in the lung than MC and the Acuros\textsuperscript{®} XB is also slightly higher, but it is closer to the MC calculated dose.
Figure 6-24: %DD for lung heterogeneities inside solid water phantom for 6 MV X-ray, 3 cm x 3 cm FS.
Figure 6-25: %DD for lung heterogeneities inside solid water phantom for 6 MV X-ray, 3 cm x 3 cm FS (Zoomed target).

With the 18 MV photon beam both TPAs estimated the target dose very well, as well as the dose at the proximal and distal interfaces. Acuros® XB, MOSkin™ and EBT2 film were all within ±2.5% agreement with MC, while AAA was within ±5% agreement for the target dose, and also before and after the first and second lung slabs, as shown in Figures 6-26, 6-27, 6-28, and 6-29. The AAA algorithm deals with lung heterogeneity much better than air and steel, which is in good agreement with studies conducted by Gagne and Zavgordni [90]. In terms of clinical use, the AAA algorithm is still probably achieving acceptable accuracy compared to the Acuros® XB dosimetry calculations for lungs, although this would be a decision best undertaken by a radiation oncologist in consultation with a physicist and would depend somewhat on each individual plan. For example small volume high dose SRS plans may possibly require the higher accuracy algorithm,
Figure 6-26: %DD for lung heterogeneities inside solid water phantom for 18 MV X-ray, 3 cm x 3 cm FS.
Figure 6-27: %DD for lung heterogeneities inside solid water phantom for 18 MV X-ray, 3 cm x 3 cm FS (Zoomed proximal interface).

Figure 6-28: %DD for lung heterogeneities inside solid water phantom for 18 MV X-ray, 3 cm x 3 cm FS (Zoomed target).
6.4 Conclusion

The results for build-up dosimetry showed an excellent agreement between the MOSkin™ and Attix IC for all cases within ±1.5%, except for the tangential field where the MOSkin™ demonstrated a 3% underestimation of the surface dose due to the different WED of measurement by the MOSkin™ and Attix IC. Acuros® XB showed an agreement (within 20%) in the surface dose against the MOSkin™ and Attix IC for 4 cm x 4 cm (MOSkin™ only), 10 cm x 10 cm and 40 cm x 40 cm, 6 MV X-ray fields, and slightly worse (30%) for 18 MV X-ray fields. The AAA algorithm agreement on the surface dose was much worse, being almost 100% different than the MOSkin™ and Attix IC. Both codes converged to MOSkin™ and Attix IC measurements with increasing depth, with faster convergence by Acuros® XB. Acuros® XB calculations may be improved more by better treatment of backscattering for larger fields and higher X-ray energies.
For interface dosimetry, Acuros® XB was within ±2.5% agreement with MC calculations for all interfaces such as air, steel, and lung, for the photon beams of 6 MV and 18 MV. With steel heterogeneity there was a slight shift from its expected position for the backscattered peak that increased in depth by ~2 mm. The AAA algorithm did not treat the air case very well in that it almost ignored the air gap. The Acuros® XB calculation appeared to be predicting the dose voids in the lung which are typical of narrow fields in low density media. As expected, as the lateral electron range for 18 MV photon beam is greater than the 6 MV photon beam, the decreasing dosages in the lung was greater at 18 MV than the 6 MV photon beams.

With the steel insert, the AAA algorithm does not calculate the backscattered peak at all, while it underestimated the dose at the proximal interface by 30% compared to MC, and ~6% beyond the distal interface. For the lung insert case, both the Acuros® XB and AAA algorithms calculated the dose with acceptable differences from MC, ±2.5% and ±5% respectively for both 6 MV and 18 MV photon beams.
7. Conclusions

Four methods for annealing for a MOSFET dosimeter, the CMRP MOSkin, were investigated in this thesis. They were Ultraviolet (UV) annealing, isothermal annealing, Direct current (DC) annealing, and pulsed current annealing. They have been investigated under medical linear accelerator beams. The dosimetric experimental characteristics of these MOSFET dosimeters, such as linearity, sensitivity, and signal stability, were investigated after each annealing procedure. The best annealing method with optimum annealing parameters has been specified and used to anneal MOSkin detectors for clinical purposes. One important clinical procedure is commissioning the treatment planning algorithm. Acuros® XB is a newly implemented treatment planning algorithm that has been benchmarked against MOSkin detectors, EBT2, and an Attix ionisation chamber. The optimum annealing method has been applied to MOSkin detectors and has been used in the build-up region measurements, while interface dosimetry inside the phantom was performed with 11 virgin MOSkin’s. Only two MOSkin detectors were used and annealed for the build-up region experiments.

7.1 Ultraviolet annealing of MOSFET dosimeter

UV annealing of MOSFET detectors can recover the threshold voltage $V_{th}$ but it does not recover the sensitivity. It seems that UV annealing induces the formation of some oppositely charged traps which compensate for the pre-existing positive charges. This makes the threshold voltage ($V_{th}$) appear to decrease, although the sensitivity is still decreasing with the annealing cycles, but with no recovery. The transconductance measurements show that UV light decreases transconductance,
which suggests that UV light introduces some interface traps which increase the
instability. The instability of the MOSFET signal increased to ±26mV compared to
±1mV of the virgin one. This increased the error in any clinical measurements up to
10 cGy, as the lowest detectable dose is 1cGy/2.6mV. Hence, this type of annealing
is not recommended for clinical use, but it could be used in a high dose environment
such as industrial or space dosimetry.

7.2 Thermal annealing of MOSFET dosimeter

Thermal annealing or Isothermal annealing at 150°C showed full recovery of the
threshold voltage ($V_{th}$) and full recovery of sensitivity, while the instability in this
case is an acceptable ±4mV. This could introduce 2 cGy of error for low dose
measurements. This small instability is a residual from the non-annealed interface
traps induced by radiation. However, to anneal this part of the interface traps may
require a longer annealing time or higher temperature. However, the time of
annealing compared to UV annealing is much longer ~ 100 hours, and results in
better dosimetric outcomes.

In conclusion, isothermal annealing recovered 99% of the sensitivity, which
maintained the calibration factor for MOSFET. However, a longer annealing time
may limit its use in busy clinical departments, and higher temperature annealing will
be limited in this case by the melting point of the soldering materials.

7.3 Direct current annealing of MOSFET dosimeter

Direct current annealing of the CMRP MOSFET, MOSkin™ was possible with full
recovery of the threshold voltage and full recovery of sensitivity. Moreover, the
radiation induced instability on the MOSFET signal had been totally eliminated, as
evidence the threshold voltage reproducibility was within ±1mV compared to ±4mV for irradiated MOSFET to 30Gy. The annealing process has a threshold temperature that started from 80°C and above, this occurred within one minute after the DC current passed though the MOSFET channel. However, the temperature measured on the gate did not need to be exact, but it did demonstrate the ability of annealing under low temperature and less power in a much shorter time compared to isothermal annealing. High current annealing leads to a rapid increase in temperature and faster annealing, but this option is limited by the melting point of the soldering materials. Thus, this point should be taken into account in clinical purposes to prevent any damage to the MOSFET that may arise from excessive heat. The annealing of un-irradiated MOSFET is possible, to extend its linear range and to set the initial threshold voltage of MOSFET batch to an identical value. Pre-irradiating MOSFET to 2.5Gy after each annealing improved its response and linearity. The physical mechanism behind this phenomenon is based on a synergy of the plasma effect, localised heating, and hot charge injection.

### 7.4 Pulsed current annealing of MOSFET dosimeter

It is concluded that the lifespan of a MOSkin dosimeter had been extended and seems to be reused many times while maintaining its sensitivity and linearity. In comparison to DC annealing, there was no need to pre-irradiate the MOSkin to improve its linearity or stability because it was ready to use directly after annealing. This will increase its importance in radiation medicine and space dosimetry, in addition to its unique advantages. It is concluded that pulsed annealing should be divided to two parts to recover its original sensitivity and linearity. The first part should start with high current pulses such as 1s – 20s to anneal the traps near the Si-SiO₂ interface. Then, a second annealing part should start with a low pulsed current.
over a longer time (~50s) to anneal the sensitive oxide layer (SiO$_2$). It was noted that the instability of the threshold voltage $V_{th}$ was in excellent stability, similar to the virgin one, ±1mV. This indicated that the interface traps had been eliminated. Another parameter that supports this is the transconductance of the annealed MOSkin detectors, which were similar to the virgin ones, as estimated from their I-V characteristic curves. The annealing time became shorter and more reliable after the consecutive annealing cycles.

### 7.5 Verification of Acuros XB on the build-up region

The verification of the build-up region for the tested setup for all cases showed an excellent agreement between Attix IC and MOSkin within ±1.5%, except for the oblique field where the MOSkin underestimated the surface dose by 3%. This was expected due to a difference in the WED of the measurements between the two detectors.

The two TPAs, Acuros XB and AAA were benchmarked against Attix IC and MOSkin. The results for 6 MV X-ray beam showed that Acuros XB was within 20% agreement with MOSkin for 4 cm x 4 cm in a surface dose. A similar agreement was observed between Acuros® XB, Attix, and MOSkin for 10 cm x 10 cm and 40 cm x 40 cm, 6 MV X-ray fields, but with an 18 MV X-ray beam, Acuros® XB was slightly worse at ~ 30%. The AAA was much worse at estimating the surface dose for all cases, with differences of up to 100% compared to Attix and MOSkin. At deeper depths, both algorithms converged to the Attix and MOSkin measurements, while Acuros® XB was faster than AAA to converge to the measurements. To conclude, Acuros® XB can be improved with better treatment for the backscattered radiation.
with higher energy and larger fields. In addition the Acuros® XB configuration can be tuned to match the measured surface dose.

7.6 Verification of Acuros® XB around high and low densities inhomogeneities

Interface dosimetry showed that Acuros® XB performed very well and was within ±2.5% agreement with MC calculations for all the interfaces of Air, Steel, and lung, and for the used photon beams of 6 MV and 18 MV. For steel heterogeneity, the backscattered peak shifted by ~2 mm from its expected depth. For air the AAA code did not treat the case very well in that it almost ignored the air gap in its calculation, which was already defined as an inhomogeneity. For cases of lungs with a narrow beam, the Acuros® XB calculated the dose voids very well. And as expected, a decrease in the dose for 18 MV photon beam was greater than the 6 MV photon beam because the lateral electron range for the 18 MV photon beam is greater than the 6 MV photon beam. In general both algorithms performed well for lung cases, with acceptable differences from MC. The Acuros® XB was within ±2.5% and the AAA was within ±5% for 6 and 18 MV photon beams respectively.

The AAA algorithm does not treat the backscattering component very well; as was shown with the steel heterogeneity when AAA did not calculate the backscattered peak at all, and it also underestimated the interfacial dose at the proximal side by 30% and 6% at the distal side compared to MC.
REFERENCES


