Evaluation of Dose Calculation Accuracy for Laryngeal Cancer Radiation Therapy Treatment Plans

by

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Abstract

Treatment planning systems use dose calculation algorithms to determine how much dose is being delivered to cancerous tumours and surrounding tissue during radiation therapy treatments. The purpose of this work is to determine the accuracy of dose calculation for treatment plans of laryngeal cancer with varying beam energies and field sizes. This work looked at Clarkson integration, FFT convolution, superposition, and Monte Carlo simulation calculation algorithms with cobalt-60, 6 MV, and 10 MV treatment beam in field sizes from 4x4 to 10x10 cm². MOSFETs and radiochromic film were characterized and used to measure absolute dose values in homogeneous and heterogeneous neck phantoms. Calculations were compared with measurements to determine the accuracy of each treatment plan.

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Glossary of Terms

Charged particle equilibrium (CPE): exists for a volume v if each charged particle of a given type and energy leaving v is replaced by an identical particle entering

Collisional KERMA: expectation value of the primary photon energy transferred to charged particles per unit mass, excluding the radiative-loss energy (units of Gy or J/kg)

Dose: expectation value of the energy imparted to matter per unit mass at a point (units of Gy or J/kg)

Energy deposition kernel (EDK): 3D matrix which describes the energy deposited by charged particles set into motion by primary or scattered photons

Field size: surface area covered by the radiation beam on the surface of the phantom or patient (units of cm^2)

Kinetic energy released per unit mass (KERMA): expectation value of the primary photon energy transferred to charged particles per unit mass (units of Gy or J/kg)

Linear attenuation coefficient: extent to which the intensity of a photon beam is reduced per unit distance as it passes through a specific material (units of cm^{-1})

Linear energy transfer (LET): charged particle energy loss per unit distance (units of MeV/cm)

Normal tissue control probability (NTCP): likelihood of complication when a healthy organ receives dose during radiation treatment

Optical density: measure of a transparent object transmittance

Pathlength: expectation value of the distance travelled by a photon as it passes through a specific medium (units of cm)

Range CSDA: average distance a charged particle will travel, calculated using the continuous slowing down approximation (units of g/cm^2)

Scatter-air ratio (SAR): ratio of scattered dose and the dose to a small mass in air at the same source to axis distance

Source to axis distance (SAD): distance from the source of linear accelerator target to a well defined point (units of cm)

Source to surface distance (SSD): distance from the source of linear accelerator target to the surface of the phantom or patient (units of cm)

Stopping power: charged particle energy loss per unit thickness of a specific medium (units of $\text{MeV}\cdot\text{cm}^2/\text{g}$)

Threshold voltage: minimum gate voltage required to create an inversion layer in the substrate region below the oxide layer in a MOSFET (units of mV)

Tissue-air ratio (TAR): ratio of the dose in water at point Q on the central beam axis in the patient or phantom to the dose to a small mass of water in air at the same point Q on the central beam axis

Total energy released per unit mass (TERMA): expectation value of the total primary photon energy transferred to charged particles or scattered photons per unit mass (units of Gy or J/kg)

Transient charged particle equilibrium (TCPE): state beyond depths equal to the maximum forward charged particle range in a field greater than twice the lateral charged particle range

Tumour control probability (TCP): probability of having no surviving malignant cells in a tumour after radiation treatment

Chapter 1

Introduction

Cancer is a diverse and deadly disease. Classified by uncontrolled cell division and failed programmed cell death, cancer can form in any part of the body and affects people of any age, race, or gender. The Canadian Cancer Society estimates that 166,400 people will be diagnosed with and 73,800 people will die from cancer in 2008 [1]. With current trends, over 40% of Canadians will develop cancer at some point in their lives, and 1 in 4 will die from it.

Laryngeal cancer forms when cells lining the larynx (voice box) become malignant. The larynx is approximately 5 cm long and connects the throat to the trachea (windpipe). It plays a key role in talking, breathing, and swallowing. Laryngeal cancer represents approximately 1% of all cancers and has a high mortality rate. It is most common in males over 50 years of age. One half of all males who develop laryngeal cancer will die from it. Major risks associated with developing the disease include excess smoking and/or drinking, exposure to sulphuric acid, having a human papillomavirus infection, and not eating enough fruits and vegetables. [1]

Radiation therapy is a common treatment for cancer. Shortly after the discovery of X-rays in 1895, German physicist Wilhem Röntgen determined that ionizing photons could be used to treat cancerous tumours. X-rays became more readily available in the early 1900's with the discovery of radium and the popularity of radiation treatment ensued. Following World War II, the production and delivery of X-ray radiation was dominated by cobalt-60 machines. Cesium units and betatrons were also popular. Although cobalt-60 units are still used in developing countries and for palliative

treatment, they have largely been replaced by linear accelerators which produce higher energy X-rays. High energy photons can be advantageous because they have greater skin sparing effects. They penetrate deeper into the patient before interacting with surrounding tissue and depositing energy. Other advancements in radiation therapy include the introduction of computed tomography (CT) imaging and multiple-beam treatment delivery. CT imaging allows radiation treatment plans to be based on the patient's 3-D anatomy. Using multiple radiation beams during the treatment of a tumour helps spare surrounding healthy tissue.

The main objective of a radiation therapy treatment plan is to maximize the amount of energy deposited to the tumour while avoiding complications. Conformal radiation therapy uses higher energy photons and more beams with smaller fields compared to traditional treatment. This helps concentrate dose to the tumour and spare healthy tissue. However, conformal radiation therapy comes with its own set of problems. Conformal treatment plans have regions with high dose gradients and patient setup must be consistent and accurate to avoid shifting the dose profile off the tumour and onto healthy tissue. Also, using multiple beams may increase the amount of energy deposited to the skin and increases the time it takes to calculate dose profiles.

Physicians and physicists use treatment planning systems to calculate how much dose is being delivered to a patient and where the dose is being deposited. Dose calculation is an important step in radiation therapy because the dose delivered to the tumour and to healthy tissue will determine a patient's outcome. Tumour control probability (TCP) is the probability of having no surviving malignant cells in a tumour after radiation treatment [2]. TCP depends on the type of cancer, size of the tumour, how

much radiation is delivered, and how many treatments (fractions) the dose is delivered in. Normal tissue complication probability (NTCP) is the likelihood of complication when a healthy organ receives dose during radiation treatment. NTCP depends on the organ, how much radiation is delivered, and how many fractions the dose is delivered in. For laryngeal cancer, the parotid gland is the most important organs at risk. If it receives too much radiation the patient may experience xerostomia (dryness of the mouth). Treatment planning systems determine the treatment course which will maximize TCP without surpassing the allowable NTCP. The calculated dose distribution should be accurate and precise to ensure the course of treatment meets the necessary TCP and NTCP levels.

There are numerous techniques used to calculate dose. The accuracy of a calculated dose value depends on which technique the system uses, the location of the tumour, and several treatment variables including beam energy and field size. For the case of laryngeal cancer, the low density trachea region may lead to calculation uncertainties. Further, the high energy photons and small fields in conformal radiation therapy may amplify any calculation uncertainty caused by low density regions. There is disagreement amongst radiation therapy specialists as to whether the benefits of highly conformal dose profiles outweigh the risks associated with lower calculation accuracy.

This objective of this thesis is to study various dose calculation techniques and determine their accuracy when calculating dose to the larynx. Doses will be calculated for different treatment plans with varying beam energies and fields. MOSFETs and radiochromic film will be characterized as absolute dosimeters and used to measure the dose delivered based on each treatment plan. Calculated doses will be compared to measured doses and the uncertainty of each calculation algorithm in the presence of a low

density region will be assessed. Accuracy levels will then be compared to guidelines set by the American Association of Physicists in Medicine to determine whether high energy beams with small fields are appropriate for the treatment of laryngeal cancer.

Chapter 2

Radiation Physics and Treatment Planning

Tumour control probability and normal tissue complication probability both depend on delivered dose and are used to evaluate a patient's course of treatment. Treatment planning systems are used to calculate radiation dose to be delivered to a patient. The total desired dose as calculated by the planning system is typically broken into 15 to 35 equal fractions and delivered to the patient 5 times per week for 3 to 7 weeks.

The difference between the desired delivered dose, as determined by a treatment planner, and the actual delivered dose is due to several sources of uncertainty. Overall uncertainty can be broken into two categories: (1) treatment uncertainties, and (2) dose calculation uncertainties. Treatment uncertainties include beam stability, beam flatness, patient setup, organ motion, and the planned target volume. The approximate extent of treatment uncertainties is currently an estimated 4.3% at one standard deviation [3]. The American Association of Physicists in Medicine (AAPM) has set the overall desired accuracy as 5%. To meet this standard, dose calculation accuracy must be better than 2% (the overall uncertainty equals the quadrature sum of treatment and dose algorithm uncertainties) [3]. While preparing a treatment plan, the physicist aims to meet the standards set by the AAPM while providing the best possible treatment course. The goal of this work is to determine which calculation algorithm and treatment characteristics obtain a dose calculation accuracy of 2%.

Dose algorithm accuracy depends on the physics model, photon energy, field size, and patient or phantom composition. Dose calculation algorithms can be broken into two

classes: those which account for tissue inhomogeneities in the primary photon attenuation calculation, and those which account for tissue inhomogeneities in calculation of electron transport and primary photon attenuation. This chapter reviews the radiation physics which governs photon beam dose delivery and various treatment planning algorithms used to calculate dose including Clarkson Integration, Fast Fourier Transform Convolution, Superposition, and Monte Carlo simulation.

2.1 Photon Beams

Photon beam radiation can be delivered from many different sources including isotope machines and linear accelerators (linacs). A common isotope machine is the cobalt-60 unit, which was designed in Canada and first installed at the University of Saskatchewan in 1951 [5]. Cobalt-60 decays by beta emission to nickel-60 which quickly emits two photons of 1.173 and 1.332 MeV, generating a stable and predictable beam. In linacs, electrons are injected into a waveguide and gain energy from an oscillating electromagnetic field. The high energy electrons bombard a heavy metal target which is commonly tungsten or gold. X-rays are produced through Bremsstrahlung radiation when the high energy electrons interact with the target's nuclei. Characteristic radiation may also produce X-rays when the high energy electrons collide with and eject atomic electrons. As outer shell electrons relax into the orbitals of ejected electron they lose energy and release a photon. Bremsstrahlung and characteristic radiation is distributed over a spectrum of photon energies. Linac photon beams are named according to the strength of the electric field in the waveguide. If the field is 60 kV/cm and the tube is 1 m long, the linac is producing a 6 MV beam. Electron accelerated in

this waveguide would gain 6 MeV of energy before hitting the target and the maximum photon would also be 6 MeV.

Linacs produce a beam with a wide energy spectrum whereas cobalt-60 machine produce a photon beam with two distinct photon energies. The higher energy photons of the linac have therapeutic benefits which include greater skin sparing (lower surface dose) and larger depths of penetration. Once the photon beams have been generated, dose is delivered to a tumour in two stages: (i) the photons interact with the patient's atoms setting high energy electrons into motion, and (ii) the electrons transfer energy to surrounding tissue through Coulomb collisions.

A photon beam is defined by its energy, field size, and source-to-surface distance (SSD). Beam energy is either the electron kinetic energy immediately before they hit the target for a linac or the energy of characteristic radiation emitted by a source machine. Field size is the area (cm^2) which primary photons pass on the surface of the patient or phantom. All fields in this thesis will be squared. Source-to-surface distance is the distance (cm) from the radiation source to the surface of the patient. SSD will always be 80 cm and 100 cm for cobalt-60 and linacs, respectively.

Depth of the measurement point 'P' in the patient or phantom will be denoted with 'z'. The distance from the source to the point of measurement (known as source-to-detector distance, SDD) the sum of SSD and depth (see figure 2.1).

Figure 2.1 Definition of SSD, SDD, field, and depth 'z' in the patient or phantom.

2.2 Dose Delivery

2.2.1 Primary Interactions

Primary photons are those which have not lost energy or changed direction since leaving the treatment machine. They can interact with medium through photoelectric absorption, Compton scattering, or pair production. Photoelectric absorption occurs when the energy of an incoming photon is absorbed and an atomic electron is ejected. Compton scattering occurs when a portion of the incoming photon energy is absorbed and used to eject an atomic electron (see figure 2.2). Pair production is the conversion of photon energy into an electron and a positron (antiparticle of the electron).

The probability of an interaction occurring is based on photon energy, effective atomic number of the interacting mass (Z_{eff}), and mass density (ρ). The total linear attenuation coefficient (μ) is the sum of all interaction probabilities. Photon pathlength (l) is the average distance a photon will travel before interacting and is equal to the inverse of the total linear attenuation coefficient. Photon fluence, the number of photons to intersect an area, attenuates exponentially through a medium according to the following equation:

$$\Phi(z) = \Phi_0 e^{-\mu z} \quad [2.1]$$

where z is the distance traveled through the medium, Φ_0 is the incident photon fluence, and $\Phi(z)$ is the photon fluence at a depth z . When photons with energies between 1 and 10 MeV interact in a water-like tissue (effective atomic number ~ 7.5), Compton interactions dominate and the total attenuation coefficient is approximately equal to the Compton linear attenuation coefficient ($\mu \approx \sigma_{\text{Compton}}$). The Compton linear attenuation

coefficient is a function of atomic number, weight, and density, according to equation 2.2:

$$\sigma_{Compton} \propto \left(\frac{Z}{A}\right) \rho \sigma_{KN} \quad [2.2]$$

where σ_{KN} is the Klein-Nishina cross-section between a photon and a free, static electron [5]. The component $\left(\frac{Z}{A}\right) \rho$ is equal to the electron density (ρ_e) of the medium. At beam energies less than 100keV the photoelectric effect becomes significant. At higher energies and atomic numbers pair production dominates [5].

Figure 2.2 Diagram of a Compton Interaction where an incoming photon sets an atomic electron into motion and is scattered away at a lower energy. The electron loses energy through Coulomb collisions and sets a second electron (delta-ray) into motion. Bremsstrahlung radiation is also created.

2.2.2 TERMA, KERMA, and Dose

Total energy released per unit mass (TERMA) is defined as the product of energy fluence (Ψ) and the total mass attenuation coefficient according to:

$$T(z) = \Psi(z) \left(\frac{\mu}{\rho} \right) \quad [2.3]$$

TERMA represents the energy transferred from primary photons to high energy electrons or scatter photons in a volume of mass dm . The high energy electrons will subsequently lose energy locally through Coulomb collisions and may set other electrons into motion (known as delta rays, see figure 2.2). If the electron kinetic energy is greater than 1 MeV, some of the energy may be converted to radiation through Bremsstrahlung events.

Kinetic energy released per unit mass (KERMA) is the fraction of TERMA which is transferred to high energy electrons. Collisional KERMA (K_c) is the fraction of KERMA which is transferred to the medium through Coulomb collisions. Collisional KERMA is defined by energy fluence and the mass-energy absorption coefficient (μ_{ab}) according to the following equation:

$$K_c(z) = \Psi(z) \left(\frac{\mu_{ab}}{\rho} \right) \quad [2.4]$$

When the primary photon energy is less than 1 MeV, KERMA and collisional KERMA are equal. Absorbed dose is the energy deposited per unit mass dm of a medium and is calculated as follows:

$$D = \frac{dE}{dm} \quad [2.5]$$

Dose is measured in units of Gray (Gy) where one Gray is equal to one Joule of radiation energy per kilogram of mass.

2.2.3 Stopping Power

Particles which contribute to collision KERMA can be distributed over a spectrum of initial energies. For an electron at a specific energy, the collisional energy lost per unit thickness ($\text{MeV}\cdot\text{cm}^2/\text{g}$) is given by the mass collisional stopping power (S_{col}) according to:

$$S_{\text{col}}(E) = \frac{1}{\rho} \left(\frac{dE}{dx} \right) \quad [2.6]$$

Linear energy transfer (L) is the collisional energy lost per unit distance (MeV/cm) and can be calculated as the product of mass collisional stopping power and mass density according to:

$$L = \frac{dE}{dx} \quad [2.7]$$

As an electron slows down, energy is lost at a greater rate. The average energy lost per unit thickness by an electron with a specific initial energy as it slows to a resting position is the mean mass collisional stopping power, $\overline{S_{\text{col}}}(\overline{E}_i)$. The average mass collisional stopping power, $\overline{\overline{S_{\text{col}}}}(h\nu)$, is the average energy lost per unit thickness for a set of electrons with a spectrum of initial kinetic energies set into motion by a photon with energy $h\nu$. Mass collisional stopping power is a function of effective atomic number and photon energy, and is nearly independent of density [3].

The range of an electron is the average distance it will travel from the photon interaction site before coming to a rest. Range is often calculated using a continuous slowing down approximation (CSDA range, R_{CSDA}) where electrons lose energy at a constant rate according to the following equation:

$$R_{CSDA} = \int_0^{E_i} \frac{dE}{\bar{S}} \quad [2.8]$$

A more realistic model would calculate electrons losing energy in discrete steps with each Coulomb collision. This would require tracking individual particles and would be computationally intensive.

2.2.4 Charged Particle Equilibrium

Charged particle equilibrium (CPE) occurs when the number of electrons set into motion by primary photons equals the number of electrons coming to rest in a small volume of mass dm . Under the condition of CPE, primary dose can be computed with equation 2.4 because dose is equal to collision KERMA. However, due to attenuation and beam divergence, photon beams cannot achieve pure CPE.

Transient charged particle equilibrium (TCPE) can occur in a photon beam along the central axis if longitudinal and lateral equilibrium are achieved. Longitudinal equilibrium occurs at depths greater than the maximum forward range of charged particles. Lateral equilibrium occurs when the distance from the central axis to the field edge is greater than the maximum lateral electron range (typically about 1/3 of the maximum forward range). Electron range is greater in low density media and larger field sizes are required to establish lateral CPE. This is also true for high energy electrons. Under the condition of TCPE, primary dose is proportionately larger than collision KERMA because the photon energy is carried downstream and deposited at an average distance $\bar{z}$ [3] according to:

$$D_p = e^{\mu\bar{z}} K_c(z) \approx (1 + \mu\bar{z}) K_c(z) \quad [2.9]$$

If TCPE is established and the linear and energy absorption coefficients are known, primary dose along the central axis can be determined from equations 2.4 and 2.9.

2.2.5 Equivalent Pathlength

Density scaling is a technique which relates photon attenuation in a heterogeneous medium to that of a homogeneous medium, typically water. The equivalent pathlength (EPL) is the pathlength of a photon in a given medium scaled by mass density according to the following equation:

$$l_{H_2O} \approx EPL = \left(\frac{\rho_{med}}{\rho_{H_2O}} \right) l_{med} \quad [2.10]$$

where l_{med} is the pathlength in the original medium and l_{H_2O} is the pathlength in water. The O'Connor density-scaling theorem uses EPL to relate the dose in different media of equal atomic composition [3]. According to the theorem, dose at corresponding points in the media are equal if the geometry and fields are scaled inversely with mass density. O'Connor assumed the ratio of primary photon fluence to scatter fluence remains constant at equal equivalent pathlengths. However, it has been shown that if the effective atomic number of a medium is close to 7.5 and the photon energy is between 1 and 10 MeV, scaling should be done with electron density in place of mass density [6]. This is because Compton interactions dominate under those conditions and the mass attenuation coefficient is proportional to electron density (see equation 2.2). The electron equivalent pathlength (eEPL) is calculated as follows:

$$l_{H_2O} \approx eEPL = \left(\frac{\rho_{e,med}}{\rho_{e,H_2O}} \right) l_{med} \quad [2.11]$$

where ρ_{med} and ρ_{eH_2O} are the electron atomic densities in the medium and water, respectively. Figure 2.3 compares pathlength with EPL and eEPL in water with various phantom materials. With the mass density scaling method (figure 2.3a) EPL overestimates the true pathlength by up to 10% in air at low energies, and 5% in acrylic at higher energies. Using electron density scaling method (figure 2.3b) eEPL correctly estimates pathlength within 3% for energies less than 10 MeV.

2.2.6 Tissue-Air Ratio

The tissue-air ratio (TAR) is defined as the ratio of dose at a specific depth in a medium to dose in a small volume of mass in free air (D_{air}) at the same source to axis distance (SAD, see figure 2.1). It depends on depth in the medium (z), field size (f), and beam energy (E). TAR is calculated according to:

$$TAR(z, f, E) = \frac{D(z, f, E)}{D_{air}} \quad [2.12]$$

Figure 2.3 Ratio of true pathlength and equivalent pathlength in water calculated by (a) mass density scaling (equation 2.10), and (b) electron density scaling (equation 2.11). Data taken from NIST X-ray attenuation database [7].

2.3 Clarkson Integration

Clarkson integration is a method to calculate delivered dose, and was first introduced in 1941 [8]. Clarkson integration algorithms calculate dose to water using measured dosimetric data and fundamental principles of physics. The algorithms separate the scatter dose component, which depends on the field size and depth, from the primary dose component, which depends only on depth. The algorithms can account for variations in primary fluence due to inhomogeneities. They do not model variations in scatter due to tissue inhomogeneities or surface curvatures [9].

2.3.1 Dose Algorithm

Clarkson integration algorithms calculate dose in a patient or phantom using known tissue-air ratios and measured doses. Calibration measurements are taken in varying fields at a calibration depth (z_c) beyond the maximum range of the charged particles. For a given field size (f) the calibration dose (D_c) is divided by the TAR with equivalent parameters:

$$D_c(z_c, f) / TAR(z_c, f) \quad [2.13]$$

Dividing by the TAR gives the dose in air at the point of calibration. To shift the dose to the point of interest, equation 2.13 is multiplied by an inverse square factor:

$$\left[\frac{D_c(z_c, f)}{TAR(z_c, f)} \right] \times \left(\frac{s_c}{s} \right)^2 \quad [2.14]$$

where ' s_c ' and ' s ' are the SADs for the calibration depth and the point of measurement, respectively. Dose is calculated by multiplying equation 2.14 by the TAR for the equivalent pathlength at the depth of interest according to the following equation:

$$D(z, f) = \left[\frac{Dc(z_c, f)}{TAR(z_c, f)} \right] \times \left(\frac{s_c}{s} \right)^2 \times TAR(l_{H20}, f) \times C_{Clarkson} \quad [2.15]$$

where $C_{Clarkson}$ is a correction factor to account for beam hardening (increase in the average primary photon energy with depth).

2.3.2 Scatter-Air Ratio

TAR parameterizes the build-up of scatter in a phantom and can be broken into the following components:

$$TAR(d, f) = TAR(d, 0) + SAR(d, f) \quad [2.16]$$

The first component, $TAR(d, 0)$, represents the contribution from primary photons and is equivalent to the TAR value in a $0 \times 0 \text{ cm}^2$ field. It is calculated by extrapolating measured TAR values as the field size approaches zero [9]. The second component represents the contribution from scatter and is titled the scatter-air ratio (SAR). SAR is calculated by dividing the field into several pie-shaped sectors and summing the scatter contributions from each section (see figure 2.4). The scatter contribution from each section depends on depth and the average distance from the measurement point to the edge of the field.

2.3.3 Accuracy

Clarkson integration accounts for variation in primary energy fluence by scaling pathlength according to electron density. It does not account for the loss of lateral CPE in low density regions. The algorithm also does not account for surface curvature. If the phantom face is curved in such a manner that it scatters photons less than the calibration phantom, the algorithm will not predict a decrease in dose. This problem is often corrected for by adding bolus material to a patient to create a flat surface.

Figure 2.4 Clarkson integration: The rectangular field is broken into 20 equal parts and a SAR value is assigned to each section according to the average distance from the calculation point to the field edge. The total SAR value for the field is the integral sum of all sectors.

2.4 FFT Convolution and Superposition

FFT convolution (FFTC) and superposition are 3-dimensional dose calculation methods which model primary and secondary particle fluence. They are identical methods with the exception of how they handle inhomogeneity. Both techniques model photons as energy sources at the point of interaction and blur the energy into surrounding tissue by convolving TERMA with energy deposition kernels. Dose is calculated according to the following equation:

$$D(\bar{z}) = \int T(\bar{z}') K(\bar{z} - \bar{z}') d^3\bar{z}' \quad [2.17]$$

where z is the dose calculation point, z' is the point of interaction, $T(z)$ is the TERMA at the primary photon interaction site, and $K(z-z')$ is the value in the energy deposition kernel at the calculation point z from a interaction at z' .

2.4.1 Polyenergetic TERMA

TERMA is the major component in determining the distribution of dose in a patient or phantom [10]. For a monoenergetic beam, TERMA is a function of incident energy fluence, depth, and linear attenuation coefficient:

$$T(z) = \Psi_0 e^{-\mu z} \left(\frac{\mu}{\rho} \right) \quad [2.18]$$

However, most clinical beams are polyenergetic and equation 2.18 does not apply. Mohan and Boyer suggested adjusting the superposition model by integrating over all energies while convolving TERMA and the energy deposition kernel [11, 12]. This technique, while properly representing dose, is computationally expensive and slows the process by a factor approximately equal to the number of energy levels [12]. Metcalfe proposed creating a single TERMA matrix from the incident polyenergetic spectrum so

only one convolution is necessary [13]. To create a single TERMA matrix, the energy spectrum is broken into discrete levels and the polyenergetic TERMA at the interaction site z' is represented by the sum of individual monoenergetic TERMA values over the energy spectrum [13]. It can be calculated according to:

$$T(z') = \sum_i T(z', E_i) = \sum_i \bar{E}_i \Phi_i \left(\frac{\mu_i}{\rho} \right) \quad [2.19]$$

where Φ_i is the photon fluence of the i^{th} energy level at the interaction site.

2.4.2 Energy Deposition Kernels

Energy deposition kernels represent the three dimensional transport and deposition of energy from a primary photon interaction site. They are geometrical matrices where the origin represents the primary photon interaction site. Each energy deposition kernel voxel depends on the distance and angle from the origin, and represents the portion of primary photon energy deposited to that volume.

Many FFTC and superposition algorithms use the monoenergetic energy deposition kernels generated by Mackie *et al.* with the EGS Monte Carlo code [14]. Primary monoenergetic photons were forced to interact at the centre of a 60 cm water sphere and the histories of secondary particles were tracked [15]. Photon energies ranged from 0.1 to 50 MeV. Five kernels were created: primary, first scatter, secondary scatter, multiple scatter, and Bremsstrahlung with annihilation photons [15]. The kernels are spherical functions dependent on the polar angle and radius.

FTTC and superposition algorithms typically use only two energy deposition kernels. The first kernel models electrons set in motion by primary photons, and deposits a significant amount of energy over a small range. The second kernel models electrons

set in motion by scattered photons, and spreads a smaller amount of energy over a larger range. Separating the primary and scatter components is a speed optimization technique because each convolution uses its own parameters. Dose from primary photons is calculated using a high resolution grid over a small region. Dose from scatter photons is calculated using a lower resolution grid over a larger region. Using high and low resolution calculations reduces computational time by approximately 65% [16].

When polyenergetic primary photons interact in a volume, dose is calculated with polyenergetic energy deposition kernels. Generating polyenergetic energy deposition kernels is not a simple superposition calculation as with the polyenergetic TERMA because photons of different energies have different probabilities of interacting and the kernel values should be weighted accordingly. The i^{th} energy level's energy deposition kernel is weighted by the fraction of photons in the bin (f_i) and the linear attenuation coefficient [10]. To generalize the equation for a spectrum of primary photon energies, the fraction of photons in the each bin is be attenuated exponentially. Under these conditions, the polyenergetic energy deposition kernel becomes a TERMA weighted superposition of the monoenergetic kernels and is calculated as follows:

$$K(z) = \frac{\sum_i f_i e^{-\mu_i z} \mu_i E_i K_i}{\sum_i f_i e^{-\mu_i z} \mu_i E_i} = \frac{\sum_i T_i(z) K_i}{T(z)} \quad [2.20]$$

where K_i is the monoenergetic kernel value and $K(z)$ is the polyenergetic kernel value at depth z .

2.4.3 Kernel-Hardening Correction Factor

In equation 2.20, a new kernel is generated for each depth increment. However, this approach is not computationally practical for clinical use. Most treatment planning

systems generate a single kernel for the surface spectrum and apply a correction factor for changes in the kernel. Hoban [10] suggested a correction factor (C_{KH}) based on the variation of collisional KERMA to TERMA with depth according to:

$$C_{KH}(z) = \frac{\left(\frac{K_c(z)}{T(z)} \right)}{\left(\frac{K_c(0)}{T(0)} \right)} \quad [2.21]$$

The kernel-hardening correction factor is applied to both the primary and scatter kernels. It correctly accounts for the increase in linear energy absorption coefficient to linear attenuation coefficient with depth. Failure to account for kernel-hardening would result in an underestimation of dose. C_{KH} neglects the change in spatial energy distribution which may be significant if the beam radius is less than the lateral size of the energy deposition kernels and lateral CPE is lost [10].

2.4.4 Fast-Fourier Transform Convolution

FFTC algorithms use spatially invariant kernels which represent the distribution of energy from a primary photon interaction site in water. The spherical energy deposition kernels developed by Mackie *et al.* [15] are interpolated to Cartesian coordinates using adaptive quadrature techniques. Because the energy deposition kernels are constant for all calculation points, equation 2.17 is simplified by converting to Fourier space according to the following equations:

$$D_p(z) = \int C_{KH}(z') T(z') K_p | \bar{z} - \bar{z}' | dz'^3$$

$$D_p(z) = \mathfrak{F}^{-1} \left[\mathfrak{F}(C_{KH}(z') T(z')) \mathfrak{F}(K_p | \bar{z} - \bar{z}' |) \right] \quad [2.22]$$

where $K_p|r-r'|$ is the primary kernel value at dose point z from the interaction point z' . A similar equation applies for the scatter dose, D_s , and total dose is the sum of the primary and scatter components.

When a field is less than twice the lateral range of scattered electrons, lateral equilibrium along the central axis is lost and dose is reduced. FFTC algorithms can account for the loss of lateral equilibrium if the calculation is done for a water phantom. Kernels with interaction points outside the photon beam are not summed and dose is properly reduced. FFTC algorithms may not properly account for the loss of lateral equilibrium in low density media. The increased electron range is not considered because the energy deposition kernels are invariant. Dose along the central axis will decrease when electron range increases because energy is being scattered away.

Figure 2.5 shows how energy is spread from a primary photon interaction site in high and low density media. The energy deposition kernel in high density media (figure 2.5a) deposits more energy close to the interaction site while the energy deposition kernel in low density media (figure 2.5b) spreads the energy further away. For electrons with the same initial kinetic energy in different density media, the sum of deposited energy over individual kernels is equal. Figure 2.5c shows an example of a water energy deposition kernel whose lateral extent is within the photon beams. Lateral CPE is established along the central axis. Figure 2.5d shows an example of an air energy deposition kernel in the same photon beam as figure 2.5c. The lateral extent of this kernel is not within the photon beam and lateral CPE is not established along the central axis. For this case, FFTC algorithms would calculate dose using the kernel of figure 2.5c and predict a higher dose than would be predicted if the kernel from figure 2.5d was used.

Figure 2.5 Energy deposition kernel for an electron in (a) high density medium, (b) a low density medium. An energy deposition kernel in (c) water whose lateral extent is within the photon beam, (d) air whose lateral extent is beyond the photon beam.

2.4.5 Superposition

Superposition algorithms are an adaptation of the collapsed cone technique [14,17]. The algorithms use variable energy deposition kernel which can account for changes in electron transport when a photon beam is in a non-water or inhomogeneous phantom. The distance travelled by a scattered electron, d , is scaled by the ratio of average electron density between the interaction and calculation points, $\bar{\rho}$, and electron density in water:

$$\tilde{d} = \left(\frac{\bar{\rho}_e}{\rho_{e-H_2O}} \right) d \quad [2.23]$$

where $\tilde{d}$ is the estimated distance the electron will travel. Equation 2.23 is similar to equation 2.11 where photon pathlength is scaled with electron density. This is meant to account for differences in electron range in water, where the energy deposition kernels were calculated, and the media in which the electrons travel. The values in the energy deposition kernel are then scaled by the ratio of electron density at the interaction point, $\rho_e(z')$, and the average electron density between the interaction and calculation points, $\bar{\rho}_e$ according to the following equation:

$$D_p(z) = \int C(z') T(z') \frac{\rho_e(z')}{\bar{\rho}_e} K_p \left(\frac{\bar{\rho}_e}{\rho_{eH_2O}} (\bar{z} - \bar{z}') \right) d^3 z' \quad [2.24]$$

The second scaling takes into account the difference in the amount of kinetic energy released per voxel caused by increasing or decreasing electron range. If electron range is increased, the energy deposition kernel becomes longer and wider. Since the total energy deposited must be constant, individual voxel values must decrease.

Figure 2.6 shows the ratio of the average linear energy transfer in air and that of water, scaled by electron density, calculated as follows:

$$\left(\frac{\overline{\overline{L}}(h\nu)_{air}}{\overline{\overline{L}}(h\nu)_{water}} \right) \left(\frac{\rho_{e - water}}{\rho_{e - air}} \right) \quad [2.25]$$

where $\overline{\overline{L}}(h\nu)$ is the average linear energy transfer of electrons set into motion by a photon of energy $h\nu$. For all photon energies below 4 MeV, the average linear energy transfer in air is less than the electron density scaled average linear energy transfer in water. Therefore, the electron density scaling method of equations 2.23 and 2.24 would underestimate the average distance travelled by the set of electrons set in motion by any photon of energy less than 4 MeV when the true medium is air.

Figure 2.6 Ratio of the average linear energy transfer in air and water scaled by electron density (equation 2.25). Data taken from *The Physics of Radiology*, Johns and Cunningham, A4 [5].

2.4.6 Computation Time

Superposition algorithms are computationally slower than FFTC algorithms but may provide more accurate dose calculations in the presence of tissue inhomogeneities. Three techniques are used to improve computation time.

Dose vs. Interaction Point of View: In the FFTC model, dose is collected from interaction sites around the point of interest. Using this ‘interaction point of view’, dose is calculated at all sites on the beam grid, even if they are not relevant points for treatment planning. Using the ‘dose point of view’, energy deposition kernels are inverted and the origin is placed at the calculation point. TERMA is then collected at the calculation point. This allows the system to calculate dose from only the relevant points.

Ray Traces: FFTC uses the original 48 evenly spaced zenith array angles presented by Mackie [14, 15]. The standard superposition algorithm uses only 16 zenith angles. Accuracy is improved by distributing most of the rays in the forward direction where the energy distribution is concentrated. The Fast Superposition algorithm uses 6 zenith rays resulting in a reduction of computation time by 250% at the cost of 1-2% accuracy [14].

Multigrid: In the multigrid technique, regions of high and low dose gradients are identified. If the calculation point is in a low gradient region, dose is calculated with coarse resolution and interpolated onto the original grid. If the calculation point is in a high gradient region, dose is calculated with the finer grid resolution.

Superposition algorithms using the multigrid technique will be known as fast superposition for the remainder of this thesis.

2.5 XVMC

X-ray voxel based Monte Carlo (XVMC) is an algorithm based on fundamental radiation physics. Clarkson integration scales a calibration dose measured in water and FFTC/superposition calculates dose with a water kernel. These methods are calculating the dose to water (D_w). XVMC algorithms simulate particle transport in the medium and calculate dose to medium (D_m). This approach is a straightforward reproduction of the true means by which energy is deposited in media, and as such produces accurate dose distributions. Historically, Monte Carlo algorithms were not practical for routine treatment planning because of long calculation times. However, computational improvements and new optimization techniques have reduced overall calculation times, and some treatment planning systems are beginning to use Monte Carlo algorithms.

2.5.1 Monte Carlo Integration and simulation

The Monte Carlo simulation technique is a statistical approach to numerical integration. If $f(x_1, \dots, x_m)$ is square-integrable and depends on m variables in a unit hypercube $[0,1]^m$ then the Monte Carlo approximation of the integral of $f(x)$ is:

$$I = \int f(x) dx \approx \frac{1}{N} \sum_{n=1}^N f(x_n) \quad [2.26]$$

where x_n is a random point on the unit hypercube. According to the law of large numbers, the Monte Carlo integral will converge to the true value with an increasing number of samples.

The Monte Carlo simulation technique applied to radiation transport tracks the deposition of photon energy by predicting the distance a photon will travel to an interaction site and the type of interaction to occur. When secondary particles are set into motion, the algorithm predicts their energy and trajectory. The first version of XVMC

considered Compton interactions and pair production, neglecting photoelectric absorption [18]. However, in the energy range of interest, photoelectric absorption is a simple process and does not consume a significant amount of processor memory. It was added to later versions of XVMC [19].

2.5.2 Primary Photon and Secondary Charged Particle Interactions

The probability of a primary photon interacting, $p(z)$, in a layer of tissue dz is defined by the exponential decay with the linear attenuation coefficient (see equation 2.1) and is given by:

$$p(z)dz = e^{-\mu z} \mu dz \quad [2.27]$$

The probability is set equal to a random number between zero and one (R_1), and integrated over all depths. The resulting depth is the point of interaction according to:

$$\int_0^z p(z') dz' = R_1 \Rightarrow z = \frac{\ln(1 - R_1)}{\mu} \quad [2.28]$$

At the determined depth z , the type of interaction to occur is based on a discrete probability function. The interaction to occur is determined by:

<i>if</i>	$R_2 < \kappa_{PP} / \mu$	the interaction is pair production
<i>else if</i>	$R_2 < (\kappa_{PP} + \tau_{PE}) / \mu$	the interaction is photoelectric
<i>else</i>		the interaction is Compton

where R_2 is a second random number between zero and one, and τ_{PE}/κ_{PP} are the linear attenuation coefficients for photoelectric absorption and pair production.

Once the type of interaction is determined, a similar process determines the direction and energy of scatter particles. In Compton scatter, one electron and one

photon are released (figure 2.2). The electron is scattered away at an angle ϕ with energy T and the photon is scattered at angle θ with energy $h\nu'$ according to:

$$T = h\nu \cdot \frac{\alpha(1 - \cos\theta)}{1 + \alpha(1 - \cos\theta)} \quad [2.29]$$

$$h\nu' = h\nu \frac{1}{1 + \alpha(1 - \cos\theta)} \quad [2.30]$$

where $h\nu$ is the incoming photon energy and α is the ratio of photon energy to the rest mass of the electron ($h\nu/m_0c^2$). The XVMC algorithm determines the ratio of scattered photon energy to incoming photon energy (ε) based on the incoming photon energy and differential Compton cross-section according to:

$$\frac{d\sigma}{d\varepsilon} \propto \frac{1}{\varepsilon} + \varepsilon \quad [2.31]$$

Once ε is determined, all other relevant variables (T , $h\nu'$, ϕ , θ) can be calculated [20].

In pair production a photon converts all its energy into the mass and kinetic energy of an electron positron pair, and is governed by the following equation:

$$h\nu = 2m_0c^2 + T_{e^-} + T_{e^+} \quad [2.32]$$

Note the photon must have more than 1.022 MeV for pair production to occur. Similar to the Compton process, the XVMC algorithm determines the ratio of electron energy to incoming photon energy based on a differential cross section, and all other relevant variables are determined from there. The positron is assumed to be an electron because their stopping power ratios are within 2% of each other [18].

In photoelectric absorption, a collision between a photon and an atom releases an electron. The kinetic energy, T , is the difference between photon energy, $h\nu$, and binding energy of the shell from which the electron was ejected, E_s , described by:

$$T = h\nu - E_s \quad [3.33]$$

The XVMC algorithm calculates the electron energy by substituting E_s with a weighted average of all the shell binding energies. The electron's direction is the same as the incoming photon. The inclusion of photoelectric absorption has a small (<1%) effect for energies of 1-2MeV, and is negligible for higher energies [19].

2.5.2 Speed Optimization Techniques

Clinical beams can have greater than 10^{16} particles to track. To track each particle would be computationally intense and not practical in the clinic. The condensed history technique was introduced by Berger in 1963. It reduces the number of histories the algorithm tracks [21]. Secondary electrons undergo $\sim 10^5$ interactions before coming to a rest, each causing very little change to the direction and energy of the electron. In the condensed history technique, interactions are grouped into discrete steps and the cumulative directional and energy change is applied. Accuracy of this approximation depends on the step size as well as the low-energy cut-off (energy level at which particles are no longer tracked). XVMC codes also use advanced variance reduction techniques which improve the accuracy of calculation while minimizing computation time, such as range rejection where an electron path is terminated if its energy is so low that it cannot escape the block it has been grouped into by the condensed history technique [21].

Chapter 3

MOSFET: A Point Dosimeter

The first metal-oxide-semiconductor field effect transistor (MOSFET) was created at Bell Labs in 1960 by Dawson Kahng and Martin Atalla [22], and has since become the most widely used type of integrated circuit. A MOSFET controls an electric signal by modulating the charge concentration of a capacitor. The capacitor is made of two conducting planes, a gate terminal and a silicon body (substrate), separated by an insulating silicon-oxide layer. Original gate electrodes were aluminum. However, modern MOSFETs use a non-metallic layer of polysilicon making the metal component of the MOSFET title misleading. The silicon body is lightly doped with either group VA (n-type) or group IIIA (p-type) impurity atoms to add abundant electron or hole charge carrier. A MOSFET also contains two highly doped regions, the drain and source, which have dopant atoms opposite to that of the body. If the drain and source are p-type while the substrate is n-type the device is known as a pMOSFET (see figure 3.1). The following discussion will cover pMOSFET design and physics unless otherwise stated.

When the body and source terminal of a MOSFET device are shorted and a negative voltage is applied to the gate, free electrons in the n-type substrate directly below the SiO_2 layer migrate away while holes build up. If the applied voltage is sufficient, a p-type channel connecting the drain and source forms and current is able to flow. The channel is known as an inversion layer because the n-type substrate material inverts to p-type. The minimum negative gate voltage required to create the inversion layer is known as the threshold voltage, V_{th} .

Figure 3.1 Diagram of a pMOSFET. Image adapted from Sedra and Smith "Microelectronic Circuits", [23]

3.1 MOSFETs as Radiation Detectors

The basic principle of MOSFET dosimetry follows the creation of electron-hole pairs in the SiO_2 layer when exposed to ionizing radiation. The number of charge carriers released is proportional to the amount of energy deposited or absorbed dose. Holes migrate to the Si/SiO_2 interface and become trapped while electrons move to the gate. A small portion of electron-hole pairs recombine, and an external positive bias at the gate helps minimize the recombination.

Trapped holes at the Si/SiO_2 interface create a stable internal polarization which attracts free substrate electrons. The minimum negative gate voltage required to create an inversion layer is now larger because it must overcome the internal polarization at the

Si/SiO₂ interface. The shift in threshold voltage before and after irradiation (ΔV_{th}) is proportional to the number of trapped holes and is calculated as follows:

$$\Delta V_{th} = \frac{Q_t X}{\epsilon_{ox} \epsilon_0} \quad [3.1]$$

where Q_t is the trapped charge at the Si/SiO₂ interface, X is the distance to the charge Q_t from the metal gate, and ϵ_{ox} and ϵ_0 are the SiO₂ and free space permittivity, respectively. The distance to the trapped charge X is approximately equal to oxide thickness, t_{ox} , and equation 3.1 can be approximated by the following:

$$\Delta V_{th} \approx \frac{Q_t}{C_{ox}} \quad [3.2]$$

where C_{ox} is the capacitance of the oxide layer. If recombination is minimal and consistent, the shift in threshold voltage will be proportional to dose.

The sensitivity of a MOSFET radiation detector is a function of the number of electron-hole pairs created which is a function of oxide thickness. The shift in threshold voltage has been found to exhibit a power-law relationship with the oxide layer thickness, and to maximize MOSFET sensitivity a thick oxide should be used [24]. However, increasing the oxide thickness adds mechanical stress to the device and the Si/SiO₂ interface may dislocate, leading to threshold voltage instability. Extremely thick oxide layers also require unpractical growth times at high temperatures. A typical MOSFET radiation detector will have an oxide layer 0.1 to 1.0 μm thick [25].

3.1.1 Single MOSFET Detectors

The original MOSFET detector designed for radiation therapy dosimetry consisted of a single n-type MOSFET [26]. This design gave linear correlation between threshold voltage and accumulated dose at a dose-rate of 100 cGy/min and a typical calibration

coefficient for this MOSFET detector was 0.2 cGy/mV. The linear response breaks down for high accumulated dose and varying temperatures [24]. The loss of proportionality between threshold voltage and absorbed dose at high accumulated doses is caused by the increasing strength of the internal electric field generated by trapped holes. The external bias becomes less effective and more recombination occurs. Linearity between threshold voltage and absorbed dose also depends on temperature because threshold voltage is inherently dependent on carrier mobility. A change in ambient temperature by a single degree can result in a 5 mV shift of threshold voltage, or 1 cGy of accumulated dose [27].

3.1.2 Dual-Bias-Dual-MOSFET Detectors

The dual-bias-dual-MOSFET detector was introduced in 1994 [24]. The new design placed two identical p-type MOSFETs on a single chip and biased them at different gate voltages. When a MOSFET biased at b_i is exposed to ionizing radiation, the number of holes (N_H) to reach the Si/SiO₂ interface is:

$$N_H(b_i) = \varepsilon(b_i)N \quad [3.3]$$

where N is the total number of electron-hole pairs created and $\varepsilon(b_i)$ is the efficiency of the device. The shift in threshold voltage is then:

$$\Delta V_{th} = \frac{N_H(b_i)}{C_{ox}} \quad [3.4]$$

where C_{ox} is the oxide capacitance. The dual-bias-dual-MOSFET detector measures the difference in threshold voltage change between the two identical MOSFETs and is calculated as follows:

$$\Delta V_{th1} - \Delta V_{th2} = \frac{N_H(b_1) - N_H(b_2)}{C_{ox}} = \frac{\varepsilon(b_1) - \varepsilon(b_2)}{C_{ox}} N \quad [3.5]$$

Equation 3.5 is proportional to the total number of electron-hole pairs created. Dose can then be calculated by multiplying the difference in threshold voltage change by an appropriate calibration coefficient (C_1) according to:

$$D = C_1(E) \cdot (\Delta V_{th1} - \Delta V_{th2}) \quad [3.6]$$

where $C_1(E)$ is a function of beam energy. When the device is exposed to a shift in temperature, the efficiency of trapping holes changes by a factor $\varepsilon(T)$ according to:

$$N_H(b_i) = [\varepsilon(b_i) + \varepsilon(T)]N \quad [3.7]$$

where $\varepsilon(T)$ is assumed to be independent of biasing. The change in threshold voltage then becomes:

$$\Delta V_{th1} - \Delta V_{th2} = \frac{\varepsilon(b_1) + \varepsilon(T)}{C_{ox}} N - \frac{\varepsilon(b_2) + \varepsilon(T)}{C_{ox}} N \quad [3.8]$$

which is equal to equation 3.5. According to equation 3.8 the dual-bias-dual-MOSFET detector is less sensitive to temperature than the single MOSFET detector.

3.1.3 MicroMOSFET Detectors

The dual-bias-dual-MOSFET detector showed an angular dependence of up to 15% and had limited clinical use [25]. Researchers discovered this dependence was a function of the silicon substrate thickness, and the need for a smaller transistor became apparent [25]. A re-designed dosimeter, the microMOSFET, has minimal angular dependence and aided in MOSFET technology's emergence into the clinic. The microMOSFET uses the dual-bias-dual-MOSFET design and is about half the size of its predecessor, $1.0 \times 1.0 \times 3.5 \text{ mm}^3$, with a $0.5 \text{ }\mu\text{m}$ thick active area of $0.2 \times 0.2 \text{ mm}^2$. They can be biased at high or standard sensitivity to accommodate the nature of the experiment. The manufacturer quotes a calibration coefficient of 0.33 cGy/mV and 1.00 cGy/mV for high

and standard sensitivity microMOSFETs with a standard bias supply. A high bias helps minimize recombination but lowers the lifetime of the device. A typical device has a lifetime of 18,000 mV, or 60 Gy for a high sensitivity and 180 Gy for a standard sensitivity [28]. The effective depth of measurement for a microMOSFET is in the SiO₂ layer, approximately 0.5 mm below the epoxy surface.

3.2 Operation of the microMOSFET System

The microMOSFET AutoSense system is comprised of standard sensitivity microMOSFETs (TN-RDM-502), reader (TN-RD-15), dual bias supply (TN-RD-22), computer (Dell Dimension 2400), and AutoSense PC software (TN-RD-49). All components except the computer were manufactured by Thomsom and Nielsen, Ottawa, Canada.

The microMOSFETs detector is mounted on a laminated polyamide cable under 0.6 mm of epoxy to protect against moisture and degradation. The cable is attached to the dual bias supply which is coupled to the reader. The manufacturer recommends attaching the detectors to the bias for one hour prior to irradiation. The dual bias supply in standard mode applies positive gate voltages of +15 V and +1 V during irradiation and intermediate periods. After irradiation the MOSFETs should sit for one minute to allow time for the holes to reach the Si/SiO₂ interface. The user can then measure the difference in threshold voltage through the AutoSense PC software. During the reading phase, an increasing negative voltage is applied to each MOSFET until a few μ A of current between the drain and source is established. Once this occurs, the corresponding gate voltage is designated as the threshold voltage. The user should not take consecutive readings within two minutes of each other because the charge injected by the measuring

system may cause a transient perturbation and the detector may over-respond. MicroMOSFETs have a lifetime of 18,000 mV, after which holes begin to saturate in the oxide layer. The detector may lose up to 5% sensitivity if irradiated after this point [28].

3.3 MOSFETs versus other point dosimeters

Thermoluminescence dosimeters (TLDs) and diodes were traditionally the most common clinical point dosimeters. TLDs are often made of doped lithium fluoride which contains traps in the band gap between the valence and conduction bands. They have an effective atomic number of 8.14 and are considered tissue-equivalent. Ionizing radiation causes excited electrons in the valence band to jump into the conduction band. The excess kinetic energy is quickly released as fluorescent radiation and most electrons return to the valence band, while a few move to the energy traps in the band gap. The total number of electrons caught in the energy traps is proportional to absorbed dose. If the TLD is later heated, electrons sitting in the energy traps can jump back to the conduction band and then return to the valence band. Each electron releases a photon as it moves to the valence band and the resulting light intensity is proportional to absorbed dose. TLDs do not require a power supply and operate without cables. However, they require pre- and post-processing at high temperatures and over long hours. After the signal is read, all information is erased. Studies have shown TLDs to have an uncertainty of 2% when handled with extreme care and typically 5% for daily clinical handling [28, 29].

Diodes operate in a similar manner to MOSFETs. They are semiconductors, commonly made from p-type silicon with an n-type surface layer. Electromagnetic forces at the interface between the two materials push the majority holes in the p-type and the

majority electrons in the n-type back, creating a depletion region with no free charge carriers. The p-type side of the junction has a net negative charge while the n-type side has a net positive charge. When the diode is exposed to ionizing radiation, electron-hole pairs are created. Minority carriers from each layer move to the depletion region. An external field attracts the carriers across the depletion region, creating a current whose magnitude is proportional to absorbed dose. Diodes require wires, and the signal readout is immediate. Diode signals may vary up to 2% per 5°C [30]. Their reproducibility is 0.5% as quoted by the manufacturer [31].

While TLDs are wireless and require less setup time they are unable to deliver information immediately. They require long processing times and extreme care to achieve low uncertainty. They also lose all information following the reading phase. Diodes require more setup time but are able to give instant readouts. Once the detector is out of the radiation field and the current stops, the diode loses all dosimetric information. MOSFETs can deliver information within one minute or after irradiation. They can also be read many times with minimal loss of information. MicroMOSFETs generally have the smallest active area and total volume of the 3 devices. A typical microMOSFET will have an active area of 0.04 mm^2 with a total volume of $1.0 \times 1.0 \times 3.5 \text{ mm}^3$ [32]. A typical TLD chip is $3.0 \times 3.0 \times 0.9 \text{ mm}^3$ with an active area of 9 mm^2 and a typical diode has a volume of $3.8 \times 5.5 \times 38 \text{ mm}^3$ with an active volume of 0.28 mm^2 [33]. TLDs can obtain a smaller active area if they are in powder form or sectioned into micro pieces.

3.4 MicroMOSFET Characterization

Thomson and Nielsen quote an uncertainty of 2% at two standard deviations for the TN-RDM-502 microMOSFETs, and claim the detectors are independent of energy, orientation, and temperature. Other studies have shown an inter-variability of 5% and a reproducibility of $\pm 2\%$ at two standard deviations, with no dose-rate, temperature, energy, or directional dependence within the calculated uncertainty [26]. This study characterizes the proportionality between absorbed dose and threshold voltage change, as well as detector inter-variability, energy dependence, and orientation effects.

3.4.1 Calibration

MOSFET detectors can be used to measure absolute dose once the proportionality between absorbed dose and threshold voltage is determined. To measure the calibration coefficient MOSFETs were irradiated in a solid water phantom to a known dose, verified by an ionization chamber. A single MOSFET was placed at 5 cm depth with 12.3 cm of solid water backscatter, using an SSD setup (see figure 3.2). The phantom had specially designed grooves to ensure no weight was placed on the detector and no air gap was present in the phantom around the detector. The MOSFETs were held in place with low-adhesion tape. A Farmer chamber was placed at 11.3 cm depth to measure absorbed dose. Dose was calculated according to the procedures describe in AAPM TG-51 [34] and scaled to the MOSFET depth using known percentage depth dose (PDD) values. The ionization chamber has a reproducibility of 0.7% [47]. The detectors were connected to the bias supply for one hour before calibration and read one minute after each irradiation, according to the manufacturer's recommendations. Each calibration coefficient was determined by irradiating a single MOSFET three times for a specific beam energy and

orientation. The calibration field size was $10 \times 10 \text{ cm}^2$ and doses ranged from 50 to 200 cGy at the depth of the detector.

To test the detectors for energy dependence, calibration coefficients were determined for a cobalt-60, 6 MV, and 10 MV using a Theratron 780 unit and Elekta Synergy linear accelerator. The source to surface distance (SSD) was 80 cm in the cobalt-60 beam and 100 cm for all other energies.

To test angular dependence, MOSFETs were irradiated with the epoxy facing up (face-up orientation) and the epoxy facing down (face-down orientation).

Figure 3.2 MOSFET calibration setup with the MOSFET and Farmer chamber at 5 and 11.3 cm depth, respectively.

3.5 Results

The manufacturer quotes a calibration coefficient of 1.00 cGy/mV in a cobalt-60 beam for standard sensitivity microMOSFETs with a standard bias. The calculated average calibration coefficient for 25 MOSFETs face-up in a cobalt-60 was (0.90 ± 0.03) cGy/mV at one standard deviation, calculated over three measurements. Figure 3.3 shows a histogram of the detector's inter-variability. The histogram bin size is 0.009 cGy/mV. A Gaussian distribution is overlaid with the same mean and standard deviation as the calibration histogram. It is scaled according to the bin width and sample size such that each plot has the same integrated area:

$$f(CF) = \left(\frac{Nw}{\sigma\sqrt{2\pi}} \right) e^{-\frac{(CF-\mu)^2}{2\sigma^2}} \quad [3.9]$$

where CF is the calibration coefficient, N is the number of data points, w is the bin width, σ is the standard deviation, and μ is the mean. The distribution of calibration coefficients follows a Gaussian distribution according to a two-tailed p-value of 0.7887. The differences between calibration coefficients of different detectors are most likely due to slight statistical variations in the thickness of the oxide layer, epoxy, or encapsulation which arise during manufacturing.

The average calibration coefficients for 10 MOSFETs face-up in 6 and 10 MV beams was (0.95 ± 0.02) cGy/mV and (0.97 ± 0.03) cGy/mV, respectively, calculated for three measurements (see table 3.1). These values are lower than the manufacturer's quote of 1.0 cGy/mV and show a slight increase with increasing energy. Figure 3.4 illustrates individual detector's variation with energy, normalized to the calculated calibration coefficient in the cobalt-60 beam. Seven of the ten detectors show significant energy dependence at the 90% confidence level according to a standard student-t test. As a

group, the average calibration factors for the MOSFETs listed in table 3.1 also show significant energy dependence at the 90% confidence level according to a student-t test.

Calibration was repeated for 25 MOSFETs in a cobalt-60 beam with a face-down orientation. The average calibration coefficient was (0.95 ± 0.02) cGy/MV for the group, showing a significant increase over the face-up orientation according to a student-t test at a 90% confidence interval. Figure 3.5 shows the ratio of face-up and face-down calibration coefficients for individual MOSFETs. Only one detector in the group had a face-up calibration coefficient greater than its face-down value.

According to these results, microMOSFETs are considered suitable point dosimeters for this project. The treatment plans in chapter 5 will be verified by the MOSFET with the lowest uncertainty in the specific beam energy. The uncertainties of the selected MOSFETs in a cobalt-60, 6 MV, and 10 MV are 1.5, 1.8, and 2.0%, respectively, at two standard deviations. MicroMOSFETs will be oriented face-up for all irradiations, and each response will be used to calculate absolute dose using energy specific calibration coefficient.

Table 3.1 Calibration coefficients for TN-RDM-502 microMOSFETs.

Beam	Data Points	Orientation	CF (cGy/mV)	$\pm 1 \sigma$ (cGy/mV)
cobalt-60	25	Face-up	0.90	0.03
cobalt-60	25	Face-down	0.95	0.02
6	10	Face-up	0.96	0.03
10	10	Face-up	0.97	0.02

Figure 3.3 MOSFET inter-variability for 25 MOSFETs face-up in a cobalt-60 beam.

Figure 3.4 Relative calibration coefficients for 10 MOSFETs, face up, in cobalt-60, 6 MV and 10 MV beams. Error bars represent one standard deviation.

Figure 3.5 Relative calibration coefficients for 25 MOSFETs in a cobalt-60 beam. The vertical axis represents the face-up calibration coefficient scaled by the face-down calibration factor for the same MOSFET.

Chapter 4

Radiochromic Film: Two-dimensional Dosimeter

A radiochromic process is the immediate and permanent colouration of a material when exposed to ionizing radiation. The technology has evolved from early self-processing prints including daguerreotype, collotype, photogravure, and photolithographic. In 1826, one of the first known radiochromic processes showed unsaturated hydrocarbon polymers cross-link when exposed to ionizing radiation, resulting in a light-scattering pattern [35]. Other radiochromic media include organic *cis-trans* isomers where the orientation of the functional group changes when exposed to ionizing radiation. More recently, thin radiochromic films have been used to measure high doses (10^4 - 10^6 Gy). These films are generally composed of nylon or vinyl and triphenylmethane leucocyanides which form a coloured dye when exposed to ionizing radiation. Colourless aqueous tetrazolium chloride salts reduce to highly coloured solids in the presence of radiation and have been useful in radiographic imaging. In radiation therapy, radiochromic films with polydiacetylene have been introduced. They turn a deep blue in the presence of ionizing radiation.

Polydiacetylene films are composed of thin, active microcrystalline monomers on a polyester film base. Ionizing radiation causes homogeneous, planar monomers to join along the carbon-chain backbone creating long polymer molecules. The reaction does not require chemical, thermal, or optical treatment. Polymerization of the active layer is a function of deposited energy and measuring the degree of colour change can be correlated to total delivered dose.

4.1 GafChromic Film

GafChromic dosimetry film is the most widely used commercial radiochromic product [36] and is produced by ISP Technology. The original model was developed by David Lewis in 1986 for electron lithography [37]. Since then several models have been developed with various structures and increasing sensitivity.

HD-810: HD-810 (also known as DM-1260) was the original commercial dosimetry GafChromic film. It was sold in 15 m rolls but is currently available in standard sheets. A 7 μm active layer is coated on a 100 μm polyester base and the film has a useful dose range of 10 to 2500 Gy. The thin active layer gives good positional accuracy however causes a relatively insensitive response compared to other models.

MD-55-2: MD-55-2 has two 16 μm active layers separated by 25 μm of adhesive and polyester. The active layers are each protected by 67 μm of polyester and the film can be emerged in water. Due to the thicker active layer this design is more sensitive than the HD-810 film but has a useful range of only 3 to 100 Gy. Positional accuracy is difficult to obtain with this model because the structure is more complex and the adhesive layer may shrink or expand.

HS: This model has a single 40 μm active layer between two 97 μm layers of polyester. There is no central adhesive layer giving greater positional accuracy compared to the MD-55-2 model. The increased thickness allows for greater sensitivity at the cost of the useful range (1 to 50 Gy). The film retained its ability to be used in water. The active layer consists of 9% hydrogen, 57% carbon, 18% oxygen, and 16% nitrogen and has an effective atomic number of 6.0 to 6.5, similar to all previous models. The HS

absorption spectrum in figure 4.1a shows a large peak at 678 nm. This result is similar for all previous models [37].

EBT: External beam therapy (EBT) film is the most recent model of GafChromic film. It is ten times more sensitive than the HS model [38] and is designed for use in high energy beams with varying field sizes. The film consists of two 17 μm active layers, separated by a 6 μm surface layer. The active layers are covered by 97 μm polyester sheets (see figure 4.1b), a similar structure to the MD-55-2 model. The active range is 0.01 to 8 Gy. The film has a different chemical makeup from the previous models consisting of 39.7% hydrogen, 42.3% carbon, 16.2% oxygen, 1.1% nitrogen, 0.3% lithium, and 0.3% chlorine. Its effective atomic number is 6.98, comparable to that of water at 7.5. Figure 4.1a shows the EBT absorption spectrum which has a similar shape to the previous models but is shifted such that the peak is at 633 nm due to the new chemical composition.

Figure 4.1 GafChromic Film (a) absorption spectra for the HD-810, MD-55, HS, and EBT models. Image taken from Devic et al 2007 [39]; (b) EBT film structure. Image taken from Devic et al 2005 [38]

4.2 Radiochromic Film as an Absolute Dosimeter

Optical density (OD) is a measure of light attenuation through a transparent object. It has a logarithmic relationship to light intensity and is calculated as follows:

$$OD = \log_{10} \left(\frac{I_o}{I} \right) \quad [4.1]$$

where 'I_o' is the incoming light intensity and 'I' is the transmitted light intensity. To measure colouration of film, the change in optical density before and after irradiation is of interest. This is known as the net optical density (netOD). If light transmission is being measured with a flatbed scanner, an additional background term (I_{bckg}) should be subtracted from the transmission value to account for the zero-light transmitted intensity value according to the following equation:

$$netOD = \log_{10} \left(\frac{I_{unexp} - I_{bckg}}{I_{exp} - I_{bckg}} \right) \quad [4.2]$$

Where I_{unexp} and I_{exp} are the unexposed and exposed film transmission values. The background term can be obtained by placing an opaque object in the scanner and measuring the intensity value. Ideally this term is zero but can be up to a few percent of the incoming light intensity value.

Because polymerization is a function of deposited energy and leads to colouration in radiochromic film, the change in optical density before and after irradiation (net optical density, netOD) is indicative of delivered dose. The light intensity term I_o/I has an exponential relationship with dose, making equation 4.1 linear with dose. However, because of the additional background terms and slight deviations from linearity at high doses it has been shown that a polynomial relationship between net optical density and

dose is most suitable [38]. Equation 4.3 represents the function which minimizes parameter uncertainty while intersecting the origin:

$$D = b \cdot netOD + c \cdot netOD^n \quad [4.3]$$

where D is the delivered dose and b, c, and n are calibration parameters. The first term represents the linearity response at low doses and the second term was introduced to account for the high dose response. The parameters b, c, and n can be determined using the Levenberg-Marquardt quasi-Newton minimization technique.

4.3 Characterization of GafChromic EBT Film

Net optical density has a sensitive relationship with dose and can depend on many factors. Colour stability is a function of postirradiation time. MD-55-2 and HS radiochromic films show a 13% and 19% increase in post-irradiation optical density over the first 12 hours. EBT film is slightly less time dependent and has been shown to stabilize to within 1% after the first 6 hours postirradiation [40]. All models of GafChromic films show some energy dependence at low X-ray energies. The films tend to under-respond in this range due to their lower effective atomic number and high carbon fraction relative to water and tissue [35]. In the external beam therapy range (1-10 MV) EBT films have been shown to have no significant energy dependence [41]. EBT films also have no significant dose-rate or fractionation dependence [35, 42, 43]. The angle of the incident beam with respect to the film may have associated problems if the film axis is placed parallel to the beam path due to varying scatter effects. This problem is minimized in EBT films compared to earlier models because of its higher effective atomic number of 6.98. Temperature and humidity can also affect film response. Films were shown to vary by 2.5% per 10°C [36]. Humidity is less significant

with only a $\pm 2\%$ variation over the relative range of 6-94% [35]. Films also have a slight sensitivity to UV light. EBT films show no significant variation on a single sheet or within a single batch. However, they may have inter-batch dependence due to differences in the active layer thickness which can arise when films are manufactured at different times [44]. Also, light transmission may depend on the angle of the film sheet with the incident readout light source. Monomers tend to align in one direction during manufacturing and can lead to a phase shifting effect on polarized light or interference patterns on coherent light [38].

4.4 Handling of Radiochromic Film

Several authors have discussed proper handling and care of radiochromic film [35, 36, 37, 38, 45]. Proper technique must be used to minimize undesirable effects caused by the sensitive relationship between optical density and the various factors discussed above.

Surgical gloves should be used when handling film to avoid any oil transfer. Individual film sheets should be separated by tissue paper to absorb any moisture and minimize scratches. Films should be stored in a dry and dark place and the temperature should remain between 20 to 25°C. If it is necessary to cut the film sheet, a sharp blade or guillotine knife should be used. This will ensure individual layers of the film do not separate. A small dot with a permanent marker should be immediately placed in the corner of the film to mark orientation (location of the marking should be consistent for all films). If a sheet is being divided into smaller sections, orientation marks for each division should be placed prior to cutting. During the irradiation process, films should be covered whenever possible to avoid excess exposure to UV light. After irradiation the

films should be left for a minimum period of 24 hours to allow the colour to stabilize. Films should be scanned at normal room temperature and humidity. The scanner should use incoherent, non-polarized light such as a flatbed CCD scanner. A laser should never be used. When analyzing films only the red channel should be considered because of the films peak sensitivity at 633 nm. [35, 36, 37, 38, 45]

4.5 Radiochromic versus Radiographic Film

Silver halide radiographic film is the traditional clinical 2-D dosimeter. Radiographic film may show several adverse characteristics because it is not tissue equivalent and its scattering effects are different from tissue and water. The film is known to have significant energy dependence due to its silver content ($Z_{Ag} = 47$). Elements with high atomic numbers are much more likely to cause photoelectric or Compton interactions when exposed to low energy photons compared to low atomic numbered elements. Radiographic films may also over-respond with increasing field size and depth [43]. With increasing field size, more scatter occurs and more low energy photons are present. This is also true for increasing depths.

Radiochromic film is self-processing while radiographic film requires a dark room and chemical processing. Because radiographic film is highly sensitive to visible and UV light it must remain covered in an envelope at all times. This can lead to uncertainty of the film's position. Also, radiographic film cannot be used in water. Processing may lead to additional uncertainty because the spectral response is strongly affected by chemical concentration, time, and temperature [46]. Further, radiochromic film has an increased sensitivity relative to radiographic film when scanned in the visible light spectrum. Figure 4.2 shows the optical density response of HS GafChromic film

and EDR-2 radiographic film when irradiated to 1 Gy. The GafChromic film has a higher response for all wavelengths and is 4 times more sensitive at its peak wavelength.

The advantages of radiographic film are that it is relatively inexpensive compared to radiochromic film, the processed image is stable, and optical density is not affected by polarized or coherent light. However, because radiographic film requires chemical processing and shielding from visible light, radiochromic film is the preferred 2-D dosimeter for this work.

Figure 4.2 Optical density response of EDR-2 radiographic and HS radiochromic film over the visible light spectrum when irradiated to 1 Gy. Image taken from Butson et al. 2003. [36].

4.6 EBT GafChromic Film Calibration

All films were handled in accordance with section 4.4. Scanning was done using an Epson Perfection V700 photo flatbed scanner with Epson Silverfast software. Colour and spatial resolution were set to 48 bit and 300 DPI, respectively. Images were saved as red-green-blue (RGB) tiff files. Analysis was done with MATLAB version 7.4.0 software using only the red image channel with 16 bits or 65536 shades per colour.

A single sheet of EBT GafChromic radiochromic film was removed from its envelope and marked for orientation. The film was centred in the scanner with the orientation dot in the upper left corner. The sheet was then divided into 1.5 cm x 2.5 cm blocks with orientation marks placed in the top left corner of each and cut using a paper guillotine. The original tissue paper was placed back on the film before it was cut and the film was shielded from light when possible. Each sheet division was given a serial number in the bottom right corner using a fine point permanent marker. The film divisions were placed in small envelopes and separated by the cut pieces of tissue.

Calibration was done with a 6 MV beam at 100 cm SSD in a solid water phantom. The field was 10x10 cm² and film pieces were placed at 5 cm depth with 15 cm backscatter. A Farmer chamber was placed at 11 cm depth to measure dose delivered to the film. Temperature and pressure were recorded. Three pieces of film were not irradiated but left with the irradiated samples as controls. The remaining pieces were irradiated individually from 50 to 600 cGy. A total of 46 divisions were irradiated. Films were given 48 hours to stabilize before scanning. Films were individually placed at the centre of the scanner with the orientation mark in the top left corner and scanned with the same resolution as given above. For consistency, films with the orientation mark in

the top left corner of the scanner are considered to have zero rotation relative to the incident readout light source.

The MATLAB program opened individual image files and located the centre of the film. The mean intensity value (I) of the red channel over a 100 pixel x 100 pixel region was calculated (approximately $7.5 \times 7.5 \text{ mm}^2$). Net optical density was calculated using equation 4.2 with I_{unexp} as the mean intensity of the unirradiated samples and I_{bckg} as the zero-transmission value. Dose values measured with a Farmer chamber were calculated according to the procedures in AAPM TG-51 [34] with corrections applied for temperature and pressure. The difference in depth between the Farmer chamber and film was corrected using known percentage depth dose values. The ionization chamber has a reproducibility of 0.7% [47]. A Levenberg-Marquardt quasi-Newton minimization technique was applied to equation 4.3 to solve for parameters b , c , and n . This process was repeated in a cobalt-60 and 10 MV beam using films from a single batch. A second 10 MV calibration was done using films from a different batch.

Positional dependence in the scanner was determined by moving film samples horizontally across the scanner (see figure 4.3a). Percentage depth-dose (PDD) curves in solid water were measured using the film both parallel and perpendicular to the beam axis in a 6 MV beam with as $10 \times 10 \text{ cm}^2$ field. For the parallel PDD case, a single whole sheet of film was placed in between two 15 cm blocks of solid water. The gantry was rotated to 90° and the phantom was adjusted to 100 cm SSD (see figure 4.3b). The film was irradiated with 200 cGy at 5 cm depth. For the perpendicular PDD case, $1.5 \times 2.5 \text{ cm}^2$ divisions of films were irradiated individually at varying depths in the solid water phantom. The gantry was set at 0° and the phantom was placed at 100 SSD (see figure

4.3b). Each film was irradiated with 200 MU which corresponds to a dose of 200 cGy at 5 cm depth.

Film stability was measured by irradiating pieces of films according to the calibration procedures and scanning them at various post-irradiation times. Films were irradiated with 50 to 500 cGy and scanned at 1 to 96 hours after irradiation. The dependence of light transmission on sheet angle to the incident readout light source was observed by scanning film samples with the orientation mark at 0° and 90° degrees from the original calibration position. The orientation mark was firstly placed in the top left corner (as per calibration) and the film was scanned. The film was then rotated 90° and the scan was repeated.

a)

Figure 4.3 GafChromic film position for (a) film position in the scanner for positional dependence experiment, (b) parallel and perpendicular percentage depth dose experiments.

4.6 Results

GafChromic EBT film has been shown to have an overall dose measurement uncertainty of 2% at one standard deviation for doses greater than 0.4 Gy [38]. The calibration results for a 6 MV beam with a 10x10 cm² field in a solid water phantom according to equation 4.3 are:

$$D_6 = (332.9) \cdot netOD + (564.4) \cdot (netOD)^{3.2} \quad [4.4]$$

The squared normal of the residual for 46 measured data points was 556.6 Gy² as calculated by equation 4.5:

$$residual = \sum_i (D_i - b \cdot netOD_i - c \cdot netOD_i^n)^2 \quad [4.5]$$

Where D_i and $netOD_i$ are the i^{th} measured dose and net optical density. The calibration parameters had a χ^2 value of 15.5 as calculated by equation 4.6:

$$\chi^2 = \sum_i \left(\frac{D_i - b \cdot netOD_i - c \cdot netOD_i^n}{\sigma_i} \right)^2 \quad [4.6]$$

where σ_i is the standard deviation of the i^{th} dose calculation:

$$\sigma_{Dose} = (b + n \cdot c \cdot netOD^{n-1}) \sigma_{netOD} \quad [4.7]$$

and σ_{netOD} is the standard deviation among pixels of the net optical densities over the measured region of interest (100 pixels by 100 pixels). The *goodness of fit* of the calibration curve passes a 99% confidence level for the 46 measured points and 3 parameters as per a standard χ^2 probability table.

Results from the cobalt-60, 6 MV, and 10 MV calibrations are shown in figure 4.4a. The error bars shown on the plot represent pixel variation at one standard deviation as calculated by equation 4.7.

Figure 4.4 GafChromic Calibration Curves in a $10 \times 10 \text{ cm}^2$ field with films at 5 cm depth for (a) a single batch with cobalt-60, 6 MV, and 10 MV beams, (b) 2 different batches in a 10 MV beam.

The results from figure 4.4a show no dependence on energy within the measurement uncertainty for doses 50 to 500 cGy in cobalt-60, 6 MV, and 10 MV beams. Figure 4.4b shows the calibration results for films from 2 different batches in a 10 MV beam. The calibration curves do not show any significant batch dependence within the measurement uncertainty. The uncertainty for dose values calculated using equation 4.3 with no dependence on energy is 1.6% at two standard deviations.

Figure 4.5a shows the variation of a single film when moved across the scanner. The horizontal axis is the distance across the scanner in cm from left to right. The error bars represent one standard deviation of the measured pixel value over a 100x100 region of interest. Each horizontal line represents the measured pixel value at the centre of the scanner (10 cm) where the calibration films were imaged. Each film shows a significant dependence on position with a peak pixel value at 7 cm. A correction factor was applied using a third order polynomial as a function of position and parameters were determined with a least-squared minimization technique. Figure 4.5b shows the pixel values after applying the correction factor. All the high dose values (300 and 400 Gy) and the majority low dose (100 Gy) fall within the measured uncertainty.

Figure 4.5 Pixel value of film samples versus horizontal position in the scanner when (a) uncorrected, (b) corrected pixel with a third order polynomial.

Percentage depth dose curves taken in a solid water phantom at 100 SSD in a 6 MV beam with a $10 \times 10 \text{ cm}^2$ field are shown in figure 4.6. Error bars on the perpendicular film samples represent the measurement uncertainty at one standard deviation. The Markus chamber has an effective depth of measurement 0.03 mm from the chamber surface and the perpendicular films samples have an effective depth of measurement at 0.12 mm. The parallel film was shifted 0.25 mm to align with the Markus chamber data, and uncertainty in the zero depth position of the parallel film is taken as $\pm 0.25 \text{ mm}$ accordingly.

Figure 4.6b shows the build-up region of the percentage depth dose curve. The Markus chamber data is extrapolated to the surface using a second order polynomial. The Markus chamber, parallel film, and perpendicular film measure surface doses of 22.5, 23.1, and 20.7%, respectively. According to these results, the GafChromic film agrees with the Markus chamber measurements within the uncertainty of measurement.

Figure 4.6 Percentage depth dose curves in a solid water phantom with a 6 MV beam and $10 \times 10 \text{ cm}^2$ field at 100 cm SSD. (b) Build-up region with a second order polynomial extrapolation of the Markus chamber data.

Results from the film stability are shown in figure 4.7 for dose levels from 50 to 460 cGy. All dose levels had a similar logarithmic time response with stabilization occurring after 24 hours. The average change in net optical density from 1 to 24 hours post-irradiation was 15%. After 24 hours there was no significant change within the measured uncertainty.

Figure 4.8 plots the pixel value with the film rotated 90° relative to the pixel values with no rotation. Error bars represent the measured uncertainty at one standard deviation. The plot shows a significant increase in pixel value when the film sheets are rotated 90° from the calibration position. The increase in pixel value appears to have a positive correlation with increasing dose levels. At 100 cGy the rotated film measured a 13% increase in pixel value compared to the sheet with no rotation, and at 300 cGy a 18% increase was measured.

According to these results, the GafChromic EBT film is considered to be a suitable dosimeter when irradiating with the beam's central axis parallel with the film sheet. Films will be scanned 48 hours after irradiation and with the same sheet orientation as in calibration. Pixel values will be corrected for position in the scanner as in figure 4.5b.

Figure 4.7 Post irradiation time colouration of EBT GafChromic film

4.8 Relative pixel value of GafChromic Film at 90° and 0°.

Chapter 5

Treatment Plan Verification

Two neck phantoms were fabricated for this project. Each phantom contained 13 acrylic cylindrical disks mounted on an acrylic stand. The disks were 2 cm thick and 13 cm in diameter. A plastic screw at the end of the stand could adjust pressure on the disks as desired. The first phantom was homogeneous and contained solid acrylic disks (figure 5.1a). The second phantom was heterogeneous and contained acrylic disks with a central air cavity (figure 5.1b). The air cavity was cylindrical with a 2.5 cm diameter and represented the trachea. The acrylic represented the surrounding muscle and tissue in the larynx region of the neck. Acrylic is considered to be an acceptable substitute for human tissue because the equivalent photon pathlength in acrylic is within 2% of that in muscle or lung, (see equation 2.11 and figure 2.3b) and electron range in acrylic is within 0.5% for the energies of interest in this project.

The phantoms were designed to hold MOSFETs and GafChromic film. The heterogeneous phantom MOSFET dose measurement points were at the two inner air-acrylic interfaces (52.5 and 77.5 mm below the surface) along the central beam axis (known as dose measurement points one and two, respectively, see figure 5.2). The homogeneous phantom holds MOSFETs at the two dose measurement points by using the heterogeneous disks with an acrylic insert in place of the solid disks. The insert was cylindrical with a 2.5 cm diameter and fits tightly into the disks. It has two small grooves at opposite ends to fit the MOSFETs and cable. GafChromic film can be mounted in between disks. The stand has thin slots at the intersection points between disks to hold the film (see figure 5.1d).

Figure 5.1 Acrylic Neck Phantom: (a) homogeneous disk, (b) heterogeneous disk (c) heterogeneous phantom (d) heterogeneous phantom with GafChromic film.

Figure 5.2 MOSFET dose measurement points along the central beam axis

5.1.1 CT Image Acquisition

CT images were acquired for both the heterogeneous and homogenous phantoms using a Philips Brilliance 16-slice large-bore CT scanner. The scanner calculates linear attenuation coefficients (μ) by irradiating the phantom with a kV photon beam and measuring beam attenuation. The CT generates 16 slices (0.75 mm per slice) per gantry rotation. A specific voxel's linear attenuation coefficient (μ_i) is calculated by filtered back projecting the attenuation information acquired over a range of angles. The CT scanner outputs a 3D image where each voxel is represented by the Hounsfield unit (HU):

$$HU_i = \frac{\mu_{x,i} - \mu_{H_2O}}{\mu_{air} - \mu_{H_2O}} \times 1000 \quad [5.1]$$

where $\mu_{x,i}$ is the linear mass attenuation coefficient of the i^{th} voxel, and μ_{air} and μ_{H_2O} are the linear mass attenuation coefficients of air and water. The CT image had $2 \times 2 \times 2 \text{ mm}^3$ voxels. The 3D HU matrix was transferred to a contouring station (Focus, CMS) where the phantom surface was defined on each CT slice. The matrix was then transferred to an Elekta Xio and a CMS PrecisePlan treatment planning system. The treatment planning systems converted HU to electron density independently. However, the Xio and PrecisePlan systems used identical conversion tables and calculated identical electron density matrices.

5.2 Treatment Plans

Single beam treatment plans were created using various algorithms to calculate dose delivered to both the heterogeneous and homogenous phantoms. All plans used the same contoured electron density matrix and therapy machines, and were delivered in a single fraction. The same therapy machine was used for each plan and subsequent irradiation since beams from different linacs may have different energy spectrums and

outputs, even if the machines have the same nominal energy. Also, an isotope machine, such as a cobalt-60 unit, will deliver radiation at varying dose-rates depending upon the strength of the source. Cobalt-60 plans were generated for a Theratron 780 unit. The 6 and 10 MV plans were generated for an Elekta Synergy linear accelerator. The beam axis was centred with the cylindrical phantom axis for all fields with the gantry at 90° (see figure 5.2). Table 5.1 shows which algorithms were used to create treatment plans in cobalt-60, 6 MV, and 10 MV. The PrecisePlan XVMC algorithm was only capable of calculating dose for the 6 MV beam because the system had not yet been commissioned with data from the other beams. Similarly, the Xio Clarkson integration algorithm was not capable of calculating dose for the cobalt-60 beam. All possible plans were generated. Table 5.2 shows which fields were used. The cobalt-60 unit was physically unable to close its collimator to a $4 \times 4 \text{ cm}^2$ field. Otherwise, plans were generated for cobalt-60, 6 MV, and 10 MV with field sizes 4×4 , 5×5 , 6×6 , 8×8 , and $10 \times 10 \text{ cm}^2$. A total of 114 plans were created.

Cobalt-60, 6 MV, and 10 MV plans used an SSD 80, 100, and 100 cm, respectively. This was consistent with the MOSFET and GafChromic film calibrations. Once a plan was generated, a 2D matrix containing absolute dose values was output and converted to a Matlab file. The matrices were in the centre of the beam and in transverse view. The Xio system used a calculation grid of $1 \times 1 \times 1 \text{ mm}^3$ and output matrices with $1 \times 1 \text{ mm}^2$ dose points. The PrecisePlan system used a calculation grid of $2 \times 2 \times 2 \text{ mm}^3$ and output matrices with $2 \times 2 \text{ mm}^2$ dose points. . The approximate energy fluence for the 6 and 10 MV beams are shown in figure 5.3. The cobalt-60 energy fluence is equally divided by photons at 1.17 MeV and 1.33 MeV.

Table 5.1 – Algorithms used in Treatment Plans

Beam	Algorithm				
	Clarkson	FFTC	Superposition	Fast Sup.	XVMC
Cobalt-60		✓	✓	✓	
6 MV	✓	✓	✓	✓	✓
10 MV	✓	✓	✓	✓	

Table 5.2 – Fields used in Treatment Plans

Beam	Field /cm ²				
	4x4	5x5	6x6	8x8	10x10
Cobalt-60		✓	✓	✓	✓
6 MV	✓	✓	✓	✓	✓
10 MV	✓	✓	✓	✓	✓

Figure 5.3 Approximate energy fluence for an Elekta Synergy (a) 6 MV and (b) 10 MV beam

5.3 Method

The treatment plans discussed in section 5.2 were verified with MOSFET point dosimeters and GafChromic 2D dosimeters. For each irradiation, the central axis of the phantom was normal to and intersected the beam's central axis (as shown in figure 5.2). The MOSFETs and GafChromic film were irradiated in separate experiments because the MOSFETs were held in the centre of a disk while film was placed in between disks, and the dosimeters could not simultaneously be aligned with the central beam axis

5.3.1 MOSFET

MOSFETs were attached to the bias supply for one hour before irradiation. They were placed in the MOSFET grooved disks and held with low-adhesion tape. The MOSFET disk was then placed in the middle of the phantom with 6 other disks already to one side. The remaining 6 disks were placed on the other side after the MOSFET cables had been strung through. The MOSFETs were aligned with the beam axis using the machine lasers. Pressure was applied to the disks to ensure no air entered between the disks. The SSD was set to 100 cm for the 6 and 10 MV beams and 80 cm for the cobalt-60 beam. Throughout irradiation the MOSFET cables were attached to the bias supply. Each MOSFET was irradiated 3 times for a given field with one minute passing between irradiation and reading.

The MOSFET's response was converted to dose using the energy specific calibration coefficients presented in chapter 3 according to:

$$D = \Delta V_{th} \cdot CF(E) \quad [5.2]$$

where D is the absolute dose, ΔV_{th} is the change in threshold voltage, and $CF(E)$ is the calibration coefficient for beam energy E . The uncertainty on dose (δD) was calculated

using both the standard deviation of the change in threshold voltage (δV_{th}) over three measurements and the calculated uncertainty on the calibration coefficient (δCF):

$$\delta D = \sqrt{(\Delta V_{th} \cdot \delta(CF))^2 + (CF \cdot \delta(\Delta V_{th}))^2} \quad [5.3]$$

5.3.1 GafChromic Film

Film sheets were marked for orientation and cut in half using a guillotine knife. One half sheet of the film was placed between two loosened central disks in the phantom. The film edge was aligned with the surface of the phantom. Pressure was applied to the disks and the phantom was adjusted to the proper SSD. The film was aligned parallel to the central beam axis using the machine lasers. Films were stored in a dark folder for 48 hours before being scanned. Scanning colour resolution was set to 48 bit RGB colour with a 300 DPI spatial resolution and the images were saved as tiff files (consistent with calibration). The image files were opened with a Matlab program and converted to absolute dose using the calibration parameters discussed in section 4.7 and equation 4.2 and 4.3. The pixel values were adjusted with the calculated scanner spatial dependence polynomial (see section 4.7 and figure 4.5).

5.4 Results

Depth dose profiles were taken along the central beam axis in the transverse plane. The GafChromic film dose values were averaged over 1 mm to coincide with the Xio treatment planning pixel size. MOSFETs were placed at both inner air-acrylic interfaces (52.5 mm and 77.5 mm depth). The point of measurement was shifted by the effective depth of measurement (0.5 mm).

Figures 5.4 through 5.6 show the results when the homogeneous phantom was irradiated according to the plans of section 5.2 with 5x5 and 10x10 cm² fields. MOSFET and film measured doses agree at both points for all energies and fields. There is no significant difference between the FFT, superposition, multi grid superposition calculated dose and the dose measured with the GafChromic film beyond the surface for all beam energies and all fields. The Clarkson integration algorithm agrees with measured dose past the point of maximum dose but estimates a larger dose than measured in the build-up region. This difference is more pronounced in 10x10 cm² fields and the 6 MV beam.

Figure 5.7 shows the variation in dose with depth in a solid water phantom with no surface curvature. Measurements and calculation were done in a 6 MV beam with a 10x10 cm² field at 100 cm SSD. The Clarkson integration algorithm predicts a surface dose consistent with the superposition algorithm. However, both algorithms predict a dose larger than the measurement data. The Markus chamber, perpendicular film, and parallel film predict a surface percentage depth dose of 22.4, 20.7, and 23.1%, respectively. The Clarkson integration algorithm and superposition algorithm predict a surface dose of 40.8 and 36.8% respectively.

Figure 5.4 Variation in dose with depth along the central beam axis in the homogeneous phantom when irradiated with a cobalt-60 (a) $5 \times 5 \text{ cm}^2$ field, (b) $10 \times 10 \text{ cm}^2$ field. Film was parallel with the central beam axis

Figure 5.5 Variation in dose with depth along the central beam axis in the homogeneous phantom when irradiated with a 6 MV (a) $5 \times 5 \text{ cm}^2$ field, (b) $10 \times 10 \text{ cm}^2$ field. Film was parallel with the central beam axis

Figure 5.6 Variation in dose with depth along the central beam axis in the homogeneous phantom when irradiated with a 10 MV (a) $5 \times 5 \text{ cm}^2$ field, (b) $10 \times 10 \text{ cm}^2$ field. Film was parallel with the central beam axis

Figure 5.7 Variation in dose with depth along the central beam axis in a solid water phantom when irradiated with a 6 MV beam and $10 \times 10 \text{ cm}^2$ field. Film was parallel with the central beam axis

Figures 5.8 through 5.10 show the results when the heterogeneous phantom was irradiated. Again, MOSFET and film measured doses agree at both points for all energies and fields. Calculation and measurements for the cobalt-60 beam agree at the first air-acrylic interface. The calculated dose is notably greater than the measured dose in the air cavity. Within a few mm of the second air-acrylic interface, the superposition algorithms agree with measured dose while the FFTC algorithm continues to estimate a dose greater than the film. The difference between calculated and measured doses appears to be greater for the smaller field ($5 \times 5 \text{ cm}^2$). Calculations and measurements for the 6 MV beam agree at the first air-acrylic interface. The PrecisePlan XVMC calculated dose agrees with the GafChromic film measured dose in the air cavity of the phantom. The Clarkson, FFT convolution, superposition, and multi-grid superposition calculations estimate a dose greater than that measured by the film. The difference between calculation and measurement is greater for Clarkson integration and FFTC.

The superposition algorithm agrees with measured dose approximately 10 mm downstream of the second air-acrylic interface. Clarkson and FFTC agree with the measured dose approximately 10 mm downstream of the second air-acrylic interface. Calculations and measurements for the 10 MV beam are consistent with the cobalt-60 and 6 MV results. All calculated doses agree with measurement at the first air-acrylic interface and predict a larger dose than measured in the air cavity. Predicted doses differ from measured dose more in the $5 \times 5 \text{ cm}^2$ field than the $10 \times 10 \text{ cm}^2$ field. FFTC and Clarkson integration algorithms show the greatest difference from measured dose. The superposition algorithm agrees with measured dose approximately 20 mm downstream of the second air-acrylic interface.

Figure 5.8 Variation in dose with depth along the central beam axis in the heterogeneous phantom when irradiated with a cobalt-60 (a) $5 \times 5 \text{ cm}^2$ field, (b) $10 \times 10 \text{ cm}^2$ field. Film was parallel with the central beam axis

Figure 5.9 Variation in dose with depth along the central beam axis in the heterogeneous phantom when irradiated with a 6 MV (a) 5x5 cm² field, (b) 10x10 cm² field. Film was parallel with the central beam axis

Figure 5.10 Variation in dose with depth along the central beam axis in the heterogeneous phantom when irradiated with a 10 MV (a) $5 \times 5 \text{ cm}^2$ field, (b) $10 \times 10 \text{ cm}^2$ field. Film was parallel with the central beam axis

The ratios of calculated to measured dose (r-value) at the second dose point are presented in figures 5.11 and 5.12:

$$r = D_c / D_M \quad [5.4]$$

where D_C is the dose calculated by a specific algorithm and D_M is the dose measured by the MOSFET at the second dose point (0.5 mm past the second air-acrylic interface). The error bars shown in figures 5.11 and 5.12 are calculated from the standard deviation on the MOSFET measurements:

$$\delta(r) = \left(D_c / D_M^2 \right) \cdot \delta D_M \quad [5.5]$$

where D_M has been calculated according to equation 5.3

Figure 5.11 Ratio of calculated dose and measured MOSFET dose for varying field sizes and calculation algorithms in (a) Cobalt-60, (b) 6 MV, (c) 10 MV beams. Error bars represent the uncertainty associated with the MOSFET dose measurement (equation 5.5)

Figure 5.12 Ratio of fast superposition calculated dose and measured MOSFET dose for varying field sizes and beam energies. Error bars represent the uncertainty associated with the MOSFET dose measurement (equation 5.5)

5.5 Discussion

Homogeneous phantom

The Clarkson Integration algorithm does not account for the curved surface of homogeneous neck phantom. A curved surface will scatter less radiation than a flat surface. The decrease in dose caused by less scatter is not predicted by the Clarkson integration algorithm, resulting in a calculated dose which is greater than the measured dose. The difference between predicted and measured dose is greater for the lower beam energy (6 MV) and larger field ($10 \times 10 \text{ cm}^2$). This is because beams with low energy photons and large fields have a higher proportion of scatter radiation.

The FFTC and superposition algorithms agree with each other in the homogeneous phantom. This is because their algorithms are nearly identical when no variations in electron density are present.

Heterogeneous phantom

Calculations and measurements agree at the first dose measurement points for all energies and fields. This is because the measurement point is upstream of the air cavity and any change in dose would be a result of a change in backscatter. Backscatter represents 1-2% of total dose. For instance, if the predicted backscatter dose was 30% greater than the true amount, this would lead to a sub-percent increase in total dose. Since the film and MOSFETs have an uncertainty of $\sim 2\%$, a sub-percent variation in dose would be considered insignificant.

The difference between measured and predicted dose is greatest for the Clarkson integration algorithm. For both the 6 MV and 10 MV beams, the difference between measured and predicted dose was 19% in the $4 \times 4 \text{ cm}^2$ field at the second dose

measurement point. This difference decreased as the field size increased because transient charged particle equilibrium becomes established and less energy is scattered away. In the $10 \times 10 \text{ cm}^2$ field, the dose predicted by the algorithm is 5% and 8% higher than measured by the MOSFET at the second measurement point in the 6 MV and 10 MV beam, respectively. Measured dose is less than the Clarkson integration predicted dose because the algorithm dose not account for changes in the range of secondary particles. The algorithm scales a measured dose to water according to TAR values and an inverse square correction (see equation 2.15). Inhomogeneities are only considered in the attenuation of primary fluence because dose is calculated at a depth equal to the eEPL. When electrons in the heterogeneous neck phantom travel through the air cavity, either because they were set into motion in the cavity or because they passed the air-acrylic interface, their range increases and more energy is scattered away from the central axis. The decrease in dose caused by an increase in electron range is not predicted by the algorithm. The difference between predicted and measured dose is larger for higher beam energies because the secondary particles will have higher initial energies and longer ranges. The difference is also less in larger fields because lateral CPE remains established through low density regions of the phantom. The minimum field required to establish CPE increases when a particles range increases. If lateral CPE is not established, more energy will be scattered away from the central beam axis.

The FFTC algorithm is similar to the Clarkson integration algorithm because the only consideration of tissue inhomogeneity is taken in the attenuation of primary fluence. The Clarkson integration and FFTC predicted doses agree with each other at the second dose measurement point for all fields and beam energies within the uncertainty. The

FFTC algorithm predicts a lower dose than the Clarkson integration algorithm in the air cavity. This is because the FFTC algorithm accounts for a decrease in TERMA caused by air's lower mass attenuation coefficient. The air mass attenuation coefficient is approximately 10% less than the water mass attenuation coefficient in the energy range of interest. A decrease in mass attenuation coefficient means fewer interactions will occur. The FFTC algorithm will predict a decrease in dose caused by the lower probability of primary photon interaction in air, whereas the Clarkson integration algorithm will not.

The superposition and fast superposition algorithms predict doses closer to the measured doses than the Clarkson integration and FFTC algorithms. Calculated dose from the multi-grid superposition algorithm is within 1% of the calculated dose from the superposition algorithm for all fields and beam energies. Doses predicted by the algorithms at the second measurement point are larger than measured doses. In a 4x4 cm² field, the 6 MV and 10 MV calculated doses are 11% and 9% greater than the measured doses, respectively. The difference between prediction and measurement is less for larger fields. In a 10x10 cm² field, the 6 MV and 10 MV calculated doses are 3% and 6% greater than measured doses. The superposition algorithms account for variations in scatter radiation by scaling the energy deposition kernels according to the average density between the interaction and energy deposition points. Figure 2.6 shows the ratio of air and water average linear energy transfer. For photon energies less than 4 MeV, the water average mass collisional stopping power is larger than that of air. Density scaling of a water energy deposition kernel to represent an air energy deposition kernel will result in a kernel which is too small if the primary photon had less than 4 MeV of energy. Calculations done with a stopping power which is too large is equivalent to using a range

which is too small. Therefore, when the superposition algorithms calculate dose deposited by photons with energy less than 4 MeV when the secondary particles travel through an air cavity, they will use an electron range which is too small. This situation is similar to when dose is calculated by Clarkson integration or FFTC because not all the energy scattered away by high-range electrons is calculated. If the photons have energies greater than 4 MeV, the effect is reversed and the algorithms will calculate a range which is too long. In that case, the algorithm will calculate too much energy being scattered away. Figure 5.3 shows the approximate energy spectra for the 6 and 10 MV beams used for this project. The 6 MV beam peaks at 0.75 MeV and has 88% of its total energy fluence is less than 4 MeV. The 10 MV beam peaks at 1.5 MeV and has 74% of its total energy fluence is less than 4 MeV. Despite the 6 MV beam having a greater percentage of photons with energy less than 4 MeV, calculations done in the 6 MV beam are closer to the measured doses for all fields except the $4 \times 4 \text{ cm}^2$. This is because the electrons set in motion by the higher energy photons of the 10 MV beam have a greater range and scatter more energy away than the electron set into motion in a 6 MV beam.

Differences between measurement and calculation are the smallest for the XVMC algorithm. The average difference between measurement and calculation at the second dose measurement point was 1.7%. The calculated dose is consistently lower than the measured dose and the difference does not appear to be a function of field. The lower calculated dose may be caused by voxel averaging effects. The XVMC calculation voxels were $2 \times 2 \times 2 \text{ mm}^3$ and the MOSFET dose measurement point was 0.5 mm below the second air-acrylic surface. If the XVMC calculation voxel at the first or second dose

measurement point contained a portion of air, the overall dose to the voxel would be reduced.

Chapter 6

Conclusion and Summary

According to the AAPM Report #85, dose algorithm accuracy should be within 2% to meet the recommended overall dose delivery uncertainty of 5%. Accuracy of an algorithm was determined by comparing precisely measured doses with the predicted dose. All of the tested algorithms meet the AAPM standard when calculating dose in a homogeneous acrylic phantom. However, when an air cavity is introduced most algorithms fail to calculate accurate dose values in the air cavity and immediately beyond.

The Clarkson integration calculated doses were 5% to 19% larger than the measured doses at the second dose measurement point. The FFT convolution calculated doses showed similar results and were 4% to 18% greater than measurements. Neither of these algorithms met the AAPM standard for any of the tested beam energies or field. Both superposition algorithms calculated doses were between 2% and 11% larger than measurements. The cobalt-60 beam with a $10 \times 10 \text{ cm}^2$ field was the only instance when the superposition algorithms met the recommended 2% accuracy standard. For fields 5×5 , 6×6 , 8×8 , and $10 \times 10 \text{ cm}^2$, the superposition algorithm performed better in the 6 MV beam compared to the 10 MV beam (7%, 6%, 4%, and 3% versus to 8%, 8%, 5%, and 6%). However, the superposition dose calculation with a $4 \times 4 \text{ cm}^2$ field showed slightly better agreement with measurements in the 10 MV beam compared to the 6 MV beam (11% in 6 MV compared to 10% in 10 MV). The XVMC algorithm calculated doses were 0% to 3% different from measurement. The calculation met the AAPM standard

when using field sizes 4x4, 6x6, and 8x8 cm². Improved results could possibly be obtained if a smaller calculation voxel was used for the XVMC algorithm.

For single beam treatment plans of laryngeal cancer when the tumour is located immediately beyond the trachea, this work indicates to use an XVMC algorithm to calculate the delivered dose. If it is not possible to use an XVMC algorithm, superposition algorithms for low beam energy and large fields give the lowest possible treatment uncertainty.

Parallel-opposed beam treatment is when the tumour is treated with two equal beams 180° apart. If a tumour located beside the trachea is treated with parallel-opposed beams, it will be immediately beyond the trachea for one beam and immediately before the trachea for the other. This will cut the calculated uncertainties in half because the first beam will have an uncertainty similar to a single beam treatment while the second beam delivers a dose approximately equal to the calculation. In this case, the superposition algorithms may meet the AAPM standard for fields 6x6 cm² and larger in a cobalt-60 beam, and 8x8 cm² and larger in a 6 MV beam. For small fields or a 10 MV beam, the standards are still not met. This work indicates that for parallel-opposed beam treatment of a tumour located beside the trachea and dose calculations done using a superposition algorithm, treatments done with a 6 MV beam should use fields 8x8 cm² and larger. A 6x6 cm² field may be appropriate if the tumour is being irradiated in a cobalt-60 beam. Clarkson integration and FFTC algorithms should not be used and treatment should not be done with a 10 MV beam.

Conformal radiation therapy uses many beams and small fields. Increasing the number of fields may help lower the calculation uncertainty if the majority of the beams

locate the tumour downstream of the trachea. Superposition algorithms may be able to calculate dose within the AAPM standards for 10 MV beams and small fields if an appropriate combination of beam angles is used. Calculation uncertainties should be taken into consideration when preparing treatment plans for high energy, small field beams.

6.1 Summary

MOSFETs and GafChromic film were characterized as absolute dosimeters and used to calculate dose delivered to an acrylic neck phantom. The measurements were compared with dose values calculated by various treatment planning algorithms. The difference between calculation and measurement was used to determine algorithm uncertainty when an air cavity is present.

MOSFET calibration coefficients were found to depend on energy and orientation, and to vary between detectors. The average calibration coefficients were 0.90 ± 0.03 , 0.95 ± 0.02 , and 0.97 ± 0.03 cGy/mV for cobalt-60, 6 MV, and 10 MV beams, respectively. Individual MOSFETs used to calculate dose at 52.5 and 77.5 mm in the acrylic neck phantoms had uncertainties less than 2% at two standard deviations. The GafChromic film calibration showed no dependence on energy or batch. The calibration parameters passed a χ^2 *goodness of fit* test at a 99% confidence level. Uncertainty in dose calculated by the calibration curve was 1.6% at two standard deviations.

Clarkson integration and FFTC calculated doses immediately beyond the air cavity were considerably larger than the measured doses. Superposition calculated doses showed an improvement over those of Clarkson integration and FFTC but were still

significantly larger than the measured doses. XVMC calculated doses showed general agreement with measurements.

For single beam treatments with the tumour located immediately beyond the trachea, it is recommended to use an XVMC algorithm. Parallel-opposed beam treatments should preferably be done in a 6 MV beam with a field $8 \times 8 \text{ cm}^2$ or larger. A $6 \times 6 \text{ cm}^2$ field may be used for treatments with a cobalt-60 beam. For high energy beams and small fields, an appropriate combination of beams which gives a total dose calculation uncertainty of less than 2% should be used.

The uncertainties in dose calculation which arise from the presence of an air cavity in the trachea should be considered when planning radiation treatments for patients with laryngeal cancer.

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