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Soh, Roger Cai Xiang

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Monte Carlo simulation and dosimetric studies for small fields in lung stereotactic body radiation therapy

Roger Soh Cai Xiang

School of Physical and Mathematical Sciences
Division of Physics and Applied Physics
Nanyang Technological University

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To God Almighty whom led me through the valley of the shadow of death,
my unconditionally loving wife Joo Leng, and my adorable son Stuart.

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The acknowledgement in this thesis was written in the form of a letter to my 9 months old boy, Stuart Soh.

My dearest Stuart boy,

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With Unconditional Love,

Your Papa

Roger Soh

2017

List of Abbreviations used

γ analysis	Gamma Analysis
3D-CRT	Three-Dimensional Conformal Radiation Therapy
4D-CT	Four-Dimensional Computed Tomography
AAA	Anisotropic Analytical Algorithm
AAAv10.0	Anisotropic Analytical Algorithm version 10.0
AAAv13.6	Anisotropic Analytical Algorithm version 13.6
AAPM	American Association of Physicists in Medicine
AIP	Average Intensity Projection
AXB	Acuros XB dose calculation algorithm
AXBv10.0	Acuros XB dose calculation algorithm version 10.0
AXBv13.6	Acuros XB dose calculation algorithm version 13.6
BEAMDP	BEAM Data Processor
BED	Biologically Effective Doses
CAX	Central Axis
CBCT	Cone-beam Computed Tomography
CC	Collapsed Cone
CIRS plastic water	Computerized Imaging Reference Systems Plastic Water Diagnostic Therapy
CM	Component Module
CPE	Charged Particle Equilibrium
CPU	Central Processing Unit
CSDA	Continuous Slowing Down Approximation
CT	Computed Tomography
CTV	Clinical Treatment Volume
dpi	Dots Per Inch
DTA	Distance-To-Agreement
DVH	Dose Volume Histograms

EBT3	External Beam Therapy 3
ECC	Element Correction Coefficient
EGSnrc	Electron Gamma Shower Monte Carlo Simulation Package
FFF	Flattening Filter-Free
FFT	Fast-Fourier Transform
FWHM	Full Width At Half Maximum
GBBS	Grid-based Boltzmann Equation Solver
GBD	Golden Beam Data
HU	Hounsfield Units
IAEA	International Atomic Energy Agency
ICRU	International Commission on Radiation Units and Measurements
IGRT	Image-guided Radiation Therapy
IMRT	Intensity Modulated Radiation Therapy
ITV	Internal Target Volume
KERMA	Kinetic Energy Released Per Unit Mass
KERMA _{col}	Collisional KERMA
LBTE	Linear Boltzmann Transport Equation
LED	Lateral Electron Disequilibrium
LiF:Mg,Ti	Lithium Fluoride crystals, doped with Magnesium and Titanium
Linac	Linear Accelerator
MC	Monte Carlo
MLC	Multi-Leaf Collimator
msr	Machine-Specific Reference Field
NSCLC	Non-small Cell Lung Cancer
OD	Optical Density
OMP	Oncentra MasterPlan
OOF	Out-of-field

PMMA	Polymethylmethacrylate
PMT	Photomultiplier Tube
PSDL	Primary Standard Dosimetry Laboratories
PTV	Planning Target Volume
QA	Quality Assurance
RCF	Reader Calibration Factor
SAD	Source-to-Axis Distance
SBRT	Stereotactic Body Radiation Therapy
SNC	Sun Nuclear Corporation
SRS	Stereotactic Radiosurgery
SSD	Source-to-Surface Distance
TCP	Tumour Control Probability
TCPE	Transient Charge Particle Equilibrium
TERMA	Total Energy Released Per Unit Mass
TG-105	Task Group 105
TLD	Thermoluminescence Dosimeters
TPS	Treatment Planning System
TRS-398	Technical Report Series 398
VMAT	Volumetrically Modulated Arc Therapy
Voxels	Volume Elements

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Abstract

The use of small fields in radiotherapy has increased over the recent years as complex and precise radiotherapy techniques are being developed. However, prediction and verification of small field doses continue to prove to be difficult due to several challenges such as lateral electron disequilibrium (LED), partial source occlusion effect and detector perturbation. Accurate prediction of small field doses can be achieved using Monte Carlo (MC) simulations. MC simulations are extremely versatile and can be used to accurately predict doses in any medium provided that the linear accelerator (linac) treatment head and associated parameters are well modelled.

The aim of this thesis is to use MC simulation to study the accuracy of commercial dose calculation algorithms used in treatment planning and to correct dose perturbation for radiation dosimeters for small lung SBRT fields. This is done using a three pronged approach, (1) by the determination of optimal small field parameters for MC simulation (2) by evaluating the accuracy of dose calculation algorithms for small lung SBRT field sizes in the clinical treatment planning system (TPS) and (3) by accurate measurement of dose profiles using calibrated film and thermoluminescence small field dosimetry.

The first step for MC simulation is to generate a clinically realistic MC linac model by identifying the optimal parameters for accurate MC simulation. Electron Gamma Shower Monte Carlo Simulation Package developed by the Ionizing Radiation Standards Group at National Research Council of Canada (EGSnrc) will be used as the MC code. Field-size specific MC simulated doses of a 6 MV photon beam were validated using 1D gamma analysis. Simulated data were benchmarked against measurements obtained using EDGE Detector (Sun Nuclear Corporation, Melbourne, FL) and Diode SRS detector (PTW-60018, Freiburg, Germany) in a 3D scanning water tank (Sun Nuclear Corporation, Melbourne, FL). Initial MC reference parameters were approximated using average percentage depth and lateral dose profiles differences as simulated from different mean electron energy and electron beam radial distribution (full width at half maximum, FWHM). Subsequently, the optimal parameters were obtained by 1D gamma analysis using increasingly stringent gamma criteria from $\gamma_{0.3\%/0.3mm}$ to $\gamma_{2.0\%/2.0mm}$ for depth dose and lateral dose profiles. By using a 95%

passing rate, a generic set of optimal primary electron beam parameters in a MC model for all field sizes was accurately determined.

The simulated 6 MV photon beam MC model was used as a benchmark to evaluate two versions of two dose calculation algorithms available in the Eclipse TPS (Varian Medical System Inc, USA), namely, Anisotropic Analytical Algorithm (AAA) version 10.0 (AAAv10.0), AAA version 13.6 (AAAv13.6) and Acuros XB dose calculation (AXB) algorithm version 10.0 (AXBv10.0), AXB version 13.6 (AXBv13.6). Depth dose distributions for 6 MV flattening filter-free (6X FFF) photon beam were also included. The study was done on a respiratory motion phantom (QUASAR, Modus Medical Devices, London, ON) with a moving chest wall. For the chest wall, a 4 cm thick wax mould was attached to the lung insert of the phantom. Average intensity projection (AIP) images from the four-dimensional computed tomography (4D-CT) scans were used for depth dose calculation. Depth doses along the central axis were compared in the anterior and lateral beam direction for field sizes $2 \times 2 \text{ cm}^2$, $4 \times 4 \text{ cm}^2$ and $10 \times 10 \text{ cm}^2$. For the lateral beam, the surface dose of the moving chest wall highlighted differences of up to 105% for AAAv10.0, 40% for AXBv10.0 and 20% for 6X FFF from MC simulations. AAAv13.6 and AXBv13.6 agrees with MC predictions to within 10% at similar depth. For anterior beam doses, dose differences for both versions of AAA, AXB algorithm and 6X FFF beam were within 7% and results were consistent with static heterogeneous studies.

Small field dosimetry within a lung phantom can be a challenge due to dose perturbation of the detector within the low-density lung medium. For thermoluminescence dosimeters (TLDs), accurate measurement can be achieved by correcting the perturbed dose using TLD correction factors calculated from MC simulations. MC calculated TLD correction factors were compared against depth and field size within a static lung phantom. It was found that the greatest TLD correction needed was for the smallest field size investigated, $2 \times 2 \text{ cm}^2$, where LED was the greatest. Correction at interfaces of cork (lung substitute medium) and water were dependent on the medium before the interface. In addition, the MC corrected TLD measurement was compared to AXBv13.6 $2 \times 2 \text{ cm}^2$ dose calculation to test the accuracy of the calculated correction factors. Finally, TLD corrected measurements and radiochromic films (Gafchromic External Beam Therapy 3 (EBT3), International Specialty Products, Wayne, NJ) were used to verify the MC and TPS simulated depth doses in the respiratory lung motion phantom. Both MC and AXB v13.6 depth doses were found to have the best agreement to within 2% deviation. A VMAT lung SBRT treatment plan with three arcs was computed with AXBv13.6, AAAv12.6, AXBv10.0 and AAAv10.0. The deliverability of the treatment plan delivery was verified using a diode array (ArcCHECK, Sun Nuclear Corporation, Melbourne, USA). Quality assurance (QA) of the plan was also

done using film dosimetry. 2D gamma analysis show that AXB v13.6 have the best agreement with film dosimetry and diode array measurements.

In conclusion, MC simulation model developed in this thesis was used to validate the accuracy of commercial dose calculation algorithms and determine the corrections needed for TLD in small fields and lung medium. The MC simulation: (1) when correctly modelled and validated by measurement, can be used as an independent benchmark for studying the accuracy commercial treatment planning systems; (2) is suitable in correcting TLD small field measurements; (3) is versatile, allowing users to verify the accuracy of treatment delivery even in extreme inhomogeneous conditions. These are important findings that will contribute to small photon field dosimetry for safer lung SBRT treatments.

Chapter 1

Radiation therapy for lung cancer

1.1 General introduction

1.1.1 A clinical case

Figure 1.1 shows a computed tomography (CT) scan of a middle-aged patient who presents with a small lung lesion. This patient represents a typical lung cancer patient with stage T1 N0 M0 (early stage tumour without nodal involvement or metastases). According to the Singapore Cancer Registry (2016), lung cancer accounts for almost a third of all cancer deaths among the male resident population and is the second most frequent cause of cancer deaths among female residents in Singapore. Treatment of lung cancer remains very challenging, where mortality rate is approximately 30%, which is the highest among other forms of cancer [97]. Patients who present with a small lung lesion as depicted in Figure 1.1 usually have 2 major options for treatment: (1) surgery or (2) radiation therapy. Both options have their advantages and disadvantages in terms of tumour control and treatment complication risks.

Figure 1.1 An axial CT slice of a sample patient with a small lung tumour in the right lung (left side of the image).

1.1.2 Surgery or conventional radiation therapy?

For early-staged lung cancer, surgery may be a good option for patients. However, if the patient is unfit for surgery due to complications such as heart disease, chronic obstructive pulmonary disease or overall poor health, the patient may not tolerate the invasive treatment of an open chest surgery [138]. Statistics showed that only 10% of all lung cancer patients diagnosed with non-small cell lung cancer (NSCLC) are suitable for surgery [47]. In such a situation, the non-invasive treatment of 3-dimensional conformal radiation therapy (3D-CRT) is the preferred option.

Figure 1.2 shows the total dose from two overlapping fields of radiation focused onto a small lung tumour. The 'colour wash' indicates the dose distribution with low doses outside the tumour region

3D-CRT is a treatment technique that uses multiple beams of x-ray radiation delivered over many treatments sessions (e.g. $2 \text{ Gy} \times 30 \text{ fractions} = 60 \text{ Gy}$). Figure 1.2 shows an example of a 3D-CRT treatment with two overlapping radiation beams. Clinical 3D-CRT treatments usually consist of 5-8 overlapping fields.

A typical 3D-CRT dose prescription for an early-staged lung cancer patient is 55-70 Gy over a 4 to 7 week period [122]. However, studies has shown that patients with these dose schedules displayed poor tumour control probability (TCP) levels of 30-40% (TCP: probability that a tumour is controlled or eradicated given a prescribed treatment) and low five year survival rates (20%) [139, 161]. Studies which increased radiation dose per fraction to improve TCP on the other hand, showed improved overall survival rates and reduced local tumour recurrence rates [90, 115]. The increasing dose levels, however, will lead to an increased probability of associated lung toxicity. Lung toxicity causes complications such as radiation pneumonitis, fibrosis and even death [15]. With the disadvantages from both standard 3DCRT and surgery, any lung cancer patient may decide to forsake treatment altogether, a decision based on poor prognosis [39]. With the current demand for an effective treatment alternative,

radiation oncologists and medical physicists have found a more effective radiation treatment method for patients with early-stage lung cancer.

1.1.3 Stereotactic Body Radiation Therapy (SBRT)

Development of SBRT

Stereotactic radiosurgery (SRS) is used to treat malignant tumours along the central nervous system such as the brain [86]. Radiosurgery is a non-invasive technique that uses a single, high dose fraction of radiation (i.e. 24 Gy in one exposure) to treat intracranial tumours. As such, radiosurgery is effective in tumour control while sparing surrounding healthy tissue [124]. Beam precision will be vital to target the tumour, which can be achieved using accompanying imaging techniques, fiducial markers and stereotactic frame to immobilize the skull during treatment [73].

It was found that SRS could be adapted to treat early-staged lung cancer patients who were not suitable for surgery. This is known as lung stereotactic body radiation therapy (lung SBRT). However, the adaptation of the SRS treatment technique from brain to lung is not straightforward due to 2 main factors among others, namely: 1) tumour motion due to respiration and 2) complicated radiation dose modelling and small field dosimetry in low-density inhomogeneous lung medium. Lung tumour motion can be several centimetres in various directions. These displacements may result in under irradiation of the lung tumour and over irradiation of the surrounding healthy lung tissue. Therefore, the aim of lung SBRT is to precisely deliver a focused ablative dose of radiation to a small mobile lung tumour.

With advances in imaging and radiation delivery techniques, small moving tumours within the thoracic cavity can be treated using lung SBRT. Dose escalation is now used for early-stage lung cancer patients with prescription doses of 48 - 60 Gy delivered in 3 to 8 fractions (this corresponds to biologically effective doses (BED) ≥ 100 Gy, which is comparable to surgery). Lung SBRT has several advantages, namely 1) reduced total treatment time to within 2.5 weeks, 2) shortened treatment times prevents tumour cell repopulation, 3) large doses per fraction may overcome ineffective cell kill rates due to hypoxic (oxygen deficient) tumours, if re-oxygenation between fraction occurs, and 4) TCP can potentially exceed 90% with low levels of toxicities [58, 85, 107]. Clinically, the advantages of lung SBRT are therefore beneficial for patients.

With the development of advanced lung-SBRT techniques, there is a possibility of increasing the TCP to beyond the 90% level by increasing the tumour dose. However, surrounding healthy lung tissues must not be compromised. A balance of maintaining lung dose to well-tolerated levels and increasing tumour dose levels is the focus of advanced lung SBRT techniques. The challenge of this development lies in the under-

standing of the complex radiation dosimetry of SBRT fields within inhomogeneous lung tissues.

Imaging techniques for lung SBRT treatments

Figure 1.3 A Varian Clinac iX Linac with CBCT module mounted.

Image-guided radiation therapy (IGRT) is an advanced treatment technique that combines the treatment capabilities of the clinical linear accelerator (linac) with the addition of an imaging module. An example is the Varian Clinac iX Linac (Varian Medical Systems, Palo Alto, USA), where a cone-beam CT (CBCT) scanner is mounted on to the linac as shown in Figure 1.3. With CBCT, the positioning of the target volume with reference to the patient anatomy can be verified before beam delivery. This is done by acquiring the time-averaged CT volume of the patient thorax during respiration when the patient is lying on the treatment couch [13]. When combined with intensity modulated radiation therapy (IMRT) (technique that shapes the photon beam to conform to the shape of the target volume) or volumetrically modulated arc therapy (VMAT) (an extension of IMRT using a novel rotational technique), small treatment fields can be accurately delivered onto a small target, reducing misalignment between patient tissues and radiation beam [87, 13] (Figure 1.4).

Volumetric images of patient anatomy can be obtained using 3-dimensional CT (3D-CT). 3D-CT images comprises of voxels (or volume elements) that assign specific CT number in accordance to the tissue attenuation of the X-ray photons (in Hounsfield

Figure 1.4 shows an example of lung SBRT VMAT treatment plan with 4 arcs. The ‘colour wash’ shows that the dose distribution of the plan is highly conformal to the target volume

Units or HU). 3D-CT images are subjected to artefacts such as cupping (dark artificially hypo-dense image regions) and streaking (lines of hyper/hypo-dense image regions), which are due to movement during scanning [110]. Under respiratory motion, motion artefacts such as blurring will be present in the 3D-CT scan. These image artefacts within 3D-CT reconstructions may impede visualisation of tumour volume definition and subsequent treatment delivery. 3D-CT scans can be unreliable as mass and electron density information derived from CT numbers used directly for dose calculations may be inaccurate.

Presently, lung SBRT techniques include 4-dimensional CT (4D-CT), CBCT and gating systems to overcome challenges in lung SBRT treatments. 4D-CT can be used to determine the *in-vivo* path of a lung tumour over the course of the respiration cycle (Figure1.5). In 4D-CT, a patient’s breathing cycle is divided into multiple phases. In each phase, a 3D-CT image is acquired to identify the position of the tumour and thoracic cavity. By taking the average HU over all phases, an Average Intensity Projected (AIP) image set that encompasses the entire tumour motion can be obtained. A treatment margin of the clinical treatment volume (CTV) and the entire path of the tumour motion can be contoured as the internal target volume (ITV) (Figure1.5). The ITV is expanded to form the planning target volume (PTV), which accounts for geometric variations due to inter-fraction uncertainties during patient setup. 4D-CT combined with gating technology allows the treatment planning and delivery of radiation to occur during a single or multiple respiration phases of the patient.

Dose calculation accuracy for treatment planning requires precise electron and mass density information acquired from CT scans. Inhomogeneous and small field conditions may pose challenges for dose calculation accuracy. Therefore, the need

for accurate lung SBRT dose prediction is crucial and will be discussed further in the following chapters.

3 Dimensional
Computed Tomography
(3D-CT)

Average Intensity Projection
Computed Tomography
(AIP CT)

Figure 1.5 Positional differences between 3D-CT and AIP CT coronal CT image slices of a lung cancer patient acquired during respiration. Contours show the ITV and PTV.

1.2 Physics of radiation therapy

The aim of radiation therapy is to maximise the dose to the treatment volume, while sparing the dose to healthy tissue. The following sections will discuss the radiation physics related to radiotherapy, in particular dose calculation algorithms and small field dosimetry.

TERMA, KERMA and KERMA_{col}

Radiation therapy begins with the distribution of photon fluence starting from the surface of a patient. As a photon beam travels through several layers of tissue within the body, the probability of photon interaction or attenuation per unit distance is given by the linear attenuation coefficient, μ [cm^{-1}]. μ is dependent on the mass density (ρ [g/cm^3]) and effective atomic number (Z_{eff}) of the medium, as well as the energy of the incident beam [MeV].

The accumulation of the energy of each photon [MeV] reaching a point within the patient determines the energy fluence Ψ [MeV/cm^2]. Therefore, the total energy released per unit mass (TERMA) at a point within the patient is defined as the product of the mass attenuation coefficient (μ/ρ) and the energy fluence at that point:

$$TERMA = \left(\frac{\mu}{\rho}\right)\Psi \quad (1.1)$$

KERMA is defined as the kinetic energy from incident photons per unit mass that are released to charged particles, such as electrons. When these charged particles dissipate their energy by ionisation or excitation only (excluding radiation losses), it is known as collisional KERMA (KERMA_{col}). Therefore, in terms of energy, $TERMA > KERMA > KERMA_{col}$.

1.2.1 Charged particle interactions, charged particle equilibrium and dose

Photon and electron Interactions in media

For the energy range in modern radiotherapy (4 - 18 MeV), there are three primary photon interactions that contribute to KERMA_{col}: Photoelectric effect, Compton Interaction and pair production. The dominant photon interaction is dependent on the photon energy as well as the medium's effective atomic number (Z_{eff}). For example, for adipose tissue ($Z_{eff}= 7.5$), Compton interaction dominates in the energy range from 0.05MeV to 10 MeV.

Within the medium, photon interactions impart kinetic energy to charged particles, such as Compton electrons. Ejected electrons undergo Coulombic collisions along its path length, converting kinetic energy into ionization and excitation events within the medium. The energy of the charged particle is deposited along its path, as opposed to energy absorbed at a point [91]. The local energy deposition by an electron along its path length is described by the mass collisional stopping power of the medium, S_{col} [75]:

$$\frac{S_{col}}{\rho} = \frac{d\bar{E}}{\rho dl} \quad (1.2)$$

where:

$d\bar{E}$ represents the average rate of energy lost locally per unit path length dl and per unit density of the medium ρ . The range of the electron path, R , can be approximated by integrating $\frac{S_{col}}{\rho}$ over the range of electron energies.

This local energy deposition due to the continuous slowing down process of a single charged particle can encompass hundreds of thousands of interactions. To treat each interaction individually would be inefficient and is not a reasonable approach. For calculation purposes, a condensed history technique condenses a large number of interactions and considered it as a single electron step [11]. The calculations used in condensed history technique are performed based on the continuous slowing down approximation (CSDA) and the multiple scattering theory. This approximation is valid as most of the charged particle interactions involve small energy losses and minor directional changes. The energy deposition events are tracked over the entire course of the particle history and converted to absorbed dose by dividing the accumulated deposited energy in all defined regions with their respective masses.

Charged particle equilibrium (CPE) and lateral electron disequilibrium (LED)

Figure 1.6 Dose delivered to point Q within a water medium from a photon source. Photon traverse within the water medium until attenuated and eject electrons within the medium. The path of the electron passes through point Q and deposit doses along its path

CPE occurs when the energy of the charged particles upon entering and leaving a small dosimetric volume of interest are equal. Ideally under CPE, photon and electron fluence is assumed to be uniform, without attenuation and beam divergence, across the dosimetric volume. Therefore, dose delivered (D) to the volume is equal to $KERMA_{col}$.

$$D = KERMA_{col} \quad (1.3)$$

Figure 1.7 Under TCPE, scattered electrons away from point Q (green arrow) are replaced by scattered electrons towards point Q (blue arrow). However, under LED, scattered electrons away from point Q travel beyond the field edge and cannot be replaced as there are no photon fluence outside the field. This causes an energy imbalance which reduces the dose at point Q

In reality, due to photon attenuation and beam divergence, the dose delivered becomes proportional to $KERMA_{col}$ [91]. Transient charge particle equilibrium (TCPE) is therefore achieved (Figure 1.8). TCPE occurs along the central-axis of a photon beam field if electron equilibrium is sustained in both longitudinal and lateral directions (longitudinal direction being parallel to the direction of the beam). For example, longitudinal disequilibrium exists at shallow depths within an irradiated medium, which represents the dose build-up region in depth dose profiles (Figure 1.6). Similarly, LED occurs when lateral range of electrons becomes greater than or equals to the radius of the field. Under LED, electrons that are ejected from the central axis of the beam are scattered beyond the edge of the beam and are not replaced. This reduces the dose along the beam's central axis. LED is dependent on radiation field size, beam energy and medium density [91]. Lateral electron range can vary for higher beam energies and in lower density media (e.g. lung tissue). For example in lung SBRT treatments, small radiation fields were delivered to target volume within the lung media, making a perfect example of LED. As such, simple dose calculation algorithms in commercial TPS will be inadequate in calculating doses for lung SBRT treatments (Figure 1.7).

1.2.2 Defining dose profile regions

Simulated dose profiles are often benchmarked against measurement data or a set of reference data known as the golden beam data (GBD). GBD is a dosimetric beam dataset, which serves as a form of dosimetric standardisation. The manufacturer, based on the construction of the standard linac model used, provides the GBD. The accuracy

of simulated dose profiles in standard conditions within water medium is evaluated across the depth dose and lateral dose profiles.

Figure 1.8 Defining depth dose profile regions

Depth dose profile typically consists of two regions, namely, buildup dose region and the transient charge particle equilibrium (TCPE) region. Buildup dose refers to the dose region between the surface (depth=0cm) to the depth of the maximum dose. The TCPE region refers to the depth dose region after the maximum dose. For relative comparison of depth dose profiles, profiles were usually normalised to 100% at depth dose maximum (D_{max}) along the central axis [96]. (Figure 1.8).

Similarly, the lateral dose profile typically consists of three regions, namely, central region, penumbra and the out-of-field (OOF) region. For relative comparison, lateral dose profiles are usually normalised to 100% along the central axis. The central region of the profile is defined as the region from 100% to 80% and the penumbra is defined as the region between 80% to 20% of the dose. Lastly, the OOF doses refer to doses that are below 20% of the dose along the central axis [96, 152] (Figure 1.9).

1.3 Dose calculation algorithms

Dose calculation algorithms in clinical TPS can be divided into three different categories, Type A, Type B and Type C models, according to their ability to model electron transport under inhomogeneous conditions [81, 103]. Type A is the most

Figure 1.9 Defining lateral dose profile regions

basic model, which uses equivalent path length correction to account for primary dose conditions under inhomogeneity corrections. Variations in lateral electron transport are not modelled [81, 103]. Nucletron's Oncentra MasterPlan (OMP) pencil beam convolution algorithm (Nucletron Oncentra MasterPlan (OMP) V1.4.1.2) and XiO's Fast-Fourier transform (FFT) convolution algorithm (Computerized Medical Systems XiO Release 4.2.0) are some examples of Type A models.

Type B models such as Analytical Anisotropic Algorithm (AAA) and Collapsed Cone (CC) algorithm, produces results with greater accuracy based on superposition and/or convolution methods [81, 103, 12]. Type B models consider both the lateral and longitudinal scaling for inhomogeneities. The longitudinal distribution is derived from Monte Carlo (MC) calculations and scaled according to equivalent path length. The lateral distribution is scaled with the density of the medium relative to water [81]. In the superposition or convolution algorithms, the primary photon interaction is regarded as a photon source and point kernels outline the divergence of energy away from the area of interaction. For a radiotherapeutic photon beam, a primary photon interaction within a patient tissue produces a subsequent spread of secondary radiation (scattered photons and electrons) and the convolution step integrates the entire process. Under heterogenous media, the attenuation of photon fluence and dose kernel divergence varies according to the media. For example, within the lung medium ($\rho \approx 0.25 \text{ g/cm}^3$), the tissue attenuation of the primary photon fluence is reduced while

the particle range increases with larger dose kernels. For lung SBRT, where CPE is mildly disrupted, convolution algorithms can provide sufficiently accurate dose calculations. However, under high charged particle disequilibrium conditions such as small field sizes ($< 5 \times 5 \text{ cm}^2$), high beam energies such as 6 MV photon beams and low densities ($< 0.4 \text{ g/cm}^3$), larger dose discrepancies will be expected.

More accurate dose calculations can be obtained using Type C models, which scores energy deposition along each electron path [103, 12, 84]. The MC method and Acuros XB (AXB) algorithm are examples of Type C models.

MC simulations can be used to stochastically simulate the behaviour of physical processes where analytical solutions are too complex. MC simulations use the well-known probability distribution of photon interactions for radiotherapy applications. An x-ray through a patient's tissues is calculated using MC simulations in 3 phases: 1) distance to interaction, 2) type of photon interaction (photoelectric, Compton or pair production) and 3) the determination of the angular and energy distribution of secondary particles such as scattered x-rays and electrons. Secondary particles are scored until either it escapes the volume of interest or its energy is exhausted. The trace of steps 1 to 3 is known as the 'particle history'.

For each history, the energy absorbed locally within each simulated voxel is scored. A random number is drawn for a particle in order to establish the length to the next particle interaction, which is dependent on the particle type, its energy and the interacting medium. After transporting this particle along the determined direction and distance, a new random number is drawn in order to determine the next interaction using the medium's cross sectional data. The particle interaction subsequently will repeat and create a chain of interactions. Furthermore, new particles may be generated as a consequence of previous interactions. This interaction process is repeated until the primary particle either reaches a predetermined cutoff energy or leaves the boundaries of the system. A new primary particle is subsequently drawn from the source and the same procedure is repeated. The final output is collected by integrating a large number of particle histories. With a large number of histories, MC generated dose distributions reaches convergence and becomes accurate. Therefore, MC results are often used as the dose calculation benchmark for dose prediction, especially for LED situations [136, 28, 52]. However, MC codes are comprehensive and have long calculation times that are unsuitable to be used clinically. On the other hand, with technological enhancements such as greater computing power and parallel computation, MC based TPS are now becoming commercially available.

An alternate Type C model, the Acuros XB (AXB) algorithm was developed to achieve both efficiency and accuracy specifically for radiotherapy applications. AXB uses a deterministic grid-based Boltzmann equation solver (GBBS) which directly

discretises the linear Boltzmann transport equation (LBTE). LBTE describes the macroscopic behaviour of particle interactions with matter, in space, and in angle and energy domains. The GBBS solves the LBTE within specified voxels iteratively to compute radiation doses. The primary photon source model, the scattered photon and scattered electron fluence are the three main factors that mainly contribute to AXB dose calculations. Ray tracing is performed from the primary photon source to calculate the particle distributions for every voxel in the medium. Subsequently, AXB solves the LBTE to obtain the scattered photon and electron fluences for every voxel in the medium. Eventually, the calculation of the dose value is obtained from the total particle fluence to dose response function. Unlike Type B algorithms, which model inhomogeneities by applying density corrections relative to water, AXB explicitly models the physical interaction of radiation in the material and solves the radiation-transport problem numerically. AXB have the same cross-section library used from the Electron Gamma Shower Monte Carlo simulation package. For each voxel in the CT image, AXB assigns a mass density and material type for a given structure. This material assignment can be done automatically by referring to the CT calibration curve in the planning system or can be assigned manually. Several studies have shown that AXB calculations were comparable to MC simulations [151, 18, 52, 60]. Their study found an excellent $\pm 2\%$ agreement in depth dose profiles through a heterogeneous unit density phantom and $\pm 2\%$ agreement in 99.9% of voxels [151]. AXB is capable of modelling radiotherapy dose deposition with accuracy comparable to MC simulations, especially in the high dose gradient regions, but with greater efficiency and shorter computation times, making it suitable for radiotherapy applications [52, 18, 60].

1.3.1 EGSnrc Monte Carlo codes

The research MC code of choice for this thesis is the Electron Gamma Shower Monte Carlo simulation package developed by the Ionizing Radiation Standards Group at the National Research Council Canada (EGSnrc). The EGSnrc MC simulation package is readily available and is one of the most used codes in MC simulation for radiation therapy applications. EGSnrc simulation accuracy can be easily found [120, 2, 129, 128]. Several literatures have demonstrated strong agreement between experimental measurements and EGSnrc dose calculations for keV and MeV photons within water and inhomogeneous medium (inhomogeneous interfaces between tissue, air, bone, and metal and ceramic implants and shielding) [89, 32, 120]. Ionisation chambers [159, 17], solid state detectors [158] and films [135] were used for experimental measurement to validate the MC model.

BEAMnrc geometry modelling

BEAMnrc, found in the EGSnrc simulation package, is an user code to simulate a clinical linac treatment head. The code allows users to build the linac treatment head geometry using various structures of independent, non-overlapping blocks known as component modules. The description of each component module is specified based on detailed information on the geometry, material specifications, input parameters and cross sectional data. For example, the flattening filter of a linac is modelled using the FLATFILT component module. This is done by specifying the number of conical layers with different radius values and material definitions. The output data is collected on a scoring plane after the component module, which outputs as phase space files. These phase space files contain data regarding energy deposition, fluence results, processing time and phase space outputs. A BEAM Data Processor (BEAMDP) utility is included within BEAMnrc and can be used to derive the particle energy spectrum, mean energy, angular distributions and planar fluence from the phase space files generated. Generally, phase space files are used as a plane source for subsequent phantom dose simulation. The simulation process is presented in Figure 1.10 below.

Figure 1.10 A flow chart of the simulation using the BEAMnrc code system

DOSXYZnrc simulation

Using the phase space file from a BEAMnrc treatment head simulation, dose calculation can be performed with DOSXYZnrc. DOSXYZnrc is the user code in the EGSnrc simulation package that is used to score absorbed dose within a voxelised phantom geometry [157]. DOSXYZnrc phantom can be built from a CT dataset for dose calculation. Users may define the size of a virtual phantom using its corresponding volume elements (voxels) in the phantom. These voxels can be assigned the desired density and materials. Cross-sectional data of the materials are defined in the PEGS4 material library. The user is required to input the distance and direction of the source plane with regard to the phantom as illustrated in Figure 1.11.

Figure 1.11 Illustration of the position of source plane with respect to the phantom coordinates during DOSXYZnrc simulation

1.4 Photon field dosimetry

1.4.1 Active and passive detectors

Radiation detectors used in photon field dosimetry can be classified under two groups, active and passive detectors.

An active radiation detector measures radiation exposure and is able to give instantaneous real time readout. Typically, active detectors require wired connection to a display, such as an electrometer, for readout.

The ionisation chamber is an example of an active radiation detector. The number of ionisations within the air filled thimble of the ionisation chamber during exposure

were used as a measure of the radiation dose. High voltages were maintained between central and outer electrodes (Figure 1.12). During ionisation, electrons emitted moves towards the positive electrode, inducing an electric current. The electrometer detects the total charge, Q , which is proportional to the dose to air volume in the thimble. Exposure, X is defined as the total number of charges produced due to irradiation in gas with mass m . Exposure will be proportional to the dose delivered and is related by the average energy expended in the production of a pair of ions in gas ($\bar{W}$) per ion pair.

$$X = \frac{dQ}{dm} \quad (1.4)$$

$$D_{air} = \frac{Q}{m} \left(\frac{\bar{W}}{e} \right)_{air} = X \left(\frac{\bar{W}}{e} \right)_{air} \quad (1.5)$$

For clinical photon beam and electron beam energies, $\left(\frac{\bar{W}}{m} \right)$ of air is approximately $33.97 J/C$ [70]. The availability of $\left(\frac{\bar{W}}{m} \right)_{air}$ makes ionisation chambers to be the preferred active detector for absolute dosimetry. Other active detectors available such as calorimetry and Fricke dosimeters are not easy to use, and hence not suitable for routine dosimetry in radiotherapy clinics. Semiconductor diodes are another example of active detectors but they are relative dosimeters and should not be used for beam calibration, since their sensitivity changes with repeated use due to radiation damage.

Figure 1.12 *Illustration of an ionisation chamber*

The other category of radiation detectors is passive detectors. Passive detectors do not provide direct readouts and can operate without any active means. It usually

requires some form of post-processing to determine the final readout. Despite this, passive detectors are useful because they are usually small in physical size and do not require any active wired connections during measurement. This makes passive detectors ideal for *in-vivo* dosimetry (direct measurement of dose delivered to the target volume). Two examples of passive detectors are radiochromic films and thermoluminescent dosimeters. Radiochromic film and thermoluminescence dosimeters are the two passive detectors, which will be introduced later in this thesis.

1.4.2 The current code of practice for clinical photon dosimetry

The International Atomic Energy Agency (IAEA) Technical Report Series 398 (TRS-398) is the current standard code of practice for clinical reference dosimetry for high energy photon and electron beams [5]. The code of practice are based on ionization chamber dosimetry and were written for a wide range of beam qualities and energies. TRS-398 ionization chamber calibration is based on absorbed dose to water measurement standards determined by Primary Standard Dosimetry Laboratories (PSDL)-traceable ionization chambers [5].

The determination of absorbed dose to water for clinical radiotherapy beams involves the application of several coefficients and correction factors to calibrate raw measurement data from standard beam quality to the specific clinic's beam quality. These measurements were conducted in a well-defined reference condition. TRS-398 reference condition setup calibrates photon beams using ionization chamber measurements under conventional field size of $10 \times 10 \text{ cm}^2$ at the isocenter with a constant source-to-axis distance (SAD) of 100 cm (Figure 1.13).

Figure 1.13 The standard reference field condition for calibration: field size = $10 \times 10 \text{ cm}^2$, SAD = 100 cm, at reference depth $z = 10 \text{ cm}$ within water medium

Absorbed dose to water ($D_{w,Q}$) for the clinic's specific beam of quality Q can be determined using ionization chamber measurement under TRS-398 reference condition is given by Equation 1.6.

$$D_{w,Q} = M_Q \cdot N_{D,w,Q_0} \cdot k_{Q,Q_0} \quad (1.6)$$

In Equation 1.6, the quantity M_Q is the ionization chamber readings after correction for ion recombination, temperature and pressure variations, polarity, and electrometer calibration in reference condition. N_{D,w,Q_0} is the absorbed dose to water calibration coefficient obtained from the national standards lab or an accredited dosimetry calibration laboratory for the standard beam quality (Q_0). k_{Q,Q_0} is the beam quality correction factor, which corrects the differences between detector response in the calibration standard beam quality (Q_0) and the specific clinic's beam quality used under reference conditions (Q).

1.4.3 Small photon field dosimetry challenges

However, unlike conventional photon field dosimetry, verification of small field doses via measurement prove to be challenging. Studies from Das *et al* [35] and Verhaegen

et al [155] concluded that significant dosimetric errors has to be accounted for comparisons between small field measurements against MC simulation benchmark. Standard active dosimetric detectors, such as ionization chambers, may not be applicable as its large sensitive volume causes perturbation to the particle fluence within the medium under small photon beams. Therefore, during dosimetric verification or quality assurance (QA) of lung SBRT treatments, further corrections are needed from the standard conventional field sizes measurement protocols for small field measurements [5, 1].

There are several challenges for small photon field dosimetry, namely, LED, partial source occlusion effect, volume averaged detector response, field size determination and detector perturbation.

Lateral electron disequilibrium (LED)

Figure 1.14 A schematic diagram of charged particle equilibrium. Electrons are produced in p due to photon interactions. Charged particle equilibrium occurs when these electrons crosses mass Δm and deposits energy that is equal to the energy of the secondary electrons produced from mass Δm (energy E in the diagram). Therefore, under small photon field conditions, LED occurs, where the energy deposited in Δm is not equal to the energy of the secondary electrons leaving Δm . [42]

During measurement, the material of the radiation detector used will generally be different from the material of the measured medium. The signal from the radiation detector will be proportional to the absorbed dose in this material. Calibration converts the detector's signal to a known reference radiation quality such as absorbed dose in the material. The introduction of a detector into a uniform medium can be analogous to having a cavity within an uniformed medium. Cavity theory relates the dose to the

detector (D_{det}) to dose to medium (D_{med}) with the factor f_Q under arbitrary radiation quality Q (under photon or electron beam)(Equation 1.7). Hence, dose calculations for dosimetry are based on cavity theory. The factor f_Q is based on the charged particle fluence incident on the detector sensitive volume.

$$f_Q = \left(\frac{D_{med}}{D_{det}} \right)_Q \quad (1.7)$$

Under conventional field size dosimetry, the size of a detector is small compared to the field (Figure 1.14). The buildup of dose due to interactions in the detector material is considered to be negligible. In other words, the presence of the detector does not perturb the charged particle fluence. Therefore, cavity theory is simplified (Bragg-Gray cavity theory) and CPE is achieved. CPE is obtained in a volume when incoming electrons deposit the same energy as the energy of secondary electrons leaving the volume (Figure 1.14). This allows the doses for the different media to be related by the material's stopping power ratios, which approximate the primary and secondary charged particle energy deposition in the detector volume.

By definition, a small field is a field in which there is LED and Bragg-Gray cavity theory cannot be easily applied. According to Alfonso *et al* [1], the definition of a small field is defined as "a field with a size smaller than the lateral range of charged particles". This typically applies to photon beams of field size smaller or equal to $4 \times 4 \text{ cm}^2$ [71, 35]. At these field sizes, the lateral range of the secondary electrons in a medium is often greater than the lateral dimension of the field, which leads to the loss of lateral CPE and this causes changes in the energy spectra along the lateral direction of the cavity (Figure 1.15). The extent of LED is also dependent on the measured medium. For example, lung tissues have very low density (0.27 g/cm^3) as compared to adipose and tumour tissues. The low density medium results in an increased range of secondary electrons which leads to a greater extent of LED. The consequence of such increased LED result in larger uncertainty of doses within lung and tumour tissues.

Under Charged Particle Disequilibrium

Figure 1.15 A schematic diagram of LED. Under small photon field conditions, the range of the secondary electrons leaving Δm is greater than the dimension of the field. Therefore, the energy deposited in Δm is not equal to the energy of the secondary electrons leaving Δm , resulting in LED.[42]

Partial source occlusion effect

Lung SBRT treatments are often associated with highly collimated fields to allow high precision delivery of doses. However, with highly collimated small fields, the Gaussian-shaped beam source will be partially shielded by the collimators (partial source occlusion), leading to a substantial decrease in the beam output. Furthermore, under small field conditions, the beam source can no longer be regarded as a point source. This results in a wider penumbra region of the beam profile, as more photons from the fringe of the source are blocked and the penumbra is extended over the entire small field [29]. In addition, partial source occlusion produces a charged particle energy spectrum with greater scatter dose component and decreased primary dose component compared to conventional fields. This implies that the dose distribution for small fields will be more sensitive to small variations such as source size, scatter contributors and detector type, resulting in greater uncertainty (Figure 1.16).

Figure 1.16 In small fields, the beam source is shielded by the collimator and is partially seen from the position of measurement as compared to conventional field sizes [6]

Volume-averaged detector response

Small and nonstandard fields do not provide flat dose profiles which results in a charged particle fluence gradient across the detector. The energy deposited in an ionization chamber located on a dose gradient is averaged over the detector volume and is used as the point dose assumption, which results in a systematic underestimation of dose. As the level of LED or dose-gradient across the detector increases, the volume averaging and dose underestimation effect increases. In the case of very small fields, significant errors can occur if the size of the field approaches that of the active volume of the detector. Using inappropriately large volume detectors can and has led to overdosing and severe misadministration. Dosimeters for small and nonstandard fields must provide increased spatial resolution for accurate dose measurements by minimizing volume averaging effects.

Field size determination

For conventional large field sizes, determination of field size is based on the full width at half maximum (FWHM) of the dose profiles, which lies at 50% of the dose profiles. However, as field size decreases, the penumbras from the opposing field edges overlap, resulting in inaccurate field size and consequently dose determinations. Conventional FWHM methods will result in an overestimation of field sizes being larger than the geometric field sizes settings (Figure 1.17). Clinically, this causes a systematic large volume exposure of healthy tissue in the region of the targeted volume, as well as a systematic under dosage of the targeted volume due to the large geometric field size. It will also lead to the miscalculation of dose volume histograms (DVH) as well as tumour control and normal tissue complication probabilities (TCP and NTCP respectively) [55].

Figure 1.17 For conventional field sizes, the field borders for the field size settings is equivalent to the FWHM of dose profiles. However, for small field sizes, the penumbra from opposing field edges overlap resulting in a discrepancy from FWHM data

Detector perturbation

During small field measurement, the presence of a detector in a small field under LED will result in the perturbation of electronic fluence which will further complicate dosimetry. The effects of perturbation are hard to quantify because the detector's medium is often different in density and composition from the measured phantom's medium. However, using passive detectors with small geometries relative to the small field can reduce perturbation effects.

1.4.4 Code of practice for small photon field dosimetry

Small field reference dosimetry

Alfonso *et al* published a formalism to correct for the small field dosimetry. The formalism was recommended for stereotactic machines that are unable to achieve the conventional reference field of $10 \times 10 \text{ cm}^2$ [1]. This practice requires an intermediate step of using machine-specific reference field (msr) to map to a conventional reference field calibration (Figure 1.18). The practice was done under reference dosimetry, which is an standard defining exercise that involves the direct measure of absorbed dose under standard reference conditions.

Therefore, the general small field equation for the absorbed dose to water, $D_{w,Q_{msr}}^{f_{msr}}$, at 10cm reference depth in water, with beam quality Q_{msr} and in machine specific reference field f_{msr} is given by

$$D_{w,Q_{msr}}^{f_{msr}} = M_{Q_{msr}}^{f_{msr}} \cdot N_{D,w,Q_0} \cdot k_{Q,Q_0} \cdot k_{Q_{msr},Q}^{f_{msr},f_{ref}} \quad (1.8)$$

where:

f_{ref} refers to the conventional reference field used during calibration according to TRS-398 code of practice,

f_{msr} denotes the machine-specific reference field for stereotactic machines that are unable to achieve conventional reference field.

Q is the beam quality of the conventional reference field f_{ref} .

Q_{msr} is the beam quality of the machine specific reference field f_{msr} .

$M_{Q_{msr}}^{f_{msr}}$ is the dosimeter reading in the field f_{msr} .

N_{D,w,Q_0} is the calibration coefficient that relates the detector response to the absorbed dose to water at a reference beam quality Q_0 . It is measured at reference conditions according to TRS-398 ($10 \times 10 \text{ cm}^2$ under 10cm water at SAD 100cm).

k_{Q,Q_0} is the beam-quality correction factor, which account for the differences between the detector response under different conditions of field size, geometry, phantom material, and beam quality

$k_{Q_{msr},Q}^{f_{msr},f_{ref}}$ accounts for the differences between the response of a detector in f_{ref} and f_{msr} , which is defined as.

$$k_{Q_{msr},Q}^{f_{msr},f_{ref}} = \frac{D_{w,Q_{msr}}^{f_{msr}} / M_{Q_{msr}}^{f_{msr}}}{D_{w,Q}^{f_{ref}} / M_Q^{f_{ref}}} \quad (1.9)$$

This factor can be obtained through direct calibration by measurement in the conventional and machine specific reference field against a primary standard or against a MC calculated dose.

Figure 1.18 Small field dosimetry formalism. Examples of stereotactic machines that are unable to obtain a reference field of $10 \times 10 \text{ cm}^2$ such as CyberKnife[®], GammaKnife[®] and an example of radiosurgical collimation are shown. [1]

Small field relative dosimetry

Most clinical measurements were done under relative dosimetry, which refers to the use of a secondary standard dosimeter. For example, film dosimetry relies on the production of a known dose exposure on the film to relate relative opacity to dose for accurate dosimetry.

Under relative dosimetry, small field measurements for the absorbed dose to water, $D_{w, Q_{clin}}^{f_{clin}}$, at a reference point for a clinical field f_{clin} of beam quality Q_{clin} and in the absence of the chamber is given by

$$D_{w, Q_{clin}}^{f_{clin}} = D_{w, Q_{msr}}^{f_{msr}} \cdot \Omega_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}} \quad (1.10)$$

$$D_{w, Q_{clin}}^{f_{clin}} = M_{Q_{msr}}^{f_{msr}} \cdot N_{D, w, Q_0} \cdot k_{Q, Q_0} \cdot k_{Q_{msr}, Q}^{f_{msr}, f_{ref}} \cdot \Omega_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}} \quad (1.11)$$

where Q_{msr} is the beam quality for machine specific reference field f_{msr} .

$\Omega_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}}$ is a field factor that correlate the absorbed dose from machine-specific reference field f_{msr} to clinical field f_{clin} . It can be determined by the ratio of measured detector readings, multiplied by MC simulated correction factor $k_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}}$.

$$\Omega_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}} = \frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{msr}}^{f_{msr}}} \cdot k_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}} \quad (1.12)$$

For lung SBRT treatment, conventional reference field delivered using a linac can be achieved, therefore there is no need for the intermediate machine specific reference field. Therefore the general equation for small field (Equation 1.8) reduces to the conventional reference field formalism according to TRS-398 (Equation 1.6). The only correction needed will be the required field factor $\Omega_{Q_{clin}, Q_{ref}}^{f_{clin}, f_{ref}}$ that converts the absorbed dose to water in f_{ref} to the absorbed dose to water in small field in f_{clin} (Figure 1.19).

$$D_{w, Q_{clin}}^{f_{clin}} = M_{Q_{ref}}^{f_{ref}} \cdot N_{D, w, Q_0} \cdot k_{Q, Q_0} \cdot \Omega_{Q_{clin}, Q_{ref}}^{f_{clin}, f_{ref}} \quad (1.13)$$

$$\Omega_{Q_{clin}, Q_{ref}}^{f_{clin}, f_{ref}} = \frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{ref}}^{f_{ref}}} \cdot \frac{D_{w, Q_{clin}}^{f_{clin}} / M_{Q_{clin}}^{f_{clin}}}{D_{w, Q}^{f_{ref}} / M_Q^{f_{ref}}} \quad (1.14)$$

For linacs, conventional reference fields can be achieved

Figure 1.19 Small field dosimetry formalism for linac that are able to achieve conventional reference field [1]

Bassinet *et al* [8] presented the small field correction factors for a range of active detectors used for a 6MV Varian linear accelerator with MLC shaped fields down to $6 \times 6 \text{ mm}^2$. These correction factors for the active detectors were based on the mean experimental output factors of passive detectors such as Gafchromic EBT2 film and small LiF TLD cubes.

Correction factors can also be obtained by simulating the detectors within the medium using MC simulation as shown by Cranmer-Sargison *et al* [31] and Czarnecki *et al* [33]. However, the course of acquiring these correction factors for small field dosimetry can be tedious due to long calculation times for MC simulation. Correction factors are dependent on depth as well as the field size and detectors used. A study in the determination of correction factors for passive detectors will be useful for medical physicist during small field dosimetric measurements.

1.5 Research question and hypothesis

Lung SBRT treatment involves the delivery of ablative levels of radiation doses in a few fractions. With technological advancements in imaging and radiotherapy, lung SBRT treatments can be improved by increasing lung tumour dose levels, which improves the overall survival rates of early stage NSCLC patients. The accurate delivery of lung SBRT remains an important topic, as it is the current conventional standard of treatment for many medically inoperable lung cancer patients. This form of treatment requires highly accurate dose delivery with little room for error. Therefore, quality assurance for lung SBRT treatment plans is vital.

However, dosimetry for lung SBRT can be challenging as it often involves small field dose calculations and dosimetric measurements.

Current commercial TPS dose calculation algorithms have shown inaccuracies under LED conditions in lung medium, and most studies done compare calculated depth doses only within static inhomogeneous phantoms [51, 143, 34, 50, 117]. It is difficult to find novel aspects of TPS dose calculation accuracy under lung respiratory motion. Few studies done were only limited to the superior-inferior tumour motion [106, 104]. A controlled depth dose study of external surface motion and anterior-posterior tumour motion will contribute to the understanding of calculated dose distributions under respiratory motion.

In order to provide a realistic dose distribution benchmark, a MC model can be built to test the accuracy of commercial treatment planning system dose calculations. The accuracy of MC dose calculations have been well validated by literature and can be considered as the gold standard for dose calculations, provided that appropriate MC input parameters was correctly assigned [96, 155, 112]. A novel approach of using gamma analysis for the commissioning of a MC model was proposed.

Under small field conditions (for field size smaller than $4 \times 4 \text{ cm}^2$), dose perturbation may occur due to the presence of detectors in low lung density medium. Perturbation correction will be needed for accurate dosimetry. However, most studies found investigated perturbation correction for active detectors only [31, 33]. To date, hardly any study was done to investigate the perturbation correction required for passive detectors such as TLDs. Passive detectors are extremely useful for measurement of lung SBRT dose distributions within lung phantoms as they do not require any wired connection and are often the preferred choice of detectors for radiation dose audits [21].

These highlights three main areas of investigation, namely, MC commissioning using gamma analysis, dose calculation under heterogeneous LED conditions, and dosimetry under respiratory motion. A study on these topics could be useful for

medical physicists to determine suitable dose calculation algorithms and dosimeters for lung SBRT dosimetry, for example in lung SBRT dose audits and clinical trials.

Therefore, this thesis' hypothesis is therefore as follows:

The accurate delivery of doses for lung stereotactic body radiation therapy requires due consideration of the appropriate choice of dose calculation algorithms to be used for treatment planning calculations, as well as accurately corrected dosimetric measurements, which can be used for quality assurance exercises under the conditions of small fields, low density lung medium and motion.

1.5.1 Research plan and objective

The work presented in this thesis focuses on three main topics, namely 1) optimal small field parameterization for MC simulation, 2) studying the accuracy of dose calculation algorithms in the presence of motion in a lung phantom and 3) investigating perturbation effects for small field *in-vivo* dosimetry for a passive detector, specifically TLDs. These topics are displayed in Figure 1.20 with regions. Three primary research objectives were developed in order to study these topics and their co-dependency.

Figure 1.20 A schematic introducing the major topics to be addressed within this thesis

- (A) The determination of optimal small field parameters for MC simulations by using 1D gamma analysis. (Region A in Figure 1.20)

- (B) Depth dose study of two commercial dose calculation algorithms, namely Acuros XB dose calculation (AXB) algorithm and Anisotropic Analytical Algorithm (AAA) against MC simulation using a respiratory motion phantom. (Region B in Figure 1.20)
- (C) Lung SBRT dosimetric measurements using Gafchromic EBT3 films and MC corrected Thermoluminescent dosimeter (TLD) dose measurements. (Region C in Figure 1.20, measurement of small field doses under motion)

Chapter 2

Monte Carlo simulation parameters and modelling

MC methods have been considered as the gold standard for accurate dose calculations in radiotherapy and its accuracy has been well validated by literature [96, 155, 112, 154, 129]. MC dose calculations are able to provide a realistic dose distribution in both homogenous and inhomogeneous media. MC simulations was even used as a benchmark to test the accuracy of commercial treatment planning system dose calculations using geometries such as square fields on slab phantoms, with particular emphasis on the dose effects of heterogeneities or at interfaces [50, 23]. Unfortunately, full MC calculations require a large amount of computational resources and specific expertise to develop the MC model. The dose computation and calculation time required are also too long to be used in radiotherapy clinics. With the current development of multicore computer processing technology, MC-based treatment planning systems was recently developed to model radiation transport with sufficient accuracy so as to provide a more realistic dose distribution in inhomogeneous media.

In this chapter, the MC simulation model for photon beams of nominal energy 6 MV was built and commissioned. As discussed in the previous chapter, the full MC simulation comprises of two main aspects, modelling of each individual component within the linac treatment head, as well as the verification of the simulated dose output in standard reference conditions. We will be discussing these two aspects in the following sections.

2.1 Modelling of the components within the linac treatment head

The geometric modelling of the linac treatment head begins at the surface of the first component module (CM), the X-ray target, where it is the origin of the Z-axis ($Z = 0$ cm). The geometry model was aligned such that the Z-axis in the model was the beam central axis (CAX). The X- axis represents the cross-plane direction and Y-axis denotes the in-plane direction in the coordinate system (Figure 2.1). Table 2.1 lists the treatment head components together with the corresponding CMs that are used to model the structures in BEAMnrc.

The X-ray target, made of a high-Z material such as tungsten, is the first component in the geometry model. It simulated using SLABS CM. Simulating the correct layer thicknesses and materials is important to model realistic photon beam output. Bremsstrahlung X-rays is generated when high energy primary electron beam strikes the X-ray target. The X-ray target is considered as a semi-infinite slab as the width of the incident primary electron beam is small in comparison to the size of the target.

The primary collimator CM is CONS3R. The collimator is made of tungsten and has a conical opening in the direction of the beam. As shown in Figure 2.1, the radii of the openings of the collimator largely affects the beam's entrance and exit paths, hence, it is important to input the correct beam opening angle. The large lateral thickness of the structure was modelled to provide sufficient shielding within the treatment head.

The vacuum window is a thin window after the primary collimator and was modelled with SLABS CM. The purpose of the window is to maintain a vacuum to allow minimal impact on the radiation beam.

Flattening filter is modelled using the FLATFILT CM. It plays the greatest role in modulating the primary photon beam and is located in a carousel within the linac head. Within the carousel, it contains different flattening filters specified for different photon energies. The conical structure flattens the forward-peaked photon beam intensity such that the beam produces a relatively flat dose distribution in standard reference conditions. The geometry has a highly non-linearly shaped surface and modelling of this component have to be highly precise, as a small misalignment or incorrect shape will largely alter the beam symmetry and flatness. It is usually made of lead, tungsten, steel or aluminium alloy high-Z material.

The dose monitor ion chamber is a safety mechanism, which monitors the beam output to ensure consistency of the beam. It consists of two stacked transmission parallel plate and has a complex structure with several plastic sheets, protective covering and electrodes separated by very small distances. The monitor ion chamber is modelled using CHAMBER CM.

The mirror is modelled using MIRROR CM. It is placed at a beam opening angle, defined by the angle between the plastic mirror sheet. The mirror projects a light field of the beam setup onto the surface of the patient to enable patient positioning during treatment.

The jaws, modelled using SYNCJAWS CM, comprises of two pairs of high-Z tungsten blocks. Y jaws are in an in-plane direction and X jaws are in a cross-plane direction. The jaws are the first tertiary beam-modifying components. Any rectangular field sizes ranging between $0 \times 0 \text{ cm}^2$ to $40 \times 40 \text{ cm}^2$ can be collimated using the jaws alone. The Y jaws move about an arc, about the center at the X-ray target, with the jaw edge parallel to the radial line from the primary electron beam. The X jaws moves horizontally, pivoting so that the jaw edge is parallel to the Y jaws. Geometrical equations were used to obtain the jaw positions from the isocenter (SSD = 100 cm) level for the required field size.

The multi-leaf collimator (MLC) model used (Varian Millennium 120 leaf MLC module), consists of 60 pairs of leaves made of tungsten and aluminium, and is modelled using SYNCVMLC CM. The MLC is difficult to model as each MLC leaf is modelled differently according to the tongues, grooves and driving screw holes. In addition, the round leaf ends curvature radius and the interleaf air gap has to be set for all leaves. MLC provide conformal shaping when rectangular shaping from jaws is not required. In this thesis, MLC was retracted in the treatment head, as the jaws are sufficient to provide rectangular beam shaping.

Figure 2.1 Varian Clinac iX MC photon geometry model diagram (For illustration purposes, diagram is not to scale)

Table 2.1 Corresponding CMs used for the 6MV photon beam geometry model

Component	Photon Beam Model
X-ray target	SLABS
Primary Collimator	CONS3R
Vacuum Window	SLABS
Flattening Filter	FLATFILT
Monitor Ion Chamber	CHAMBER
Mirror	MIRROR
Secondary Collimator: Jaws	SYNCAWS
MLC	SYNVMMLC

2.2 Method of verification for MC simulated data

Although MC simulations are the gold standard for dose calculation in radiotherapy [54, 103, 81], linac beam characteristics are often unique due to variation in accelerator designs and on-site beam tuning. It is necessary, therefore, to simulate beam parameters within each linac individually to generate a clinically realistic MC simulation [74]. The report of Task Group 105 from the American Association of Physicists in Medicine

(AAPM TG-105) highlighted the importance and challenges of using MC simulations for treatment planning such as determination of beam parameters and commissioning of beam data [27].

The first fundamental step in simulating the photon beams is the accurate determination of the primary electron beam parameters. However, the actual shape and spectra of the primary electron beam are rarely known. To determine it experimentally requires a portable spectrometer with a degree of sophistication, which is not readily available to radiotherapy clinics [112, 96]. As a first approximation, the primary electron beam is modelled as a source with mono-energetic spectra and gaussian spatial distribution by the mean electron energy and the radial distribution (which is described by the full width at half maximum (FWHM)). Despite the irregularity of the actual shape of the primary electron beam, a Gaussian model FWHM provides a good initial approximation [112]. Several studies used different procedures, such as trial and error methods, for optimising the electron parameters [74, 19, 130, 77]. Therefore, determination of the primary electron beam parameters can prove to be a difficult task for users as it is tedious and time-consuming.

Commissioning procedure for a MC model is usually conducted by comparing a set of measurements at a single reference field size (generally $10 \times 10 \text{ cm}^2$) to the corresponding MC simulation [112, 111]. The commissioning procedure was not standardised and several studies have reported several different procedures for analysing the accuracy of the simulated profiles [112, 140, 24, 74, 27, 19, 130].

2.2.1 Direct dose difference

The most basic evaluation method during the commissioning for a MC model is to take the direct dose difference between the two datasets and check if each simulated and measured point agrees within a specified tolerance value. This works well in regions where there is a low dose profile gradient. However, in dose profiles with regions of high dose gradients (especially in lung SBRT treatment fields), this evaluation method can report a large dose error due to a small misalignment, which may not be clinically significant. The dose difference is calculated using equation 2.2

2.2.2 Distance-to-agreement

The distance-to-agreement (DTA) is the closest distance between the reference point and the evaluated point in the distribution that has the same dose value. In contrast to the dose difference test, DTA test response well in high dose gradient regions as high dose gradient regions can be interpreted as the spatial offset between the two

Figure 2.2 Diagrammatic representation of the concept of dose difference and DTA. In this example, the reference dose profile is shown as the black line and the evaluated as the red line.

distributions. The offset can regard distance misalignment as experimental setup error instead of inaccurate dose distribution. DTA is very sensitive in low dose gradient regions. As shown in Figure 2.2, it is possible for a large misalignment between two points with same dose in a low dose region. This will lead to a clinically significant error even though the dose difference in the region may be small.

2.2.3 Gamma analysis

As outlined above, the dose difference and DTA tests have advantages and disadvantages associated with them. The dose difference test is useful in low dose gradient regions but overly sensitive in high dose gradient, whereas the opposite is true for DTA. Therefore the ‘gamma analysis’ was developed as a combination of the two tests, such that one can test both the dose and spatial agreement of two distributions. The user sets a pass/fail criterion for the dose difference and DTA. The gamma analysis using gamma index (γ) evaluation is the standard technique used to evaluate measured distributions from commercial radiation detector systems against the dose distribution calculated from commercial TPS. It combines the dose difference criterion and the distance difference criterion and to calculate a dimensionless metric for each point for the evaluated distribution. Simulated data points will be considered a pass if $\gamma \leq 1.0$ and fail if $\gamma > 1.0$. The gamma index is calculated for each reference point in the reference dose distribution against each point in the evaluated distribution (Figure 2.3). To summarise, the two comparisons are:

1. the distance between reference to evaluated point: $\Delta r(\mathbf{r}_R, \mathbf{r}_E)$
2. the dose difference between the reference and evaluated points; $\Delta D(\mathbf{r}_R, \mathbf{r}_E)$

For each point in the evaluated distribution, the gamma index is calculated using the following equation:

$$\Gamma(\mathbf{r}_R, \mathbf{r}_E) = \sqrt{\frac{\Delta r^2(\mathbf{r}_R, \mathbf{r}_E)}{\delta r^2} + \frac{\Delta D^2(\mathbf{r}_R, \mathbf{r}_E)}{\delta D^2}} \quad (2.1)$$

where δr and δD is the distance and dose difference criterion respectively which is determined by the user.

Figure 2.3 Diagrammatic representation of the gamma index method in one dimension adapted from Low et al [88]. The y-axis represents Dose, D , and the x-axis represents Distance, r . In this schematic, the reference point is illustrated by the blue circle at the origin of the graph. The black line represents the evaluated dose profile with individual points represented by red circles.

An array of gamma index values for all of the points of the reference dose distribution can be constructed. It is then common to report the percentage of points that passes with $\gamma < 1$. This is known as the gamma index passing rate. The standard nomenclature used to report the passing criteria is in the format $\gamma_{\delta D(\%)/\delta r(mm)}$. For example, 1%/1mm gamma criteria will be denoted as $\gamma_{1.0/1.0}$.

The most common passing criteria being used is $\gamma_{3.0/3.0}$, which was originally recommended in the work by Low et al [88]. The gamma index was originally designed to compare measured dose distributions in water tank against calculated dose

distributions from TPS algorithm. The choice for the appropriate passing criteria were investigated and discussed further throughout this chapter. The potential limitations of the gamma index will also be discussed further.

2.3 Obtaining optimal beam parameters for the MC model

The aim of this chapter is to provide an evaluation approach that will clearly identify the optimal parameters for MC simulation and investigate the optimal parameters differences between conventional and small field sizes. A generic set of optimal primary electron beam parameters for all field sizes will be also determined. The current study uses gamma (γ) analysis to evaluate the match of the simulated depth doses and lateral profiles to measured data of 6 MV photon beam generated from Varian Clinac iX linac (Varian Medical Systems, Palo Alto, USA). The study will further extend to small fields as well. Measured data were obtained by cross-checking measurements using EDGE Detector (Sun Nuclear Corporation, Melbourne, FL) and PTW Diode SRS (PTW-60018) in a Sun Nuclear 3D water tank. Different mean energies and FWHM of the primary electron beam were simulated. Comparisons were done for conventional field sizes $10 \times 10 \text{ cm}^2$, $6 \times 6 \text{ cm}^2$ and $4 \times 4 \text{ cm}^2$ and small field sizes $2 \times 2 \text{ cm}^2$, $1 \times 1 \text{ cm}^2$ and $0.5 \times 0.5 \text{ cm}^2$ to measured data. Measured data for conventional field sizes were also cross-checked with GBD provided from manufacturer.

2.3.1 Experimental methods

Calculations were done using BEAMnrc and DOSXYZnrc user codes (National Research Council of Canada, Ottawa, ON) on a Linux operating system computer cluster with 12 Intel Xeon central processing units (CPUs). The CPUs operated at processing speeds of 2.67 or 3.4 GHz. The total RAM available to the cluster was 40 GB.

BEAMnrc was used to generate the phase space file for field size $10 \times 10 \text{ cm}^2$ (100cm Source-to-Surface Distance (SSD)). The linac model, Varian Clinac iX (Varian Medical Systems Inc., Palo Alto, CA), was defined by specifications provided by the Varian Monte Carlo data package: High Energy Accelerator [144]. 6 MV photon beam of Varian Clinac iX linac head was simulated. The primary electron beam was assumed to have a Gaussian radial intensity distribution (source number 19 in BEAMnrc).

EGSnrc based DOSXYZnrc code was used to calculate the dose in the region of interest. Water in the EGSnrc materials library, H20700ICRU, was assigned to simulate the water medium. Voxel size was set as $0.1 \times 0.1 \times 0.1 \text{ cm}^3$, which was consistent with the smallest calculation grid size in clinical Eclipse TPS.

For each of the field sizes, depth dose simulations of 5 different incident mean electron energies ($\epsilon = 5.8 - 6.2 \text{ MeV}$) were done. Normalisation was done at 1.6 cm depth, which is the depth at maximum dose (D_{max}). Comparisons of the varying FWHM parameter (FWHM = 0.1 - 0.5 cm) were done for the lateral dose profile along the x-axis at D_{max} , depth 1.6cm as well.

2.3.2 Initial reference simulated data

Initial reference primary electron beam parameters and initial reference jaw thickness were obtained by interpolating from the relationship of several ϵ and FWHM at reference field size $10 \times 10 \text{ cm}^2$ [24, 111]. These initial reference parameters were obtained by evaluating the best-fit curve of the direct dose difference between the average percentage dose difference between all normalised simulated and measured data, defined as follows:

$$\frac{\sum_n \frac{D_{MC} - D_{measureddata}}{D_{measureddata}} \%}{n} \quad (2.2)$$

where D_{MC} and $D_{measureddata}$ represent MC simulation and measured data depth dose measurement respectively; n is the total number of measured points [24, 83].

It was found that $\epsilon = 6.1 \text{ MeV}$ and FWHM = 0.3 cm primary electron beam was optimal for reference field size $10 \times 10 \text{ cm}^2$ based on the average dose difference relationship from Equation 2.2. Variations of other parameters for different field sizes will be referenced to $\epsilon = 6.1 \text{ MeV}$ and FWHM = 0.3 cm. Each time, only one parameter was varied while the other parameter was kept constant for comparison. It was also found that using reference field size $10 \times 10 \text{ cm}^2$ parameters resulted in an underestimation of lateral profile doses, especially in the OOF region[83, 3].

2.3.3 γ analysis

MC simulated lateral and depth dose profiles were analysed using 1D dose profile γ analysis as presented by Low *et al* [88]. In order to clearly identify the optimal parameters, γ criteria was varied between 0.3% to 2.0% dose difference and 0.3mm to 2.0mm distance-to-agreement (DTA) ($\gamma_{0.3\%/0.3mm} - \gamma_{2.0\%/2.0mm}$). Determination of the optimal parameters for each field size was based on the highest percentage of passes using the most stringent γ criteria, $\gamma_{0.3/0.3}$. Parameters with 95% passes using

Figure 2.4 The relationship between ϵ and average dose difference against measured data. The best fit curve was fitted with a second order polynomial function, with $R^2 = 0.985$. 6.1MeV was chosen as the initial reference electron energy with the lowest percentage dose difference.

the respective γ criteria were identified. Analysis of the depth dose and lateral profile for field size $10 \times 10 \text{ cm}^2$ by varying only one γ criterion (dose difference or DTA) was also included in the study.

Figure 2.5 A flow chart summarizing the steps involved in the determination of the optimal parameters for each field size in the MC simulation

2.4 Results

2.4.1 Depth dose profile comparison with varying ε and FWHM

By applying γ analysis, a clear trend was observed which allows the determination of optimal ε for each field size. For all field sizes, $\gamma_{2.0/2.0}$ and $\gamma_{1.0/1.0}$ was unable to differentiate the optimal ε as most of the data points passes the acceptance criteria with 95% confidence. Results show that optimal mean electron energy for field sizes larger than $2 \times 2 \text{ cm}^2$ was $\varepsilon = 6.1 \text{ MeV}$, and for field sizes smaller than $2 \times 2 \text{ cm}^2$, variation in ε was not significant.

Variations in ε show greater dependance on the depth dose profile as compared to FWHM variations (Figure 2.7 and Figure 2.10). Results were consistent to the study done by Chetty *et al* [27], where it was reported that incident electron radial distribution have little observable effect on central-axis depth dose.

2.4.2 Lateral profile comparison with varying ε and varying FWHM

Variations in ε show weak dependance on the lateral dose profile as compared to FWHM variations (Figure 2.8). Most of the data points due to the variations in ε passes the acceptance criteria with 95% confidence. For field size $10 \times 10 \text{ cm}^2$, percentage of passes for the data points decreases as there were larger statistical uncertainties in simulating larger field sizes.

It was found from previous studies that the FWHM parameter affects the central and penumbra regions, particularly the 'horns' of the lateral profile [27, 96, 112]. Since variations were localised, comparison of the optimal FWHM parameters was limited to those regions. Variations were as not as distinct as the variations in the ε γ analysis. Results show that optimal FWHM parameter decreases with smaller field sizes. At extreme small field size $0.5 \times 0.5 \text{ cm}^2$, primary electron beam distribution is simulated closer to a point source with optimal FWHM = 0.1. Optimal FWHM = 0.3 for field sizes greater than $4 \times 4 \text{ cm}^2$. Percentage of passes for FWHM = 0.3 within 95% confidence for field size $0.5 \times 0.5 \text{ cm}^2$ and $1 \times 1 \text{ cm}^2$ for $\gamma_{0.4/0.4}$

2.4.3 Summary of results

2.4.4 Discussions

Initially, doses were compared directly, with an acceptance tolerance of 2% for the difference between measured data and calculated doses within the central region of the beam [152, 153]. However, it was found that in high-dose gradient regions such

Table 2.2 Optimal Parameters for MC simulation of measured data based on 1D γ analysis

Optimal Parameters for MC simulation		
Field Size	Mean Electron Energy/ ϵ (MeV)	FWHM (cm)
FS0.5	5.8 - 6.1	0.1
FS1	5.9 - 6.2	0.1
FS2	6.1	0.1
FS4	6.1	0.3
FS6	6.1	0.3
FS10	6.1	0.3

as penumbra and the OOF regions, a small spatial error might result in a large dose difference between measured data and calculation [88, 68, 62]. Therefore, it is difficult to compare the trend using a simple dose-difference distribution across different field sizes. Therefore, the concept of a distance-to-agreement (DTA) distribution in the form of γ analysis was used to determine the acceptability of the dose calculation. γ analysis provides an excellent measure of the calculation quality in regions of high-dose gradient such as penumbra and OOF doses, and it was extremely sensitive in regions of a low-dose gradient, where a slight difference in dose yields a large γ index [62]. Van Dyk *et al* [43] provided guidelines and recommended γ criterion to be $\gamma_{3\%/4mm}$ for the evaluation of TPS algorithms for photon beams. Other recommendations include $\gamma_{3\%/3mm}$ and $\gamma_{2\%/2mm}$ as proposed by Fogliata *et al* in her study [50]. As a gold standard calculation algorithm, MC simulation will often be used as a dose calculation benchmark. Hence, the extremely stringent γ passing criteria used in this study will be robust to determine the optimal parameters for a good simulation model.

It was shown from the comparison in Figure 2.10, both the depth dose and lateral dose profiles are more sensitive to dose differences than to DTA. This is because the beam profiles simulated generally have small disagreements in DTA from the measured data, unless there is a significant experimental setup error or calculation error. As lateral profiles have higher dose gradients than depth dose profiles, lateral profiles will be more sensitive to DTA than dose profiles as shown in Figure 2.10d.

Previous studies suggested that the determination of the optimal parameters was conducted by comparing a small set of measurements of a single field size generally $10 \times 10 \text{ cm}^2$ or $20 \times 20 \text{ cm}^2$ to their correspondent simulations [64, 140, 24]. By

employing only a single wide field, the photon fluence can be accurately matched, avoiding collimator scatter contributions. This method of determination will be acceptable provided that the exact geometry and composition of the linac are well known and the primary radiation source is well characterised [130]. However, an independent determination of optimal parameters for field sizes smaller than $10 \times 10 \text{ cm}^2$ will be required for high accuracy simulation [112]. Contrary to previous studies, each field size has its unique set of optimal parameters for MC simulations of photon beams and should be tuned individually to pass a stringent γ criteria of $\gamma_{0.3/0.3}$, especially for field sizes smaller than $2 \times 2 \text{ cm}^2$ [112].

However, imposing a stringent criteria and simulating unique parameters for different field sizes may not be practical, as users may wish to use just one set of parameters for all field sizes for MC simulation work. Based on this study, optimal set of parameters for all field sizes can be derived by choosing parameters that have at least a 95% percentage passes with $\gamma_{0.5/0.5}$ criteria (Table 2.3).

Table 2.3 Optimal Parameters for all field sizes based on 95% passes with $\gamma_{0.5/0.5}$ criteria

Optimal parameters for MC simulation for all field sizes	
Mean Electron Energy/ $\epsilon(\text{MeV})$	FWHM (cm)
6.1	0.3

The Monte Carlo model was simulated using a large number of primary histories (1×10^9) together with several variance reduction techniques to minimise the statistical uncertainty. The combined standard uncertainty of all of the Monte Carlo calculation was reported as less than 1% as a conservative estimation, which is consistent to the study done by Renner *et al* [119].

As the simulated field size decreases, statistical uncertainty decreases as the same number of primary histories is being simulated in a smaller region of interest. Results show that for field size $0.5 \times 0.5 \text{ cm}^2$, statistical uncertainty for the associated simulated point can be as low as 0.12%.

One may argue that by having tight gamma criteria such as $\gamma_{0.3\%/0.3\text{mm}}$, it is unrealistic as it will make a large percentage of points fail. However, the main aim of the 1D gamma analysis study is to sieve out profiles with the highest number of points with percentage of passes. An example can be seen from the analysis of PDD for field size $0.5 \times 0.5 \text{ cm}^2$ in Figure 2.6a, where most of the parameters have 95% of percentage passes. As compared to the $\gamma_{0.3\%/0.3\text{mm}}$ analysis of PDD for field size $10 \times 10 \text{ cm}^2$ in Figure 2.6f, where most of the parameters have points that have less

than 10% of percentage passes, but 6.1MeV shows the highest amount of percentage passes. This analysis method will allow us to differentiate the appropriate parameters for each specific field size.

2.5 Conclusion

Determination of the primary electron beam parameters for 6 different field sizes was performed. Incident mean electron energy, and FWHM was unique for different field sizes for highest accuracy to measured data. Incident mean electron energy shows little dependence on lateral dose profile, and FWHM shows little dependance on depth doses. A set of optimal parameters for all field sizes was determined based on the result using 95% passes with $\gamma_{0.5/0.5}$ criteria.

(a) Small field size $0.5 \times 0.5 \text{ cm}^2$ Depth Dose (b) Small field size $1 \times 1 \text{ cm}^2$ Depth Dose (c) Small field size $2 \times 2 \text{ cm}^2$ Depth Dose (d) Conventional field size $4 \times 4 \text{ cm}^2$ Depth Dose (e) Conventional field size $6 \times 6 \text{ cm}^2$ Depth Dose (f) Conventional field size $10 \times 10 \text{ cm}^2$ Depth Dose

Figure 2.6 γ analysis of the depth dose profile with varying ϵ for small and conventional field sizes.

(a) Small field size $0.5 \times 0.5 \text{ cm}^2$ Depth Dose

(b) Small field size $1 \times 1 \text{ cm}^2$ Depth Dose

(c) Small field size $2 \times 2 \text{ cm}^2$ Depth Dose

(d) Conventional field size $4 \times 4 \text{ cm}^2$ Depth Dose

(e) Conventional field size $6 \times 6 \text{ cm}^2$ Depth Dose

(f) Conventional field size $10 \times 10 \text{ cm}^2$ Depth Dose

Figure 2.7 γ analysis of the depth dose profile with varying FWHM for small and conventional field sizes.

(a) Small field size $0.5 \times 0.5 \text{ cm}^2$ Lateral Dose (b) Small field size $1 \times 1 \text{ cm}^2$ Lateral Dose (c) Small field size $2 \times 2 \text{ cm}^2$ Lateral Dose (d) Conventional field size $4 \times 4 \text{ cm}^2$ Lateral Dose (e) Conventional field size $6 \times 6 \text{ cm}^2$ Lateral Dose (f) Conventional field size $10 \times 10 \text{ cm}^2$ Lateral Dose

Figure 2.8 γ analysis of the lateral dose profile with varying ϵ for small and conventional field sizes

(a) Small field size $0.5 \times 0.5 \text{ cm}^2$ Lateral Dose (b) Small field size $1 \times 1 \text{ cm}^2$ Lateral Dose (c) Small field size $2 \times 2 \text{ cm}^2$ Lateral Dose (d) Conventional field size $4 \times 4 \text{ cm}^2$ Lateral Dose (e) Conventional field size $6 \times 6 \text{ cm}^2$ Lateral Dose (f) Conventional field size $10 \times 10 \text{ cm}^2$ Lateral Dose**Figure 2.9** γ analysis of the lateral dose profile with varying FWHM for small and conventional field sizes

(a) Conventional field size $10 \times 10 \text{ cm}^2$ depth dose with varying dose difference and constant 1.0 mm DTA

(b) Conventional field size $10 \times 10 \text{ cm}^2$ depth dose with varying DTA and constant 1.0 mm dose difference

(c) Conventional field size $10 \times 10 \text{ cm}^2$ lateral profile dose with varying dose difference and constant 1.0 mm DTA

(d) Conventional field size $10 \times 10 \text{ cm}^2$ lateral profile dose with varying DTA and constant 1.0 mm dose difference

Figure 2.10 γ analysis of the depth dose and lateral profile by varying only dose difference or DTA for the γ criteria.

Chapter 3

Depth dose study between clinical TPS algorithms against MC simulation

The calculation of photon field dose distributions within low-density inhomogeneity in the presence of motion artefact such as blurring can be a challenge for algorithms in clinical treatment planning systems (TPS). Such clinical cases with low-density inhomogeneity and motion artefacts are common for lung stereotactic body radiation therapy (SBRT) treatments. During lung SBRT planning, dose calculations are performed on average intensity projected (AIP) images derived from four-dimensional computed tomography (4D-CT) images [69]. This accounts for any tumour motion due to respiration during treatment. As high doses are being delivered in few fractions during lung SBRT, dosimetric precision for dose calculations are relevant and very important.

Most inhomogeneity studies compare calculated depth doses within static inhomogeneous phantoms [51, 143, 34, 50, 117, 49, 101, 48, 41]. Other respiratory motion studies done were only limited to the superior-inferior tumour motion [106, 104]. A controlled depth dose study of external surface motion and anterior-posterior tumour motion will contribute to the understanding of calculated dose distributions under respiratory motion. Furthermore, small photon fields such as $2 \times 2 \text{ cm}^2$ can impose problems due to LED and changes in energy spectrum. Under the presence of motion artefacts, small photon fields and low-density medium such as lung, the limitations of clinical dose calculation algorithms can be tested.

Previous inhomogeneous phantom studies compared the performance of Anisotropic Analytical Algorithm (AAA) and Acuros XB dose calculation (AXB) algorithm (both from Varian Medical Systems, Inc., Palo Alto, CA), version 10.0.28, against MC

simulation. It was found that the accuracy of the algorithms depends on the photon beam energy, field size and density of the medium [51, 143, 34, 50, 117, 49, 101]. The recent release of an improved version for both AAA and AXB algorithms, version 13.6.23, highlighted key changes in the dose prediction performance, especially in how each algorithm account for inhomogeneities.

Flattening filter-free (FFF) beams are also often used during lung SBRT treatments, as they are able to deliver approximately four times higher dose rate, reducing delivery time. Previous studies claimed that FFF beams can produce dosimetrically equivalent plans when compared to conventional flattened beams [151, 162, 22, 7]. The inclusion of FFF beams in the extreme inhomogeneous dosimetric study will test the robustness of this claim.

To our knowledge thus far, no study was done for the depth dose differences between a full MC simulation, 6X FFF predicted doses and two versions of AAA and AXB algorithm under the presence of a moving tumour and chest wall.

This study will systematically investigate the comparison of MC calculated depth doses against 6X FFF beam depth doses and depth doses calculated using AAA and AXB algorithm, versions 10.0.28 and 13.6.23. Dose calculations were done on the QUASAR respiratory motion phantom (Modus Medical Devices Inc., London, ON). 6X FFF beam will be obtained from MC simulation based on a provided phase space file [30]. EGSnrc simulation package developed by the National Research Council of Canada was used for the full MC simulation [80]. MC simulated data were validated against measurement under homogenous condition in water, as well as inhomogeneous condition in a static lung phantom. MC was used as a benchmark for depth dose comparison in the respiratory phantom for this study. Anterior and lateral beam depth doses were compared to investigate depth dose differences for both superior-inferior and anterior-posterior motion of the tumour.

3.1 Experimental methods

3.1.1 Full MC simulation model

MC simulations were done using EGSnrc V4-r2.4.0, consisting of BEAMnrc and DOSXYZnrc user codes developed by the National Research Council of Canada. A Linux-based computer cluster which comprised of 12 Intel Xeon central processing units (CPUs) with processing speeds of 2.67 or 3.4 GHz and a total of 40 GB of RAM was used for the calculations.

Phase space files for a Varian Clinac iX linear accelerator treatment head (Varian Medical Systems Inc., Palo Alto, CA) of 6 MV photon beam for field sizes 2×2

cm^2 , $4 \times 4 cm^2$ and $10 \times 10 cm^2$ (100 cm SSD) were generated using BEAMnrc. The primary electron beam was 6.1 MeV with focal spot width of 0.3 cm at full width at half maximum. The energy spectrum of the bremsstrahlung beam was matched with the phase space file released by International Atomic Energy Agency (IAEA) phase space database [20]. Computational efficiency was increased using variance reduction technique such as directional bremsstrahlung splitting (DBS). The global photon and electron cut-off energies were 0.01 MeV and 0.7 MeV respectively. The splitting number was set to 1000 and the electron splitting was performed in the lower layers of the flattening filter as recommended [121]. Range rejection was turned on with varying ECUTRR and was considered for electrons with energy less than 2 MeV ($ESAVE_GLOBAL = 2$) [66]. EXACT was chosen for the boundary crossing and electron step algorithms. Both BEAMnrc and DOSXYZnrc usercodes used similar settings. All DOSXYZnrc dose calculations were performed using 1×10^{10} histories with statistical uncertainties below 1%. EGSnrc calculation voxel size was set similar to Eclipse treatment planning system (TPS) (Varian Medical Systems, Palo Alto, CA) calculation grid size of 0.25 cm. The uniform calculation grid size across considered algorithms was an improvement from the previous inhomogeneity dose modelling done by Zvolanek *et al* [165].

3.1.2 6X flattening filter-free beam

6X FFF photon beams were simulated based on the phase space files provided from Varian MC research team for TrueBeam Linac on <https://myvarian.com> [30]. The phase space files provided were written out above the jaws for TrueBeam, therefore, an additional jaws component module was added to create field size specific phase space file using BEAMnrc. Parameters for the calculation of doses using DOSXYZnrc were similar to the full MC model.

3.1.3 AAA and AXB algorithm

AAA and AXB algorithm were configured for Varian Clinac iX using Eclipse TPS. Calculations were done across AAA version 10.0.28 (AAAv10.0), AAA version 13.6.23 (AAAv13.6), AXB algorithm version 10.0.28 (AXBv10.0) and AXB algorithm version 13.6.23 (AXBv13.6) against MC simulation. Dose to medium (D_m) reporting mode was chosen for AXBv10.0 and AXBv13.6, similar to previous studies [61, 18, 117, 79, 49]. The only reporting mode for AAAv10.0 and AAAv13.6 is dose to water (D_w). The differences between these two reporting modes will be further discussed. From here on forth, the terms “AAA” and “AXB algorithm” will be used to

describe both versions of their algorithms (i.e. AAAv10.0, AAAv13.6 and AXBv10.0, AXBv13.6 respectively). Depth dose profiles for field sizes $2 \times 2 \text{ cm}^2$, $4 \times 4 \text{ cm}^2$ and $10 \times 10 \text{ cm}^2$ were calculated with calculation grid size of 0.25 cm.

3.1.4 CT calibration curve

The CT calibration curve in Eclipse TPS was calibrated using a multi-plugin CIRS[®] phantom 062 (Computerized Imaging Reference Systems, Norfolk, VA, USA). Both AXBv10.0 and AXBv13.6 refers to the same CT calibration curve for the HU value to mass density relationship in each voxel of the image grid. EGSnrc have a default CT calibration curve. For consistency, the default EGSnrc calibration curve was configured to be the same CT calibration curve as AXBv10.0 and AXBv13.6 simulations. Cross sectional data for air, lung, adipose, muscle, cartilage and bone were generated according to AXBv10.0 material cross section data using PEGS in EGSnrc [80, 145]. The HU value to relative electron density calibration curve for AAA calculations was also calibrated using the same phantom.

Figure 3.1 CT calibration curve for AXBv10.0 and AXBv13.6 and the default EGSnrc CT calibration curve.

Mass density to material mapping

In order to perform dose calculations, AXB algorithm and EGSnrc require the macroscopic cross section for each material in its computational grid [18]. The algorithms, however, assign the material based on the HU value of each voxel differently. AXBv13.6 allows the overlapping of mass density ranges in CT to material mapping [146] (Figure 3.2). In contrast, EGSnrc and AXBv10.0 do not allow the overlapping of mass density ranges. The mass density to material mapping for EGSnrc was configured to be the same as AXBv10.0, except with the addition of air material in EGSnrc. The air material added was identical to AXBv13.6. This was similar to previous work done by Ojala *et al.* and Onizuka *et al.* [108, 102].

Figure 3.2 Material assignment for EGSnrc MC simulations, AXBv10.0 and AXBv13.6.

3.1.5 Variations in HU profile under static and respiratory motion

(a) HU profile of the superior-inferior tumour motion in the anterior beam direction

(b) HU profile of anterior-posterior tumour motion in the lateral beam direction

Figure 3.3 A comparison of the static tumour CT image (solid line) with the AIP CT image (dotted line).

The moving lung tumour and chest wall causes motion artefact such as blurring, forming HU value gradients. This causes several regions of the lung phantom to exhibit contrasting HU values relative to the denser tumour polyoxymethylene sphere [59], as shown in Figure 3.3a and Figure 3.3b. The AIP HU profile in the anterior beam direction shows less CT contrast and greater similarities to static HU profile. HU profile differences could only be observed for the superior-inferior tumour motion, with only a minor difference between HU value gradients.

In the lateral beam direction, the moving chest wall had a HU value gradient between -900 to -100 HU, representing the highest CT contrast due to motion. The moving chest wall and tumour dimensions were enlarged by twice the motion amplitude in the lateral beam AIP HU profile, with ill-defined boundaries between different mediums.

Validation of MC doses under homogeneity condition

MC depth dose profiles for field sizes $2 \times 2 \text{ cm}^2$, $4 \times 4 \text{ cm}^2$ and $10 \times 10 \text{ cm}^2$ were re-verified against an IBA PFD Photon Diode Dosimetry detector (IBA/Scanditronix Medical AB, Uppsala, Sweden) in a water tank. The experimental setup comprises of a $30 \times 30 \times 30 \text{ cm}^3$ water tank phantom placed on the patient couch with the linear accelerator gantry at 270° . Correspondingly, water phantom was simulated using H2O700ICRU with a voxel size of $0.25 \times 0.25 \times 0.25 \text{ cm}^3$ in DOSXYZnrc. MC simulated data agreed with measured data and passed gamma analysis with passing criteria of 1%/1mm for the required field sizes in this study [88] (Figure 3.4).

Figure 3.4 Verification of MC simulated PDD in water for field sizes $2 \times 2 \text{ cm}^2$, $4 \times 4 \text{ cm}^2$ and $10 \times 10 \text{ cm}^2$

Validation of MC calculated doses under inhomogeneous condition

MC depth dose profiles for field sizes $2 \times 2 \text{ cm}^2$, $4 \times 4 \text{ cm}^2$ and $10 \times 10 \text{ cm}^2$ were validated under inhomogeneous condition against film dosimetry. Phantom measurements

were done using GafchromicTM External Beam Therapy 3(EBT3) films (International Specialty Products, Wayne, New Jersey). Absolute dosimetry was obtained by calibration of the required films to a known dose distribution during each measurement session. For each measurement position, the average data from two pieces of films was evaluated to increase the accuracy of the measured plane [141, 106]. Films were scanned at 75 dots per inch (dpi) using Epson Expression 11000XL scanner (Seiko Epson Corporation, Nagano, Japan) at 15 hours after the irradiation. Evaluation of film measurements was done using SNC Patient software (version 6.2.1, Sun Nuclear Corp., Melbourne, FL, USA).

Figure 4.8a shows the experimental setup of the static lung phantom used. It consists of composite cork slabs sandwiched between 4 cm thick CIRS plastic water (Computerized Imaging Reference Systems Plastic Water Diagnostic Therapy, Norfolk, VA, USA) above and below. 14 pieces of 1 cm thick, composite cork slabs, with density 0.27 g/cm^3 were used. 14 film measurement positions were done along the central axis at 1.5 cm, 2 cm, 3 cm, 4 cm, 6 cm, 8 cm, 11 cm, 14 cm, 16 cm, 18 cm, 18.5 cm, 19 cm, 20 cm and 21 cm below the surface. Only one piece of film was placed at each position along the central axis per measurement, so as to prevent overshadowing. The lung phantom was simulated in MC according to the weight fractions of the elements within the materials, as reported by Ramaseshan et al [116] and Chang et al [23]. Calculation voxel size was $0.25 \times 0.25 \times 0.25 \text{ cm}^3$. Results show that MC simulated data agreed with film measured data to within 2% tolerance for the field sizes investigated (Figure 4.8b).

3.1.6 QUASAR respiratory motion phantom setup

The technical specifications of the QUASAR respiratory motion phantom have been included in Appendix A. The respiratory motion phantom was configured to 1 cm amplitude sinusoidal respiratory motion, with a breathing frequency of 0.25 Hz. This corresponds to an average patient breathing amplitude and rate of 15 breaths/min [53]. The cedar wood lung insert holds a 3 cm diameter polyoxymethylene sphere, which simulates the target lung tumour. (Figure 3.6).

Dental modelling wax was moulded to the lung phantom insert to create a 4 cm chest wall (Figure 3.6). The dental modelling wax used (Metrowax, Metrodent Limited, United Kingdom) is a tissue equivalent material that allows the simulation of the buildup dose before the cedar wood lung phantom insert [163]. Table 3.1 shows the density and material composition of the QUASAR lung insert materials.

(a) Inhomogeneous lung phantom used

(b) Film measurement positions

Figure 3.5 Verification of MC simulated depth dose in inhomogeneous lung phantom using film dosimetry at various measurement depth for field sizes $2 \times 2 \text{ cm}^2$, $4 \times 4 \text{ cm}^2$ and $10 \times 10 \text{ cm}^2$

Figure 3.6 The QUASAR Lung Insert

Table 3.1 QUASAR lung insert materials

	Density (g/cm^3)	Material Composition
Metrowax	0.85	Paraffin and Microcrystalline Waxes [94]
Polyoxymethylene sphere	1.41	$(CH_2O)_n$ [26]
Cedar Wood	0.26	6% H, 43% O, 50% C [105]

3.1.7 Image acquisition

4D-CT images were acquired using a single-source multi-detector CT scanner (Siemens Somatom Definition AS, Siemens Medical Solutions, Forchheim, Germany). The 4D-CT images using 1 mm slice thickness, 0.5 second per rotation and 0.1 pitch was obtained to construct the AIP images of the phantom. AIP images were obtained by averaging the ten phases of gated 4D-CT images based on the helical Computed Tomography (CT) images [133]. These phases were obtained from the respiratory signal using the real-time position management optical system of the 4D-CT (RPM; Varian Medical Systems, Palo Alto, CA).

Figure 3.7 Experimental configuration of QUASAR respiratory motion phantom

3.1.8 Beam delivery and normalisation

6 MV photon beam was planned orthogonally at 100 cm fixed SSD setup when the lung insert is in the equilibrium position. The anterior beam was delivered orthogonal to the superior-inferior motion. The lateral beam was delivered in the direction of anterior-posterior motion with the moving chest wall, as shown in Figure 3.7.

Figure 3.8 Normalisation point was defined at 3 cm depth within the static chest wall for the anterior beam and in the middle of moving chest wall at 3 cm depth for the lateral beam.

For anterior beams, central axis depth dose profiles were normalised to the dose at 3 cm depth within the static chest wall (Figure 3.8). For lateral beams, central axis depth dose profiles were normalised to the dose at 3 cm depth in the middle of the moving chest wall (Figure 3.8). Normalisation was done within the chest wall instead of at the centre of the tumour because of statistical fluctuation of calculated dose to tumour from the full MC dose profile.

3.2 Results

Depth dose comparison in the anterior beam direction

Anterior depth doses are shown in Figure 3.9. By comparing with MC results, AAA tends to predict lower doses after static chest wall-lung interface for $2 \times 2 \text{ cm}^2$ and $4 \times 4 \text{ cm}^2$ and higher doses for all field sizes after lung-tissue interface. The greatest difference occurred for $10 \times 10 \text{ cm}^2$ field, where deviations up to 7% was observed.

Predictions from AXB algorithm are generally within $\pm 6\%$ against MC simulation for all field sizes. The highest dose difference from MC was about 5.2% at lung-tissue interface in $2 \times 2 \text{ cm}^2$ field by AXBv10. Deviations of AAA and AXB algorithm from MC simulations were within 7% even for the surface and buildup doses in the static chest wall. Collectively, 6X FFF doses predicted were lower and deviations found were less than 5%. Results were consistent with previous static heterogeneous studies found [50, 117, 51].

(a) $2 \times 2 \text{ cm}^2$ depth dose in anterior beam direction (b) $4 \times 4 \text{ cm}^2$ depth dose in anterior beam direction

(c) $10 \times 10 \text{ cm}^2$ depth dose in anterior beam direction

Figure 3.9 Depth dose profile of phantom in the anterior beam direction for field sizes $2 \times 2 \text{ cm}^2$, $4 \times 4 \text{ cm}^2$ and $10 \times 10 \text{ cm}^2$. The subplot below the depth dose profile represents the percentage deviation of AAA, AXB algorithm and 6X FFF predicted doses from MC simulation.

Depth dose comparison in the lateral beam direction

Figure 3.10 shows the comparison of AAA and AXB algorithm predicted depth doses against MC simulation in the lateral beam direction. The largest deviation of up to 105% from MC simulation occurs at the surface and buildup doses in $2 \times 2 \text{ cm}^2$ for AAAv10.0. For $10 \times 10 \text{ cm}^2$, AXBv10.0 had 40% deviation in the surface and buildup doses from MC simulation. 6X FFF predicted doses have approximately 20% deviation in the surface and buildup doses for all field sizes. This disagreement of surface doses occurs at the high HU value gradient region in the moving chest wall. AAAv13.6 and AXBv13.6, on the contrary, show agreement within 10% to MC simulation for all field sizes.

Excluding the surface and buildup regions, predictions by AAA algorithms show deviations in the secondary buildup regions such as the lung-tumour interfaces. This is especially so for field size $2 \times 2 \text{ cm}^2$ (Figure 3.10a). Predictions by AXB algorithm and 6X FFF doses within the phantom were in good agreement with MC simulations for all field sizes (within $\pm 4\%$) even in the presence of the anterior-posterior tumour motion.

(a) $2 \times 2 \text{ cm}^2$ depth dose in lateral beam direction (b) $4 \times 4 \text{ cm}^2$ depth dose in lateral beam direction

(c) $10 \times 10 \text{ cm}^2$ depth dose in lateral beam direction

Figure 3.10 Depth dose profiles of moving tumour for field sizes $2 \times 2 \text{ cm}^2$, $4 \times 4 \text{ cm}^2$ and $10 \times 10 \text{ cm}^2$. Beam was delivered in the lateral direction. The subplot below the depth dose profile shows the percentage deviation of AAA, AXB algorithm and 6X FFF predicted doses from MC simulation. A separate plot shows the same result for the buildup region from 0 - 2 cm depth.

Differences between the two versions of AAA and AXB algorithm

Generally, differences for AAA and AXB algorithm predicted depth doses, excluding the surface and buildup doses, agrees well with each other to within $\pm 1.5\%$ (Figure 3.11).

The surface and buildup doses for AAAv10.0 differ by up to 300% from AAAv13.6. Deviations in the surface and buildup doses between AXBv10.0 and AXBv13.6 differ by up to 60%.

(a) Differences between AAAv10.0 and AAAv13.6

(b) Differences between AXBv10.0 and AXBv13.6

Figure 3.11 Comparison of the depth dose profiles of the two versions of the algorithms for field size $2 \times 2 \text{ cm}^2$. Beam was delivered in the lateral direction. The subplot below the depth dose profile shows the percentage deviation between v10.0 with respect to depth doses predicted in v13.6. A separate plot shows their differences for 0 - 2 cm depth.

3.3 Discussions

The depth dose comparison of a full MC simulation against 6X FFF beam and dose calculations using AAA and AXB algorithm, versions 10.0.28 and 13.6.23 was investi-

gated. The full MC simulation was verified under homogeneous and inhomogeneous conditions and was taken as the benchmark for this study.

Our results have highlighted large dose differences of up to 105% for AAAv10.0 and AXBv10.0 against the MC benchmark in the lateral beam direction. These differences were found in the region of the moving chest wall. Smaller dose differences were reported for 6X FFF beam, AAAv13.6 and AXBv13.6. As discussed previously, the AIP CT of the moving chest wall has the highest HU value gradient due to the anterior-posterior motion. Previous studies claimed that anterior-posterior tumour motion was not clinically important because superior-inferior tumour motion was greater [106, 131]. However, this study shows that by having an experimental setup for anterior-posterior tumour motion, together with a 4 cm moving chest wall and 1 cm amplitude, it creates a controlled, yet challenging and demanding inhomogeneity condition. Results show that this setup was capable of differentiating the dosimetric capabilities of each algorithm and its respective versions. Anterior beam direction for superior-inferior tumour motion results was in fact, consistent with previous static inhomogeneous phantom and inhomogeneity dose modelling with respiratory motion studies [50, 117, 51, 165, 106, 104] but it was unable to differentiate the dosimetric capabilities of different algorithm versions. Dose differences, especially in the surface and buildup regions in the moving chest wall, were not identified in the static chest wall as shown in the anterior beam depth dose calculations.

The revelation of the dose prediction differences between two versions of AAA and AXB algorithm led to the investigation of the key changes made from version 10.0.28 to version 13.6.23. For AAAv13.6, it was found that a layer of water with a thickness of half CT slice spacing was no longer automatically added to the surface of the medium contour [147, 149, 150]. Subsequently, AAAv13.6 dose calculation was also corrected from producing “staircases” in the dose distribution for the buildup region, especially when very low densities were present in that region [149]. AAA calculates doses based on the CT image electron density using a water-based dose kernel. It does not consider the material’s elemental compositions. The layer of water added to the surface of the medium contour by AAAv10.0 previously was to approximate the surface and buildup doses within a water medium. This may be a good approximation for a static chest wall as shown in Figure 3.9. However, for low HU value surfaces such as the moving chest wall, this approximation results in an overestimation of doses by AAAv10.0. Changes made in AAAv13.6 has shown that surface and buildup doses have better agreement with AXBv13.6 and MC simulations (Figure 3.10).

For AXB algorithm, it was found that air medium was newly included in the material library of AXBv13.6 [146]. Furthermore, AXBv13.6 currently allows the

overlapping of mass density ranges in CT to material mapping [146, 145]. Previously, AXBv10.0 assigns voxels with low HU values as low-density lung, in contrast to the air medium assigned by AXBv13.6. The difference in material assignment will result in different surface and buildup dose predictions due to different mass collision stopping powers and mass energy-absorption coefficients calculations. The overestimation of doses by AXBv10.0 in low HU value medium was also consistent with previous results from Ojala *et al* and Onizuka *et al*. They compared the dose calculations of AXB version 10 against AXB version 11 within air medium [108, 102].

Other key changes for AAA and AXB algorithm version 13.6.23 include the improvement of image alignment and structure contouring in each voxel to calculation grid [147] and the changes made to the optimisation routine for the electron contamination model. The electron contamination model is sensitive to changes to dose deposition calculations and algorithm configuration routines [148]. The overall changes made to AAA and AXB algorithm in version 13.6.23 resulted in the agreement of dose prediction with MC benchmark as shown by the results in this study.

In addition, for small fields such as $2 \times 2 \text{ cm}^2$, AAA generally shows less agreement with AXB algorithm and MC simulations. Under small field conditions, there is an increase of LED along the central axis. AAA accounts tissue heterogeneity using the concept of radiological rectilinear scaling, where it simulates energy deposition as a function proportional to the ratio of the electron density of medium to water [145]. Rectilinear kernel scaling does not model small fields LED well as the different elemental composition of the simulated materials was disregarded. In reality, photon and electron scatter do not adhere to the rectilinear paths assumed in AAA. This is particularly pronounced in low-density lung material, where LED along the central axis becomes larger with smaller field sizes [137]. Furthermore, the previous study done by Fogliata *et al*. pointed out that AAA has an inability to model thin inhomogeneities accurately [50]. Changes in the HU value gradient (Figure 3.3) were considered as several thin layers of heterogeneity as AAA assigns different electron densities to the voxel with respect to the voxel's HU value. These discrepancies were most noticeable in dose regions when there was a presence of a high HU value gradient as predicted by AAAv10.0 in Figure 3.10. This results in the discrepancies between AAA, MC and AXB algorithm.

Andreo *et al* investigated the impact of dose calculations by using the 'water-equivalent' tissues with densities assigned from CT calibration curve (dose to water approach), to dose calculations using material compositions derived from CT number to material assignments (dose to medium approach) [4]. It was found that dose to tissue approach of using mass stopping-power ratios to water and electron fluence, are more dependent on mean excitation energies of the material, than on mass density.

Despite this, doses calculated using dose to water approach seem to agree with dose to medium approach except for high-density bone and in some cases, adipose tissue.

In terms of clinical practicality, tissue composition and its respective mean excitation energies cannot be specifically obtained for each specific patient. Furthermore, normalisation of the clinical treatment plan to the reference dosimetry is always calibrated in terms of absorbed dose to water. Since both approaches lead to small absolute dosimetric differences, dose to water approach is still the current preferred clinical dosimetric setting for dosimetric calculations.

However, Andreo *et al* claimed that dose calculations in terms of dose to medium approach should be the preferred choice, so as to avoid approximations involved in dose to medium to dose to water conversions [4]. Therefore, it is noted that for the above comparison for the depth doses between clinical TPS algorithms against MC simulation and also for subsequent parts of the thesis, dose to medium was chosen as the reporting mode for AXBv10.0, AXBv13.6 and MC dose calculations. However, for AAAv10.0 and AAAv13.6, the only available reporting mode is in dose to water approach. Nevertheless, comparison of calculated depth doses were done under relative dosimetry, the systematic uncertainty between dose to water and dose to medium were decreased during normalisation as cancelations in dose ratios decreases the differences between both approaches.

The agreement found for 6X FFF with the full MC model supports the claim that FFF beams can produce equivalent depth doses when compared to flattened beams, even in the presence of inhomogeneity, small fields and motion artefact. However, the limitation of this claim was supported only by simulation results. Further validation by conducting phantom depth dose measurements in the QUASAR respiratory motion phantom will be required to substantiate this claim.

The limitation of this study was not validating the considered depth doses by conducting dosimetric measurements. Although dosimetric measurements will prove to be a robust standard, *in-vivo* dosimetric measurements will introduce several other challenges. The low density of the lung insert, as well as small field sizes such as $2 \times 2 \text{ cm}^2$, causes potential LED. This will result in significant perturbation to the energy spectrum and the detector's dose response. Perturbation corrections must therefore be applied to correct the change in dose response. These corrections can only be accurately pre-calculated using MC simulation of the detector in the measured medium. Eventually, the accuracy of the detector's dose reading will still be dependent on the accuracy of the MC simulation. Furthermore, due to the 1cm amplitude sinusoidal motion, to determine the actual dose to the intended position of measurement will be challenging. For instance, the surface dose measurement of the moving chest wall at depth 0 cm will be difficult to obtain, as the detector will most likely measure doses

at the equilibrium of the sinusoidal motion, i.e. at depth 1cm. To date, no suitable options for conducting point dose dosimetric measurements in the presence of motion were found. Proposed solutions usually involve a certain amount of motion stochastic modelling and estimation of radiation doses delivered [164, 56].

3.4 Conclusion

6X FFF was capable of producing equivalent depth doses to MC benchmark. AAAv13.6 and AXBv13.6 show better agreement with MC benchmark than AAAv10.0 and AXBv10.0 even for extreme inhomogeneous doses. Key changes between version 10.0.28 and version 13.6.23 were identified for both AAA and AXB algorithm. With different dose prediction response, physicists and clinicians should consider the version of clinical algorithm used for dose calculation of treatment plans with extreme inhomogeneities and motion artefact.

Chapter 4

Small field dosimetric measurements for lung SBRT treatment plans

4.1 Dosimetric systems used

4.1.1 Radiochromic External Beam Therapy 3 (EBT 3) film dosimetry

Radiographic films were often used for diagnosing bone damage, dense formations such as kidney stones, dental cavities and osteoporosis. These diagnoses are based on the ability of radiation penetrating soft tissue to interact with the film while denser tissues, such as bone, attenuates the beam [67]. In radiotherapy, radiographic films are often used to verify dose distribution of the treatment field [96]. Films, in general, are extremely useful in verifying small field doses as they are thin (approximately $280\text{ }\mu\text{m}$ thick), and hence dose not result in any dose perturbation [96]. Furthermore, being a passive detector, it is easy to use because no wired connection or bias voltage is required for use. Films are also commonly used to measure flatness and symmetry of open radiotherapy photon beams and shown to have similar linear response to ionisation chambers in a relatively narrow range of doses delivered [40].

In this study, the third and latest generation of radiochromic film model, GAFCHROMICTM External Beam Therapy (EBT) 3 film (International Specialty Products, Wayne, NJ) was used. Each EBT3 film consists of a $30\text{ }\mu\text{m}$ active layer sandwiched between $125\text{ }\mu\text{m}$ matte polyester as illustrated in Figure 4.2 [40]. The matte polyester is a superficial layer, which provides protection to the active layer during handling. In addition, to avoid the creation of Newton's Rings interference patterns, a gap of approximately ten times the wavelength of visible light was maintained between the film surface and the glass window of the flatbed scanner [14]. The active layer contains

the marker dye, stabilizers and the active component giving the radiochromic film its near energy-independent response[14]. The active component contains microcrystals of a monomer dispersed in gelatin binder, which undergoes polymerisation when irradiated with ionising radiation. This polymerisation causes the radiochromic film to progressively change colour to different shades of blue[99]. The darkness of the radiochromic film is proportional to absorbed dose and can be measured as a function of Optical Density (OD) using a commercial flatbed document scanner in transmission mode (Equation 4.1) [40]. The yellow marker dye decreases UV and light sensitivity, which compliments the RGB film scanner during multichannel scanning dosimetry. It was noted that the thickness of the active layer will vary slightly between different production lots.

$$OD = -\log\left(\frac{I_0}{I_t}\right) \quad (4.1)$$

where I_0 is the intensity of the incident light and I_t is the intensity of the transmitted light through the film.

Figure 4.1 *GAFCHROMICTM EBT3 film for 0, 3, 10Gy and 20Gy.*

Radiochromic film is often used for patient-specific pre-treatment QA and has many advantages such as insensitivity to visible light and does not require tedious chemical processing. It is suitable for the verification of a wide range of megavoltage photon energies, which include most energies used in clinical photon beams. The main advantage of radiochromic film is the near tissue equivalence of the sensitive layer that leads to near energy-independence response, which is an essential characteristic to eliminate energy dependent uncertainties during lung SBRT QA [14].

The main disadvantage of radiochromic film is the large discrepancies in inter-batch sensitivity of up to 13.7% [118]. This requires the re-calibration of the sensitometric curve for each batch of film, resulting in extra workload. Although radiochromic

film does not involve chemical processing, the read-out process still requires careful handling and strict adherence to standardised protocol. The accuracy of results can be affected by factors such as time of scan, scanner warm up, scan orientation and film position within the flatbed scanner. Because the radiochromic film continues to polymerise after irradiation, a consistent wait-time of at least 24 hours (preferably 48 hours) is needed, delaying the availability of results [99, 118].

Figure 4.2 Cross-sectional view of GAFCHROMIC™ EBT3 film structure [40]

Calibration of EBT3 films

Firstly, EBT3 films were cut to 2.5 inches squares ($6.35 \times 6.35 \text{ cm}^2$). Calibration of the EBT3 films in this study were irradiated perpendicularly under 10cm depth of $30 \times 30 \text{ cm}^2$ CIRS plastic water phantom (CIRS Plastic Water Diagnostic Therapy, Norfolk, VA), SSD 100 cm setup. 6 MV photon beam with field size at $10 \times 10 \text{ cm}^2$ at isocenter from a Varian Clinac iX linac (Varian Medical Systems, Palo Alto, CA) was used. A calibrated CC13 compact chamber (Scanditronix Wellhofer, Germany) was inserted below the film plane in the plastic water phantom. This is to determine the absolute dose delivered to the film by applying the IAEA-TRS 398 code of practice, as well as to check the linac output during the irradiation.

EBT3 films were scanned on the flatbed scanner, Epson Expression 10000XL (Seiko Epson Corp., Nagano, Japan), and read using the software, EPSON SCAN v3.04. The film scanning position was kept constant so as to minimise the non-uniform response of the readout. This is due to the light scattering caused by the EBT3 film active layer particles known as lateral response artefact [92]. A $32 \times 46 \text{ cm}^2$ cardboard template was placed in the scanner to position films at a reproducible central location.

EBT3 films were also marked and scanned in landscape orientation as compared to portrait orientation as lateral response artefact is smaller [92]. RGB-positive images were scanned with a spatial resolution of 75 dpi saved in tiff format. Scan repeatability using a piece of unirradiated EBT3 film at the film scanning position shows that the deviation error was found to be less than 0.6%.

To study the reproducibility of the response of the film, the standard deviation of the net optical density from 15 pieces of film irradiated at 2 Gy under reference conditions was obtained. The 30 pieces of films were cut from different regions of 5 sheets of films from the same batch. The analysis was conducted using red channel and the standard deviation of the net optical density from all 30 pieces of films were found to be less than 1%.

Figure 4.3 Epson Expression 10000XL flatbed scanner and the film cardboard template

SNC Patient software (Sun Nuclear Corporation, Melbourne, FL) was used for the analysis of the raw EBT3 film images. The scanner response values were calibrated for red channel to net OD. Subsequently, a calibration curve that relates optical density and dose was obtained by exposing films at dose levels of 0, 1, 2, 3, 5, 8, 10, 15, 20, 25 and 30 Gy as recommended by Devic *et al* [40]. By scanning a EBT3 film repeatably, the reproducibility of the flatbed scanner was found to be less than 0.2%. As recommended by van Battum *et al* [142], the overall uncertainty of the film was found to be about 1.7%, which takes into account several sources of uncertainties such as lateral correction artefacts, background, scanner response and

intrinsic film heterogeneity. To avoid confusion, 'EBT3 film' will be referred to as 'film' in subsequent usage.

Figure 4.4 Calibrated Film Response Curve

4.1.2 Thermoluminescence dosimetry

Thermoluminescence Dosimeters (TLDs) are Lithium Fluoride crystals, doped with Magnesium and Titanium ($LiF:Mg,Ti$) to create traps in the lattice structure. Electrons that are excited during irradiation can fall into these traps to create a metastable state. When heated, these trapped electrons are released, producing photons, a process known as thermoluminescence. A photomultiplier tube (PMT) then reads the signal of the photon intensity and converts it to an amplified electrical signal which can be related to the absorbed dose to TLD.

TLDs are available in many different forms (such as powder or solid) and shapes (such as chips, rods or pellets), making it a very adaptable dosimeter for *in-vivo* dose measurement in various situations. Similar to semiconductor dosimeters, TLD has high sensitivity and is small in size. But unlike diodes, TLD has small temperature and angular dependence [72].

The major disadvantage of TLD is the time-consuming pre- and post-processing for the dose readout. A typical TLD processing involves TLD annealing for 1 hour at $400^{\circ}C$ and subsequently cooling it for 24 hours at $80^{\circ}C$. The processing of TLD must be carried out carefully and consistently to get accurate and reproducible results. Generally, the reproducibility of TLD was around 2% [72]. The poor reproducibility

of TLD may be attributed to the deviation in the crystal and impurities concentration as well as other uncertainties related to the PMT, annealing and readout procedures. TLD response is only linear to dose up to 1 Gy for LiF:Mg,Ti. For dose above the threshold, the TLD response is supralinear and may cause deviation up to 9% if uncorrected [72]. Thermoluminescence Dosimeters used in this study are Harshaw TLD-100 chips. TLD-100 chips are most used clinically as a small passive detector for *in-vivo* dosimetry.

Figure 4.5 An image of Harshaw TLD-100 chips in stored in an oxidised aluminium tray on the left and an acrylic tray on the right. Each TLD chip is unique and has identified by the rows and columns of the right tray

TLD calibration and irradiation procedures

TLD calibration is a tedious process that requires several calibration factors for accurate dosimetry. For an initial sorting, 240 TLD-100 chips were irradiated one at a time, for 1 Gy at standard reference condition (under conventional field size of $10 \times 10 \text{ cm}^2$ under 10 cm depth plastic water with SAD 100 cm)). The charge reading of each TLD was recorded before annealing. The entire process was repeated 5 times to achieve 5 individual charge readings for each of the 240 TLDs. 184 TLDs were subsequently sorted from the 240 TLDs based on its sensitivity (within 3% of the total mean charge reading) and reproducibility (within 3% of the total mean charge reading from the 5 irradiations) [63]. Out of the 184 TLDs, 12 TLDs with sensitivity within 1% from the mean reading for 5 separate measurements were identified and sorted out as calibration TLDs. The remaining 172 TLDs were used as field TLDs.

Figure 4.6 Placement of the TLD chip in the plastic water phantom for calibration

A cross-calibrated secondary standard PTW Farmer ionisation chamber (NR30013, PTW, Freiburg, Germany) was used to calibrate absorbed dose to water according to TRS-398 protocol for TLDs in CIRS plastic water. TLDs were placed in 10cm depth plastic water, field size $10 \times 10\text{cm}^2$, 6MV photon beam, at SAD 100cm. At each irradiation, 4 TLDs were held by a $30 \times 30\text{cm}^2$, 0.1cm thick polymethylmethacrylate (PMMA) positioning jig with a 0.5cm radius circle. MC simulations show a negligible effect on dose uniformity across the TLD plane due to the presence of the PMMA jig. TLD dose reading will be reported as absorbed dose to water in medium.

TLDs were annealed prior to use, so as to prevent a residual signal from uncleared traps. The annealing procedure consisted of a 400°C anneal for 60 minutes followed by an 80°C anneal for 12 hours. TLDs were annealed in oxidised aluminium (Al_2O_3) trays with individual holes to hold the TLD in place (Figure 4.5). TLDs were irradiated within 12 hours after annealing.

Figure 4.7 *Harshaw Model 3500 Manual TLD Reader*

Irradiated TLDs were read using Harshaw 3500 TLD reader (ThermoScientific, Franklin, MA) within least 24 hours after exposure (Figure 4.7). The time-temperature profile for the reader was as follows: 50°C preheat, followed by 10°C/s heating rate to a maximum temperature of 300°C. The preheating removes low temperature peaks emitted from the TLD tray and recording of the glow curve starts directly after the preheating phase. Photomultiplier tube (PMT) charge collection was integrated over the temperature ramp-up and plateau regions of the heating profile to obtain the final TLD reading in nanoCoulombs. Post-irradiation anneal at 80°C were used for 12 hours to reduce the signal from lower energy traps with low half-lives.

PMT dose-response of the TLD reader was found to have high temporal variations in its response sensitivity [65, 76] (up to $\pm 5\%$). To minimise such uncertainty, all experimental readings were normalised to the response of calibration TLDs.

Calibration TLDs were placed at the beginning and end of every sequence of 10 TLDs to establish control points. Each control point consists of 4 calibration TLDs that were exposed to 0.5 Gy in standard reference condition. The average TLD dose reading of all calibration TLDs were used for the calculation of field TLDs' dose response in the study (Equation 4.2).

Reader Calibration Factor (RCF) relates the average charge reading of a calibration TLD chip to a known absorbed dose. RCF takes into account any stochastic errors between measurements. For an accurate RCF, calibration TLDs were re-calibrated periodically from sources that are traceable to recognisable absorbed dose standards [63].

$$RCF = \frac{\langle Q \rangle}{D} \quad (4.2)$$

where $\langle Q \rangle$ refers to the average charge reading of the calibration TLDs and D represents the delivered dose.

The thermoluminescent efficiency of individual field TLD varies as they may originate from different batches of LiF crystals. Hence, Element Correction Coefficients (ECCs) relates the efficiency of individual field TLD to the average thermoluminescent efficiency of the batch, decreasing the variance of the thermoluminescent signal from 10-15% to 1-2%. [125]

$$ECC_j = \frac{RCF \times L}{q_j} \quad (4.3)$$

L is the dose delivered to the field TLDs and q_j is the charge of TLD_j .

Therefore the dose response for a specific field TLD_j will be

$$D_j = \frac{q_j \times ECC_j}{RCF} \quad (4.4)$$

4.1.3 Thermoluminescence dosimetry formalism in small static fields

In this section, the small field formalism for TLDs was derived based on the small field formalism and TLD formalism discussed previously in chapter 1. The purpose of this derivation is to understand the need for correction of TLD response in small fields.

For linacs, conventional reference field can be achieved, therefore there is no need for the intermediate machine specific reference field. Hence, the small field formalism from Equation (1.13) will be used:

$$D_{w,Q_{clin}}^{f_{clin}} = M_{Q_{ref}}^{f_{ref}} \cdot N_{D,w,Q_0} \cdot k_{Q,Q_0} \cdot \Omega_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}} \quad (1.13 \text{ revisited})$$

And the field factor formalism is as such:

$$\Omega_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}} = \frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{ref}}^{f_{ref}}} \cdot \frac{D_{w,Q_{clin}}^{f_{clin}}/M_{Q_{clin}}^{f_{clin}}}{D_{w,Q}^{f_{ref}}/M_{Q}^{f_{ref}}} \quad (1.14 \text{ revisited})$$

N_{D,w,Q_0} refers to the calibration coefficient for the absorbed dose to water to TLDs at reference beam quality Q_0 .

$$N_{D,w,Q_0} = \frac{ECC_j}{RCF} \quad (4.5)$$

Combining Equations (1.6), (4.4), (1.13), we derive the relationship for the dose delivered to TLD at standard reference conditions

$$D_{w,Q_{clin}}^{f_{clin}} = q_j \cdot \frac{ECC_j}{RCF} \cdot k_{Q,Q_0} \cdot \Omega_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}} \quad (4.6)$$

The above equations accounts for TLD measurements in small field with respect to standard reference conditions. For small field TLD measurements within a lung inhomogeneous phantom, field factor $\Omega_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}}$ will be required to account for the TLD response correction within inhomogeneity media.

4.2 TLD measurements for lung SBRT fields

Radiotherapy errors may occur at several stages between transitions from planning to treatment. Therefore quality control and quality assurance (QA) of the delivered treatment plans are necessary to ensure the precise delivery of treatment. Most QA systems includes the need for an external audit of the clinic's linac radiation beam output against a radiation standard traceable to the international measurement system. Most often, many radiotherapy departments participate regularly in these audits.

Commonly, the verification of absolute dose delivered at a reference point is referred to as the standard level I dosimetric audit. These audits were carried out via post to verify the Treatment Planning Systems (TPS) dose calculations in homogeneous phantom (water) in reference conditions. TLDs are one of the detectors used for this type of audit as they are physically small, passive and is easily handled via post.

However, level I audits are often deemed inadequate for verification of complex treatment plans such as Lung Stereotactic Body Radiation Therapy (SBRT). The accuracy of the dose calculations from various calculation algorithms has to be verified using phantoms that are more representative of patients. Such verification is a typical level II audit, which aims to verify doses under inhomogeneous non-reference conditions along the beam axis.

The challenges of the dosimetric audit for Lung SBRT lies in the verification of small fields doses in low-density lung substitute material. TLDs, even though small in physical size, may show significant dose perturbation during measurement in low-density medium and may not be meaningful without correction [95, 37, 100, 109, 38, 36, 113, 160, 57, 98, 123, 154]. Several studies show that this dose perturbation may even be further emphasised under small field conditions [38, 36, 57]. The TLD dose perturbation could only be accurately corrected by the determination of TLD

correction factors using MC methods, which can be tedious, time-consuming and requires certain specialised MC expertise [95].

In this section, we would like to investigate the correction needed for TLD small field measurements within a lung phantom. Investigations were done using a static lung phantom prototype, created for lung SBRT dosimetry audit. The trends of the simulated correction needed across several field sizes at the mid-plane of the lung phantom were investigated. A feasibility test for the dosimetry audit was done on Acuros XB (AXB) $2 \times 2 \text{ cm}^2$ dose calculation to test the accuracy of the corrected TLD measurements. In addition, the validation of lung sbrt fields was extended to QUASAR respiratory motion phantom. Modifications were done to the phantom to allow the TLD measurements within the lung tumour. TLD measurements were checked against EBT3 film measurements as well as MC simulated depth dose profiles obtained from previous chapters.

4.2.1 Static lung phantom prototype

The lung phantom was built using a composite cork cavity sandwiched between 4 cm thick of $30 \times 30 \text{ cm}^2$ CIRS plastic water above and below as shown in Figure 4.8a. 14 pieces of $15 \times 15 \text{ cm}^2$, 1 cm thick, composite cork slabs, with density 0.27 g/cm^3 were used.

Several studies did not specify the type of cork used as the lung substitute material in the lung phantom [109, 93, 16]. Chang *et al* [23] concluded that composition cork was the most suitable material based on dosimetric and physical properties. The electron density, physical density, and effective atomic number of composition cork were found to be similar to the lung tissue standard from the International Commission on Radiation Units and Measurements (ICRU, ICRU-44 lung tissue) [23]. Hence, composition cork (Advance Cork International, Delhi, India) was used as the lung substitute material for the static lung phantom prototype. The physical density of the composition cork slabs was verified to be 0.27 g/cm^3 , similar to the data from Chang *et al* [23]. The physical density of cork was calculated using measured mass and volume values. The mass of the slabs was measured using a calibrated electronic balance and the volume obtained using measurements from the vernier scale. A composition cork sample was analysed using secondary ion mass spectrometry to detect the elements reported by Chang *et al* [23]. Composite cork will be referred to as ‘cork’ from here on forth.

14 TLD measurement positions were done along the central axis at 1.5 cm, 2 cm, 3 cm, 4 cm, 6 cm, 8 cm, 11 cm, 14 cm, 16 cm, 18 cm, 18.5 cm, 19 cm, 20 cm and 21 cm below the surface. Interface dose was also measured at 4 cm and 16 cm positions. One

(a) Lung Phantom

(b) TLD measurement positions

Figure 4.8 Proposed static lung phantom prototype for lung SBRT dosimetry audit

Figure 4.9 Simulated absorbed dose to TLD in water against absorbed dose to TLD in cork

TLD was placed at each position per measurement, so as to prevent overshadowing between TLDs. Since the cork material was soft, TLDs were found to sit into the cork material and CT scans found negligible air gap at the TLD placement positions. This eliminates dosimetric error caused by air gaps, as proposed by Charles *et al* [25].

4.2.2 TLD corrections in cork

Typically, the response of TLDs was calibrated according to absorbed dose to water in water medium in standard reference condition. However, due to different mass energy absorption of lithium fluoride (*LiF*) material of TLD, the TLD response will be different from the TLD response for absorbed dose to water. It was found that mass energy absorption coefficient of *LiF* was lower than that of water and lung at mono-energetic 2MeV photon energy, as shown in Table 4.1. Therefore, MC predicted absorbed dose to *LiF* TLD would be lower as compared to the predicted absorbed dose to water and lung as shown in Figure 4.9.

Table 4.1 Mass energy absorption coefficients at 2MeV photon energy [10]

Material	μ_{en}/ρ
Water	2.608E-02
Lung	2.586E-02
LiF	2.173E-02

Additionally, in order to measure the absorbed dose to the lung substitute material such as cork, TLD absorbed dose response in cork must be accounted. The low density of composite cork, as well as small field sizes such as $2 \times 2 \text{ cm}^2$, causes LED, results in significant perturbation to the energy spectrum. Corrections must therefore be applied for the change in TLD dose response as a function of photon energy. This

is usually referred to as “energy response” in literature, which is analogous to field factor $\Omega_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}}$ in Equation 1.12. This includes corrections such as volume averaging, detector self-absorption, medium displacement. Further correction will be needed to account for the conversion from absorbed dose to TLD to absorbed dose to water.

Hence, the total TLD correction for small field measurements in cork, which will be denoted as K_{cork} , will be determined by $\Omega_{Q_{clin}, Q_{ref}}^{f_{clin}, f_{ref}}$. This is done by taking the ratio of the MC simulated dose to cork (D_{cork}) per unit MC simulated TLD response to dose in cork (M_{cork}) for the linac beam quality Q to the simulated dose in water (D_{water}) per MC simulated TLD response to dose in water (M_{water}) under calibration conditions (Equation 4.7). The field factor $\Omega_{Q_{clin}, Q_{ref}}^{f_{clin}, f_{ref}}$ accounts for the small field TLD response for dose to water (M_Q) to the TLD response for dose to water in standard reference field (M_{Q_0}). Therefore, dose to cork for beam quality Q ($D_{Q,cork}$) can be obtained using Equation 4.8 by correcting the TLD dose response obtained experimentally ($M_{experimental}$) with correction factor (K_{cork}).

$$K_{cork} = \Omega_{Q_{clin}, Q_{ref}}^{f_{clin}, f_{ref}} = \frac{(D_{cork}/M_{cork})_Q}{(D_{water}/M_{water})_{Q_0}} \cdot \frac{M_Q}{M_{Q_0}} \quad (4.7)$$

$$D_{Q,cork} = M_{experimental} \cdot K_{cork} \quad (4.8)$$

TLD response to absorbed dose within a certain medium can be determined independently by MC simulation. Calculations for the correction factor K_{cork} were done using BEAMnrc and DOSXYZnrc user codes using phase space files of the respective 6MV photon beam field sizes from Chapter 2. TLDs were simulated according to the exact weight fractions of elements contained in TLD-100 (LiF:Mg,Ti) as reported by [156]. Composite cork material was simulated according to the weight fractions of the element as reported by [23]. Dose response of water and CIRS plastic water was assumed to be the same, as simulations show that the dose response of water and CIRS plastic water differs less than 0.6% [116]. The size of the calculation voxel was set as 1.0 mm, and was consistent with AXB calculations. The number of histories generated was sufficient to produce an uncertainty of less than 1% for each of the MC simulations.

4.2.3 K_{cork} results and discussion

(a) K_{cork} calculated for 6MV photon beam with respect to depth

(b) K_{cork} for 6MV photon beam with respect to field size at 11cm depth (midplane)

Figure 4.10 Proposed lung phantom prototype for lung SBRT dosimetry audit

(a) TLD corrected measurements against AXB predicted dose profile

(b) Percentage deviation of corrected TLD measurements against uncorrected percentage deviation

Figure 4.11 Feasibility test of K_{cork} for 6MV photon beam field size $2 \times 2 \text{ cm}^2$

The calculated K_{cork} with respect to depth and field sizes within the static lung phantom was plotted in Figure 4.10. K_{cork} was found to increase as a function of field size. The larger the field size, the closer K_{cork} was to unity. For field sizes larger than $6 \times 6 \text{ cm}^2$, K_{cork} may be approximated to unity ($K_{cork} \approx 1$). MC predicts the TLD dose response using the specific composition of the TLD, LiF:Mg,Ti. Experimental results show that

TLD correction factors predicted by MC simulation for TLD within water medium was found to be near unity for all field sizes.

At water-cork interface at 4cm depth, K_{cork} was found to be near unity, which was similar to K_{cork} in water. On the contrary, at 18cm depth cork-water interface, the corrections have similar trends to K_{cork} in cork. When a photon beam passes from one medium to another medium having a different atomic number, the CPE was disturbed at the interface. Results show that the medium before the interface will impact on the interface dose measurement. TLD dose responses at the interface were predominantly due to the forward direction of the secondary electrons from interactions within the medium before the point of measurement. The magnitude of this disturbance is a function of the effective atomic number of the media [38].

The feasibility test for AXB predicted depth dose with corrected TLD measurements in the static lung phantom for field size $2 \times 2\text{cm}^2$ was presented in Figure 4.11a. Uncorrected TLD measurements deviates up to 9.8% while TLD corrected measurements were within 2% as shown in Figure 4.11b. TLD, being a high density detector, tend to over respond in low-density lung medium, especially under small field conditions [127]. DaRosa *et al* [34] made a similar study for field sizes $1 \times 1\text{cm}^2$ and $2 \times 2\text{cm}^2$, however, uncorrected TLD measurements were not presented.

In a preliminary study, K_{cork} was approximated using AXB calculations. Since AXB do not support LiF:Mg,Ti in its material library, TLDs were simulated by inserting a small water volume of 0.01cm^3 drawn in the approximate dimensions of the TLD ($3\text{mm} \times 3\text{mm} \times 1\text{mm}$), positioned within the cork medium. TLD-100 density of $2.64\text{g}/\text{cm}^3$ was assigned to the water volume [126]. By assigning the TLD density to the small water volume, TLD dose response was approximated by the number of atoms a particle encounters on a given path and the electron stopping power due to polarisation effect. However, absorbed dose due to mass stopping power was not approximated. This approximation was similar to the previous study done by Scott *et al* [127]. Generally, by comparing between AXB prediction and MC simulations of K_{cork} , was 3% lower than MC simulations. The differences are due to AXB approximation of absorbed dose to TLD and was unable to account for the difference between stopping power ratio of LiF and water.

In addition, Andreo *et al* highlighted that conversion from dose to medium to dose to water in TPS dose calculations will increase the final uncertainty [4]. This is due to the difference between electron fluence distributions in water and in medium, which in this case in cork. Hence, the conversion requires an additional fluence correction, which may have been neglected by many studies. This additional correction is included in the fluence perturbation correction factor used when calculating the TLD correction factor K_{cork} . K_{cork} corrects the dose to TLD to the measured dose in terms of dose to

water in LiF, hence, the results for TLD presented are corrected for this systematic error.

TLD perturbation is dependent on three main aspects, namely, LED, over-response of TLD in broad beam and partial volume averaging.

LED occurs when there is an imbalance of in and out scattered electrons within a small medium. The extent of LED depends on the field size, the TLD material, and the density of the surrounding medium [9, 95]. Under small field conditions, Compton interaction may generate an electron is able to transfer its energy to a point outside the field. This occurs when the distance of the point of measurement to the edge of the field is less than or equal to the Compton range of the beam. As such, interactions that produce an electron that transfers its energy to outside the radiation field will result in LED [34] (Figure 1.15). In addition, electrons have a wider range in low-density materials, such as composite cork, as compared to water. Hence, LED occurs predominantly in composite cork, resulting in the overestimation of TLD response for absorbed dose as shown (Figure 4.10).

Secondly, the over response of TLD in broad beam was due to TLDs having a higher atomic number than that of water [9, 44, 45]. 6MV photon beams have greater amounts of low-energy photons in the spectrum and are largely absorbed in higher atomic number materials due to large cross-section for photoelectric effect. This low energy photon absorption component increases with increasing field sizes. Furthermore, the stopping power ratio differences between cork and water must be accounted. For large field sizes such as $10 \times 10 \text{ cm}^2$, even though LED was not present, slight corrections will be needed as shown in Figure 4.10. For smaller fields such as $2 \times 2 \text{ cm}^2$, LED have greater significance. $4 \times 4 \text{ cm}^2$ being the intermediate field size between broad beam and small fields, K_{cork} for was also found to have an intermediate magnitude between K_{cork} for field size $10 \times 10 \text{ cm}^2$ and K_{cork} for $2 \times 2 \text{ cm}^2$.

Partial volume averaging effect will be present in the detector if the active volume of the detector was larger than the beam itself. However, due to the small size of TLD-100, this effect was deemed to be negligible in this study.

4.2.4 K_{cork} TLD corrections conclusion

The results presented in this study indicate that corrections will be required for TLD measurements in the proposed static lung phantom for lung SBRT fields. TLD correction factors calculated were dependent on the depth of measurement and field sizes. The greatest TLD correction was for the smallest field size investigated, $2 \times 2 \text{ cm}^2$, where LED was the highest in the low-density lung medium. A feasibility test was conducted successfully on AcurosXB depth doses using the simulated TLD correction

Figure 4.12 TLD Placement a QUASAR Lung Insert

factors and lung phantom prototype. TLD correction factors found will be helpful when used as a standard set of correction for Lung SBRT dosimetry audits in the proposed phantom.

4.3 QUASAR respiratory motion phantom depth dose profile measurements

In this section, the validation of lung SBRT fields was extended from the static lung phantom to the QUASAR respiratory motion phantom. This is to validate the simulated depth dose profiles in Chapter 3 with TLD and film dosimetry. Therefore, the aim is to achieve quality assurance for lung SBRT treatment plans.

Furthermore, the physical thickness of the TLD-100 chips (0.9 mm) was found to be too thick to be sandwiched by the cedar wood section of the QUASAR lung insert. The QUASAR lung insert is only suitable for EBT3 film dosimetry, where approximately 0.3 mm thickness is allowed. Hence, TLD measurements for QUASAR respiratory motion phantom could only be done for the tumour volume by replacing the polyoxymethylene sphere with a dental wax moulded sphere with an indentation for TLD-100 chips (Figure 4.12). For TLD measurements within the lung medium, a static lung phantom prototype was suggested. This allows us to study the perturbation effects of TLD within the low-density lung substitute material.

In this section, lateral and anterior beam predicted depth doses in the QUASAR respiratory motion phantom were verified via measurements using the proposed passive detectors. Both EBT3 films and TLD-100 chips were used. The lung insert was set in motion with similar settings in chapter 3 during measurement, 1 cm amplitude and 0.25 Hz sinusoidal motion.

4.3.1 EBT3 film measurement setup

EBT3 films were used to measure the depth doses across the tumour volume when under motion, within the QUASAR lung insert. The film was sandwiched between the cedar wood lung insert and the polyoxymethylene sphere tumour (Figure 4.13). The position of the film was marked on the lung insert and kept constant at lateral depth 10.325 cm to 16.675 cm for all measurements. Film measurements were normalised to dose measured at 3 cm depth of the moving chest wall.

Figure 4.13 Position of EBT3 film within the lung insert for irradiation

4.3.2 TLD measurement setup

As mentioned previously, the physical thickness of the TLD-100 chips (0.9 mm) was too thick to be sandwiched by the QUASAR lung insert. Hence, the polyoxymethylene sphere was replaced with a moulded dental wax sphere with an indentation for the TLD-100 chip. The indentation allows the TLD measurement position to be kept constant (Figure 4.12). The limitation of TLD is that it could only be used to measure point doses. TLD measurements were done at the center of the tumour volume, and were corrected for dose perturbation and normalised to dose measured at 3 cm depth of the moving chest wall.

4.3.3 Film and TLD measurement results

EBT3 film and TLD measurements for the anterior and lateral depth doses were shown below. Predicted MC depth doses agree with EBT3 film measurement of up to 2% for all field sizes. TLD measurements were in agreement with MC depth doses to within

4% for all field sizes. The highest deviation was 3.2% for TLD measurements found in lateral beam depth dose for field size $2 \times 2 \text{ cm}^2$.

Figure 4.14 Film and TLD measurement for anterior beam field size $2 \times 2 \text{ cm}^2$

Figure 4.15 Film and TLD measurement for anterior beam field size $4 \times 4 \text{ cm}^2$

Figure 4.16 Film and TLD measurement for anterior beam field size $10 \times 10 \text{ cm}^2$

Figure 4.17 Film and TLD measurement for lateral beam field size $2 \times 2 \text{ cm}^2$

Figure 4.18 Film and TLD measurement for lateral beam field size $4 \times 4 \text{ cm}^2$

Figure 4.19 Film and TLD measurement for lateral beam field size $10 \times 10 \text{ cm}^2$

4.3.4 Film and TLD measurement discussions

Results show that the full MC simulation data agrees well with film and TLD measurements. This support the claim that full MC simulation data used in this thesis is a reliable benchmark for dose calculation, even in extreme inhomogeneities and in the presence of motion artefact.

Furthermore, results show that among the clinical algorithms tested, AXBv13.6 algorithm has the best agreement with MC benchmark and measurement data. This is consistent with the conclusion from the depth dose comparison study in the previous chapter.

For the moving tumour under lateral beam, measurement positions were based on the position of dosimeter placement at the lung insert equilibrium position. This may result in an increased uncertainty in the position of measurement as well as measurement accuracy. To minimise this uncertainty, average of five sets of measurements were taken for each position.

Corrections for TLD dose perturbations were found to be negligible in the tumour volume. During TLD measurement, the tumour volume was replaced with a water equivalent wax phantom to allow the placement of the TLD in the lung insert. Therefore, due to the water equivalent medium, correction for dose perturbation due to the surrounding medium is close to unity as shown in Figure 4.10.

As discussed in the previous sections, the moving chest wall introduces motion artefacts such as blurring in the CT image. Motion may introduce differences in scatter conditions in calculated doses as compared to the film and TLD measurements. In addition, the position of film and TLD measurements may introduce a certain level of uncertainty due to motion, such as having a smeared out measured dose along the axis of motion in the dose gradient regions. Despite the presence of motion artefacts, both AAA and AXB algorithms still show acceptable agreement to film and TLD measurements.

4.4 Lung SBRT film dosimetry QA using QUASAR respiratory motion phantom

Typical lung SBRT multi-fraction dose treatments include 48-60 Gy in 3-5 fractions. High geometric and dosimetric precision is important as high doses are being delivered in only a few fractions for lung SBRT treatment. Quality assurance of the treatment plan delivery is, therefore, vital for reliable lung SBRT treatment. The purpose of this section is to validate a clinically implemented SBRT lung treatment delivery using

the QUASAR respiratory motion phantom, both in static and moving configuration, against film dosimetric measurements.

Firstly, a VMAT lung SBRT clinical plan with 3 arcs was chosen (Figure 4.20) and calculated using AXBv13.6, AAAv13.6, AXBv10.0 and AAAv10.0 in Eclipse TPS. This treatment plan was then delivered onto an ArcCHECK Phantom (Sun Nuclear Corporation, Melbourne, FL) in static configuration. The ArcCHECK is a hollow cylindrical phantom with 1386 diodes placed with 1 cm geometry spacing in a spiral array (Figure 4.21). It is specifically build for routine patient specific IMRT or VMAT QA. The purpose of the ArcCHECK measurements was to obtain confidence for deliverability of the clinical plan calculated using different algorithms.

Subsequently, the same VMAT lung SBRT clinical plan was calculated onto the QUASAR respiratory motion phantom using AXBv13.6, AAAv13.6, AXBv10.0 and AAAv10.0 in Eclipse TPS. Static and moving 4D computed tomography (4D-CT) scans were acquired for QUASAR respiratory motion phantom (Modus Medical Devices Inc, London, ON). A cedar wood insert containing a 3 cm diameter polyoxymethylene sphere target volume was used as the lung phantom. For the moving configuration, a sinusoidal breathing cycle of 15 breaths per minute (BPM) with 1cm amplitude was used. 2D dosimetry QA was done using Gafchromic EBT3 film (ISP, Wayne, NJ) for both the static and moving configuration.

The clinical treatment plan was delivered using a 6 MV beam from a Varian Clinac iX linear accelerator (Varian Medical Systems), with Advanced On-Board Imaging cone beam CT (OBI CBCT, Varian Medical Systems) for phantom setup verification. ArcCHECK and film dosimetric measurements were compared and analysed against calculated Eclipse TPS calculated doses using SNC Patient QA software (Sun Nuclear Corporation, Melbourne, FL).

Gamma analysis of the measured doses was evaluated using global normalisation, where percentage dose difference was calculated relative to the maximum dose of the measured plane. In addition, a 10% threshold value was set for the gamma analysis, such that only measurement points with dose values larger than 10% of the maximum dose were included in the analysis. The purpose of excluding several measurement points from evaluation and normalising the measurements to a reference point was to exclude sufficient low dose regions in the treatment plan for analysis. These excluded regions may not be as clinically significant and dosimetrically important when compared to the high dose regions within the treatment volume.

Figure 4.20 Film $\gamma_{3.0/3.0}$ analysis against AXB calculated dose plane

Figure 4.21 ArcCHECK phantom

4.4.1 A lung SBRT QA example using ArcCHECK measurement

The VMAT lung SBRT clinical plan was calculated using AXBv13.6, AAAv13.6, AXBv10.0 and AAAv10.0 in Eclipse TPS and delivered to the ArcCHECK Phantom. The purpose of this is to gain confidence for the deliverability of the treatment plan from the linac output, as well as to detect any deliverability differences from the treatment calculation between different algorithms.

Absolute dose calibration was done on the ArcCHECK phantom as the first step. Absolute dose calibration converts the ArcCHECK measured relative dose values to absolute dose values. A CT image of the ArcCHECK phantom was uploaded into

the Eclipse TPS for calculation. AAA and AXB algorithms account for the CT to electron density or mass density of the ArcCHECK phantom differently. For AXB algorithms, the ArcCHECK CT structure was assigned as water material and HU value was entered iteratively until 1.1836 g/cm^3 mass density was reached in the TPS, as recommended by the manufacturer [134]. Similarly for AAA, the ArcCHECK CT structure was assigned as water and HU value was entered iteratively until 1.150 g/cm^3 electron density was reached in the TPS, as recommended by the manufacturer [134]. Both AAA and AXB will have these two specific calibration settings during ArcCHECK measurement. Preliminary absolute dose calibration of the ArcCHECK phantom was done under a 6 MV photon beam, SAD = 100 cm and $10 \times 10 \text{ cm}^2$ open field. The point dose reading of the ArcCHECK anterior detector at 2.9 cm depth was assigned as the dose measurement in water 3.3 cm depth from the earlier TRS-398 reference dose calibration. The calculated point dose for the ArcCHECK anterior detector for both AAA and AXB algorithm was found to be consistent to within 0.2%. Reproducibility of the response of the ArcCHECK was also validated. 1 Gy was delivered to the ArcCHECK for field size $10 \times 10 \text{ cm}^2$ and 5 measurements were taken. The reproducibility of the response of the ArcCHECK was found to be below 0.5%.

Figure 4.22 ArcCHECK measurement results

Table 4.2 ArcCHECK gamma analysis results

Static Configuration	ArcCHECK			
Calculation Algorithm	AXBv13.6	AAAv13.6	AXBv10.0	AAAv10.0
% Gamma (3.0 %/3.0 mm)	99.7	99.7	100	99.4

Consequently, the lung SBRT treatment plan was delivered onto the ArcCHECK phantom. Lung SBRT QA results from the ArcCHECK were summarised in Table 4.2.

Gamma analysis of threshold 10% with global dose normalisation, as proposed by the AAPM Task Group 119 [46], was used with $\gamma_{3.0\%/3.0mm}$ criteria.

Generally, results show consistent pass rates to within 0.6% when compared with each other. Results indicate that deliverability of the lung SBRT treatment plan by the linac is acceptable and consistent regardless of the version and the type of calculation algorithm used under homogeneous conditions. Furthermore, the slight discrepancy in the ArcCHECK measurement result from AAAv10 may be due to the difference in couch correction using different algorithm in the VMAT plan.

4.4.2 A lung SBRT QA example using film dosimetry

2D film dosimetry analysis was done using Gafchromic EBT3 film for both the static and moving configuration of the QUASAR respiratory motion phantom. Analysis was done using relative dosimetry under inhomogeneous conditions. The moving configuration of the QUASAR respiratory motion phantom challenges the QA of clinical treatment plans under the presence of motion artefacts.

Figure 4.23 Film $\gamma_{3.0/3.0}$ analysis against AXB calculated dose plane

Table 4.3 Film Gamma Analysis results

% Gamma (3.0 %/3.0 mm)	Film			
Calculation Algorithm	AXBv13.6	AAAv13.6	AXBv10.0	AAAv10.0
Static Configuration	99.9	99.6	99.9	99.6
Moving Configuration	98.8	98.1	98.1	97.9

Dosimetric QA results are summarised in Table 4.3. Similarly, gamma analysis of threshold 10% with global dose normalisation was used with $\gamma_{3.0\%/3.0mm}$ criteria. Generally, the moving configurations of the clinical plan have the highest disagreement between the calculated treatment plan and measured 2D dose distribution. Most of the disagreements for film measurements were found at the high dose gradient region in the anterior-posterior direction, which is along the axis of the phantom motion. Generally the film measurements were consistent in reporting a lower point dose than the TPS. For the static configuration, gamma 3.0 %/3.0 mm analysis show that QA

results from film dosimetry for different algorithms show consistent results to within 0.4% pass rate when compared with each other.

4.4.3 Dosimetric QA discussions

For the dosimetric QA, gamma analysis show consistent results for all investigated calculation algorithms. One may argue that by using global normalisation and 10% threshold for the gamma analysis, there is a bias towards the best possible gamma passing rate and not justly quantifying dose distribution agreements. Local normalisation, in which percentage dose difference was being calculated relative to dose at each reference point, should be used. However, in practice, physicists use several commercial QA tools that have different criteria regarding acceptance levels for gamma analysis results. By using local normalisation, the number of reference points being compared will vary between different QA tests. Furthermore, local normalisation gamma analysis usually result in lower passing rates only at low dose and low gradient regions, which are often not within the region of interest and not clinically significant. In addition, it has been highlighted that commercial TPS algorithms have poor modelling of low dose profiles [132]. It seems therefore acceptable that by evaluating based on the normalised measurement points in the high dose region, which consist of an average of about 95% of the points on the plane within the region of interest, to be included in the analysis [46]. These points analysed should pass the set gamma criteria of 3%/3 mm with a confidence limit that ranges down to 88% as suggested by Ezzell *et al* [46]. Moreover, there was a fair inter-comparison of the dosimetric QA as the same criteria of gamma analysis of 3%/3 mm, with global normalisation and 10% threshold was kept constant for all analyses.

ArcCHECK CT structure was assigned as PMMA for AXB algorithm calculations initially. However, dose discrepancies were found for AAA and AXB algorithm absolute point dose calculations against the ArcCHECK anterior diode detector measurement. This is due to different the different cross-sectional data assigned to the ArcCHECK CT structure by AXB algorithm, as compared to the automatic assignment of water through the CT to electron density calibration curve by AAA. It was found that AAA and AXB algorithm point dose calculation for absolute dose calibration will have consistent agreement when ArcCHECK CT structure was assigned as water in AXB algorithm calculations.

Generally, under static conditions, gamma analysis show that dose distributions calculated using AAA and AXB algorithm generally have consistent gamma pass rates to within 0.3% against film dosimetric measurements, even in inhomogeneous conditions with small field sizes and small treatment target. The marginal difference

of 0.3% pass rate is due different inhomogeneity correction from AXB algorithm as compared to AAA. This trend was found to be consistent between both version 10.0 and version 13.6. However, gamma analysis with $\gamma_{3.0\%/3mm}$ criteria may be too lenient for the analysis of inhomogeneous dose distribution differences between AAA and AXB algorithm. The previous depth dose comparison study done between AAA and AXB algorithm was more capable in identifying dose distribution differences with greater certainty.

Under the moving configuration, lower gamma pass rates were seen for all dose distribution calculated. As discussed in the previous sections, this blurring in the CT image is caused by motion. Motion may introduce differences in scatter conditions in calculated doses as compared to the film measurements. In addition, the position of film measurement may introduce a certain level of uncertainty due to motion, such as having a smeared out measured dose along the axis of motion in the dose gradient regions. Despite the presence of motion artefacts, both AAA and AXB algorithms still show acceptable gamma pass rates of above 95%. Gamma pass rates for the dose distribution calculated from AXBv13.6 show that it is able to agree with film dosimetric measurements better than AAAv13.6, AAAv10.0 and AXBv13.6.

Overall, film dosimetry QA and ArcCHECK 2D measurements agree with the depth dose measurement comparison study from the previous chapter. AXBv13.6 algorithm was found to be suitable and well validated for calculation of dose distributions in lung, even in extreme inhomogeneous small field conditions. Further studies may be done to pose a greater challenge to the performance of AXBv13.6 algorithm. Dose calculations within high density heterogenous media such as bone and titanium, as well as complex target shapes and very small lesions may be investigated. This will be useful for the investigation of dose distributions within patients with cardiac implantable electronic devices, high-density implants and prostheses. Such cardiac implantable electronic devices are susceptible to ionizing and electromagnetic radiation, which may result in transient or permanent device defects if irradiated. Furthermore, irradiation of high-density implants and prostheses are known to cause local dose perturbations due to lateral and backscatter of radiation, which will result in the depletion of downstream dose due to attenuation [78]. Prisciandaro *et al* investigated methods to estimate doses to cardiac implantable electronic devices using published peripheral dose data and calculations from three algorithms, namely UMPlan, AAA and AXB algorithm [114]. Differences of up to 3 Gy in calculated and measured doses were observed for AXB algorithm. However, the version used in the study was v11.031 for AAA and AXB algorithm. A more recent investigation of the updated versions of AAA and AXB algorithm may shed some light to this topic.

The ArcCHECK QA of the lung SBRT VMAT treatment plan was done as per clinical routine QA verification using ArcCHECK calculated dose. The methods used are similar to routine clinical protocol used during the verification of a calculated clinical treatment plan. Measurements done within the static lung phantom and the moving QUASAR lung phantom were to illustrate measurements done during radiation audits and credentialing.

Kry *et al* and Carson *et al* highlighted that passive detectors, such as TLDs and films, are robust and reliable detectors to be used for credentialing and QA exercises [21, 82]. This is consistent to the proposed detectors used in the thesis.

However, both authors argued that routine QA phantoms are not a suitable replacement for a credentialing phantom. Their study show poor agreement between routine QA results when compared to measurements done on the credentialing phantom [82]. A more stringent criteria was also suggested for the dosimetric audits [21].

Credentialing phantoms and routine QA phantoms have different functions. The main function for credentialing phantoms is to conduct dosimetric audit for institutions to deliver accurate and consistent radiation therapy. Such credentialing will be most suitable especially at the time of commissioning of a new radiotherapy department, or during an annual or bi-annual radiation quality audit. It is, however, not practical to do end-to-end test for all clinical patient-specific QA. To do such audits for all clinical cases will be too time and labour intensive. Therefore, the function of routine QA phantoms is to provide efficient routine dose verification and monitoring during day-to-day delivery. The results from routine QA should be monitored to fall within a set baseline established during credentialing.

Credentialing phantom results also include the summation of all errors that occurred during the entire workflow, including scanning, contouring, planning, setting up, and delivery. Dosimetric errors may result from inaccuracies such as during beam modeling within the TPS or incorrect calibration. Routine QA usually uses several different devices, delivery techniques, and analyses, to pick up the specific cause of these errors. As such, with consistent and accurate dose delivery as the goal, the credentialing phantom and routine QA phantoms can be used together to achieve irradiation results that are processed, analysed, and evaluated in a consistent robust manner.

By conducting measurements within static and moving lung phantom, as well as the inclusion of the measurements on an ArcCHECK phantom, this chapter illustrates a possible future quality control platform that covers both the radiation credentialing phantom as well as routine QA phantoms.

4.4.4 Conclusion

In this chapter, dosimetric studies for lung SBRT small fields were done. It was found that TLD measurements require corrections especially for small field measurements within low density lung medium. This is because the high TLD physical density causes dose perturbation in low density lung medium. Correction factors for TLD in cork varies from 0.9 to 1 in cork medium for the smallest field size investigated, $2 \times 2 \text{ cm}^2$. On the other hand, dose perturbation for EBT3 film measurements were negligible as they are extremely thin. Subsequently, depth dose measurements using TLD and film were done to verify the depth dose comparison results in the previous chapter. Predicted MC depth doses agree with EBT3 film and TLD measurement of up to 4% for all field sizes. AXBv13.6 prove to have the best agreement to measured data, which is consistent with the previous conclusion with MC benchmark. For both static and moving configurations, AXBv13.6 shows the smallest deviation from film and TLD measurements. Under ArcCHECK and film dosimetry QA, AXBv13.6 show the highest gamma pass rates with 99.9% $\gamma_{3.0\%/3.0\text{mm}}$ passing rate for static configuration and 98.8% $\gamma_{3.0\%/3.0\text{mm}}$ passing rate for moving configuration. The findings from this chapter may be used to set possible benchmarks for future quality control for lung SBRT QA.

Chapter 5

Conclusion

5.1 Summary of thesis

The broad theme of this thesis is to improve the accuracy in dose calculation for lung SBRT treatments in radiotherapy. Chapter 1 briefly outlined the development and challenges of clinical radiotherapy for lung cancer, particularly lung SBRT treatment, so as to ensure safety and accurate delivery of treatment to patients. There are many various practices for lung cancer radiotherapy. Conventional radiotherapy treatment such as 3D-CRT may be unsuitable as they have poor TCP levels and low five-year survival rates. Lung SBRT treatment was adapted from brain stereotactic radiosurgery treatment and was found to be suitable to treat early-staged lung cancer patients who were not suitable for surgery. However, adaptation was not as straightforward as lung tumour motion due to human respiration and complicated radiation dosimetry in low-density inhomogeneous lung medium may result in challenges in dose prediction during treatment planning. Advanced imaging techniques such as 4D-CT scan were introduced to complement lung SBRT treatment to reduce uncertainty associated with these challenges. Hence, the aim of this thesis is to determine the accuracy of different dose calculation algorithms under small field and low density lung medium, in the presence of tumour motion. Chapter 1 also summarised the background and theory of the radiation physics associated with radiotherapy. The definition, code of practice and challenges associated with small field radiation photon beams found in lung SBRT treatments were also discussed. Additionally, dose calculation algorithms used in this thesis for the prediction of doses were also highlighted.

EGSnrc MC simulation was used as the dose calculation algorithm benchmark in this thesis. In order to obtain a reliable and robust MC simulation, the MC simulation calculated dose output was verified against measurements within a water phantom. The first fundamental step in MC simulation is the accurate determination of the primary

electron beam parameters. Chapter 2 concluded that the accurate determination of the primary electron beam parameters for MC simulation can be achieved using 1D gamma analysis. A set of optimal parameters for all field sizes was obtained based on the result using 95% passes with the most stringent $\gamma_{0.5/0.5}$ criteria investigated.

Chapter 3 studied the dose prediction capabilities of several algorithms using QUASAR respiratory motion phantom to simulate small field low-density lung conditions, in the presence of motion artefacts. It concluded that 6X FFF was capable of producing equivalent depth doses to MC benchmark. AAAv13.6 and AXBv13.6 also proved to be more competent in extreme inhomogeneous dose prediction than AAAv10.0 and AXBv10.0. Key changes between version 10.0.28 and version 13.6.23 were identified for both AAA and AXB algorithm. With different dose prediction response, physicists and clinicians should consider the version of clinical algorithm used for dose calculation of treatment plans with extreme inhomogeneities and motion artefacts such as lung SBRT treatment plans.

Chapter 4 investigated *in-vivo* dosimetry for lung SBRT treatment fields. MC simulation shows that dose perturbation is negligible for EBT3 films in small fields. However, despite the small physical size of TLD chips, MC simulations show that TLD require significant correction for measurements within small field low density medium. TLD small field formalism for dose perturbation was formulated and dose perturbation correction factors were calculated for field size $2 \times 2 \text{ cm}^2$, $4 \times 4 \text{ cm}^2$ and $10 \times 10 \text{ cm}^2$. A feasibility test for the perturbation correction factors calculated was done against AXB calculated $2 \times 2 \text{ cm}^2$ depth dose and was found to be accurate. Dosimetric study was done using TLD and film dosimetry at the isocenter of the phantom for the depth doses compared in Chapter 3. Measurements for both static and moving configuration were conducted. Results shown were consistent with previous study from Chapter 3. AXBv13.6 was the clinical algorithm that has the best agreement with film and TLD measurements.

Furthermore, a clinical lung SBRT VMAT treatment plan was calculated using both versions of AAA and AXB algorithms on a ArcCHECK phantom to check the plan's deliverability. Subsequently, the plan was also calculated by both versions of AAA and AXB algorithm on the static and moving configuration of the QUASAR respiratory motion phantom. Film dosimetry QA was done to compare the agreement of dose distribution calculated using different algorithms for the same plan. QA results show extremely good agreement between measurements and calculated plans for static configuration. However, poorer agreement was found for film measurements done in the moving configuration. Due to the blurring and motion artefacts, film measurements resulted in smearing out of measured dose in the gradient regions resulting in higher measurement uncertainty. Despite this, AXBv13.6 shows consistently higher gamma

pass rates when compared with other calculated algorithms in both static and moving configurations.

5.2 Study limitations and suggestions for future study

Clinical treatment plan studied should have been calculated using MC and included in the QA study for fair comparison. MC simulation was not included because we lacked the resource for the high computational power needed for such a computation, and therefore may result in extremely long computational time. However, assuming that QA treatment plan is dependent on the calculated depth doses, depth dose comparison study from the previous chapter concluded the accuracy of AXBv13.6 has the best agreement to MC. MC simulated depth dose also show good agreement when validated against film and TLD dosimetry.

ArcCHECK and film dosimetry QA study was also limited to one VMAT lung SBRT clinical plan. Future analysis may include larger numbers of lung SBRT clinical plans to achieve a statistically robust sample size. Nonetheless, the aim of ArcCHECK and film dosimetry QA is to investigate the deliverability differences and the dose calculation accuracy between different algorithms. The same clinical plan studied was a constant factor that acts as a benchmark to compare the differences between calculations. Future studies may include different clinical plans, to investigate the consistency of the results.

Motion study done was only restricted to sinusoidal motion for a controlled motion study. Patient specific breathing patterns could be studied and uploaded to the QUASAR respiratory motion phantom for a more realistic and detailed study. Non-sinusoidal motion may result in different dose distribution, as doses may not be smeared evenly in the direction of motion. Further studies in the future could be done to investigate the impact of different breathing patterns on dose calculations.

5.3 Final conclusion

A robust MC simulation benchmark can be achieved by analysing the simulated dose distributions from simulated field sizes with increasingly stringent gamma criteria. Furthermore, it was also found that for *in-vivo* measurements of lung SBRT fields within lung medium, corrections would be needed for accurate TLD readings. TLD tend to over respond in low density lung medium due to dose perturbation. On the other hand, no perturbation correction will be required for film dosimetry.

Both MC simulation and dosimetric studies concluded that, comparing both versions of AAA and AXB algorithms, clinical AXBv13.6 algorithm have the best dose prediction capability for lung SBRT treatment planning, even in extreme inhomogeneous condition with the presence of motion artefacts.

Accurate dose corrections and accurate dose predictions are of extreme importance in the field of medical physics. Any miscalculation of corrections or predictions will result in the misadministration of radiation doses, which may lead to an increased probability of associated toxicity for the organ at risk, or an ineffective treatment from the underdosing of the target volume. The findings in this study will assist physicists and clinicians in considering the version of clinical algorithm used for lung SBRT treatment plan dose calculation, as well as assist in the choice of detectors used for lung SBRT treatment plan quality assurance.

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List Of Publications for thesis:
Monte Carlo studies for small field dosimetry in
Lung Stereotactic Body Radiotherapy

Roger Soh Cai Xiang
Nanyang Technological University

2017

List of original work and publications

List of publications

- I **Roger Cai Xiang Soh**, Lloyd Kuan Rui Tan, Mark Sen Liang Goh, Sitti Yani, Freddy Haryanto, Wen Siang Lew and James Cheow Lei Lee “A gamma analysis approach to determine optimal parameters for Monte Carlo simulation of 6MV Varian ClinacIX photon beam”, Journal of Health Science, **5**, 36-43, 2017
- II **Roger Cai Xiang Soh**, Guan Heng Tay, Wen Siang Lew, Shaun Baggarley and James Cheow Lei Lee “TLD Correction Factors For Field Sizes Used in Lung Stereotactic Body Radiation Therapy Dosimetric Measurements”, Journal of Medical Physics, **42**, Supplement 1, AB-3 ,2017
- III **Roger Cai Xiang Soh**, Lloyd Kuan Rui Tan, Mark Sen Liang Goh, Sitti Yani, Freddy Haryanto, Wen Siang Lew and James Cheow Lei Lee “1D Gamma Analysis on the effect of Mean Electron Energy, FWHM AND XY Jaws thickness for small fields”, Medical Physics International Journal, **4**, 2, 157, 2016
- IV **Roger Cai Xiang Soh**, Khong Wei Ang, Wen Siang Lew and James Cheow Lei Lee “A depth dose study between AAA, AXB algorithm and 6X FFF beam against Monte Carlo simulation under respiratory motion with a moving chest wall”, Reports of Practical Oncology and Radiotherapy, Under Review
- V Chloe Chiew Hyen Ong, Khong Wei Ang, **Roger Cai Xiang Soh**, Kah Ming Tin, Jerome Haw Hwong Yap, James Cheow Lei Lee, Christopher Bragg “Dosimetric Comparison of Peripheral NSCLC SBRT using Acuros XB and AAA Calculation Algorithms”, Medical Dosimetry, **42**, 3, 216-222 ,2017
- VI Sitti Yani, I Gde Eka Dirgayussa, Mohamad Fahdillah Rhani, **Roger Cai Xiang Soh**, Freddy Haryanto and Idam Arif, “Monte Carlo study on electron contamination and output factors of small field dosimetry in 6 MV photon beam”, Smart Science, **4**, 2, 2016
- VII Sitti Yani, Ridwan Ramdani, Mohamad Fahdillah Rhani, **Roger Cai Xiang Soh**, Freddy Haryanto and Idam Arif, “Monte Carlo Simulation of Varian Clinac iX 10MV Photon Beam for Small Field Dosimetry”, International Journal of Radiation Research, **15**, 3, 2017

Conference Proceedings

- I **Roger Cai Xiang Soh**, and James Cheow Lei Lee “A TLD Study of Acuros XB for Lung SBRT Using A Lung Substitute Material”, 12th Asia-Oceania Congress of Medical Physics & 10th South East Asia Congress of Medical Physics, December 2012, Chang Mai, Thailand
- II **Roger Cai Xiang Soh**, Benjamin Min Cheng Chua and James Cheow Lei Lee “Acuros XB study of small field correction factors for TLD in heterogeneous medium”, 13th Asia-Oceania Congress of Medical Physics & 11th South East Asia Congress of Medical Physics, December 2013, Singapore
- III **Roger Cai Xiang Soh**, I G. E. Dirgayussa, Freddy Haryanto, Wen Siang Lew and James Cheow Lei Lee “Determination and Comparison of Correction Factors Obtained for TLDs in Small Field Lung Heterogeneous Phantom Using Acuros XB and EGSnrc”, American Associations of Physics in Medicine (AAPM) Annual Meeting, 2014
- IV I Gde Eka Dirgayussa, Sitti Yani, Mohamad Fahdillah Rhani, **Roger Cai Xiang Soh**, Freddy Haryanto, “Investigation of the Inaccuracy of MLC Leaf Position on VMAT Delivery Technique using EGSnrc Monte Carlo Simulation”, 14th Asia-Oceania Congress of Medical Physics & 12th South East Asia Congress of Medical Physics, October 2014, Vietnam
- V Sitti Yani, Rasito Tursinah, Mohamad Fahdillah Rhani, **Roger Cai Xiang Soh**, Freddy Haryanto and Idam Arif “Neutron contamination of Varian Clinac iX 10 MV photon beam using Monte Carlo simulation”, Journal of Physics Conference Series, **694**, 2016

Appendix A

Technical specifications of the QUASAR respiratory motion phantom

General Information

The QUASAR respiratory motion phantom is a patient breathing simulator to allow quality assurance tests on clinical radiotherapy systems. Specific patient respiratory waveforms may be programmed to the QUASAR respiratory motion phantom using the software application to simulate the breathing function. The phantom includes several interchangeable inserts for several quality assurance tests and is flexible to work with several motion tracking systems. The phantom is designed to move cylindrical inserts in the superior-inferior direction within a body shaped acrylic phantom. The speed and amplitude of the motion can be varied. The motion of the insert is linked to a moving platform designed to carry a respiratory tracking device.

Specifications

- 7 kg acrylic body oval: 30 cm W x 20 cm H x 12 cm L
- 5 kg drive unit: 20 cm L x 15 cm W x 12 cm H
- Total weight is approximately 20 kg including moving insert and drive unit
- 2 openings in body oval: 8 cm diameter each, for drive unit and moving insert

- Moving insert : 8cm diameter cylindrical cedar wood lung insert with 3 cm diameter solid polyoxymethylene spherical target volume. An additional 4cm thick chest wall wax mould was added to the cedar wood lung insert.
- The dental modelling wax used is a tissue equivalent material that allows the simulation of the buildup dose before the cedar wood lung phantom insert.
- Moving platform: 13 cm diameter, carries up to 500 g
- Power supply: input, 100 - 240 V AC, 47 - 63 Hz
- Output, 24 V DC 2.1 A, 50 W

Table A.1 QUASAR lung insert materials

	Density (g/cm^3)	Material Composition
Metrowax	0.85	Paraffin and Microcrystalline Waxes [94]
Polyoxymethylene sphere	1.41	$(CH_2O)_n$ [26]
Cedar Wood	0.26	6% H, 43% O, 50% C [105]

[H]

Figure A.1 A schematic diagram of the QUASAR lung insert with the chest wall

Figure A.2 Components of the QUASAR respiratory motion phantom

Figure A.3 Front View of the QUASAR respiratory motion phantom

Figure A.4 Side View of the QUASAR respiratory motion phantom

Figure A.5 Top View of the QUASAR respiratory motion phantom

Appendix B

EGSnrc input files

BEAMnrc Input file

The section below shows a BEAMnrc input file for 6 MV Varian Clinac iX 10×10 cm^2 photon beam. The input file parameters were used to build the geometry of the component modules of the linac head in BEAMnrc. The description of these parameters can be found in the BEAMnrc Users Manual [121].

```
Fullrogerhead_5.7_IAEA
AIR700ICRU
0, 0, 0, 0, 0, 2, 0,  IWATCH ETC.
1000000000, 1534, 1447, 10000, 2, 1000, 0, 0,  NCASE ETC.
20, 100, 4, 21, 1, 13.83942,  DIRECTIONAL BREM OPTIONS
-1, 19, -0.3, 0, 0, 1, 0, -0.3, 0.0, 0.0,  IQIN, ISOURCE + OPTIONS
0, MONOENERGETIC
6.1
0, 0, 0.7, 0.01, 0, 1, 2,  0 , ECUT,PCUT,IREJCT,ESAVE
0, 0, 0, 0, 0,  PHOTON FORCING
1, 10,  SCORING INPUT
1, 1
30,
0,  DOSE COMPONENTS
0.0, Z TO FRONT FACE
***** start of CM SLABS with identifier Xraytgt *****
10, RMAX
Xraytgt
2, NSLABS
0, ZMIN
```

```

0.0889, 0.7, 0.01, 0, 0, 0
W700ICRU
0.1575, 0.7, 0.01, 0, 0, 0
CU700ICRU
***** start of CM CONS3R with identifier pricol *****
25, RMAX
pricol
0.2464, ZMIN
7.3836, ZTHICK
2, NUM_NODE
0.2464, 1,
7.63, 2.09,
0.7, 0.01, 0, 0, 0,
VACUUM
0.7, 0.01, 0, 0, 0,
W700ICRU
***** start of CM SLABS with identifier vaccwin *****
5, RMAX
vaccwin
2, NSLABS
7.631, ZMIN
1.029, 0.7, 0.01, 0, 0, 0
VACUUM
0.00254, 0.7, 0.01, 0, 0, 0
BE700ICRU
***** start of CM FLATFILT with identifier flatfilt *****
30, RMAX
flatfilt
11.782, ZMIN
20, NUMBER OF LAYERS
1, 0.02794, # CONES, ZTHICK OF LAYER 1
0,
0.0635,
1, 0.02794, # CONES, ZTHICK OF LAYER 2
0.0635,
0.127,
1, 0.0381, # CONES, ZTHICK OF LAYER 3
0.127,

```

0.1905,
1, 0.04064, # CONES, ZTHICK OF LAYER 4
0.1905,
0.254,
1, 0.07366, # CONES, ZTHICK OF LAYER 5
0.254,
0.381,
1, 0.13462, # CONES, ZTHICK OF LAYER 6
0.381,
0.508,
1, 0.10922, # CONES, ZTHICK OF LAYER 7
0.508,
0.635,
1, 0.10922, # CONES, ZTHICK OF LAYER 8
0.635,
0.762,
1, 0.11176, # CONES, ZTHICK OF LAYER 9
0.762,
0.889,
1, 0.10414, # CONES, ZTHICK OF LAYER 10
0.889,
1.016,
1, 0.20828, # CONES, ZTHICK OF LAYER 11
1.016,
1.27,
1, 0.1905, # CONES, ZTHICK OF LAYER 12
1.27,
1.524,
1, 0.18542, # CONES, ZTHICK OF LAYER 13
1.524,
1.778,
1, 0.16764, # CONES, ZTHICK OF LAYER 14
1.778,
2.032,
1, 0.15494, # CONES, ZTHICK OF LAYER 15
2.032,
2.286,
1, 0.14224, # CONES, ZTHICK OF LAYER 16

```
2.286,  
2.54,  
1, 0.07366, # CONES, ZTHICK OF LAYER 17  
2.54,  
2.684526,  
3, 0.05588, # CONES, ZTHICK OF LAYER 18  
2.684526, 3.3655, 3.81,  
2.794, 3.3043368, 3.81,  
3, 0.09652, # CONES, ZTHICK OF LAYER 19  
2.794, 3.3043368, 3.81,  
3.0607, 3.302, 3.81,  
1, 0.1651, # CONES, ZTHICK OF LAYER 20  
3.81,  
3.81,  
0.7, 0.01, 0, 0,  
CU700ICRU  
0.7, 0.01, 0, 0,  
AIR700ICRU  
0.7, 0.01, 0, 0,  
CU700ICRU  
0.7, 0.01, 0, 0,  
AIR700ICRU  
0.7, 0.01, 0, 0,  
CU700ICRU  
0.7, 0.01, 0, 0,  
AIR700ICRU  
0.7, 0.01, 0, 0,  
CU700ICRU  
0.7, 0.01, 0, 0,  
AIR700ICRU  
0.7, 0.01, 0, 0,  
CU700ICRU  
0.7, 0.01, 0, 0,  
AIR700ICRU  
0, 0, 0, 0,  
CU700ICRU  
0, 0, 0, 0,  
AIR700ICRU
```



```

0, 0, 0, 0,
AIR700ICRU
0.7, 0.01, 0, 0,
CU700ICRU
0.7, 0.01, 0, 0,
AIR700ICRU
0, 0, 0, 0,
CU700ICRU
0, 0, 0, 0,
AIR700ICRU
0, 0, 0, 0,
CU700ICRU
0, 0, 0, 0,
AIR700ICRU
0, 0, 0, 0,
CU700ICRU
0, 0, 0, 0,
AIR700ICRU
0, 0, 0, 0,
CU700ICRU
0, 0, 0, 0,
AIR700ICRU
0.512, 0, 0, 0,
CU700ICRU
0.512, 0, 0, 0,
AIR700ICRU
***** start of CM CHAMBER with identifier chamber *****
7.622, RMAX
chamber
14.21, ZMIN
1, 20, 1, N_TOP, N_CHM, N_BOT
0.68, 4.9, 0, ZTHICK, RCYS, FLAG FOR LAYER 1 IN TOP
0.7, 0.01, 0, 0,
AIR700ICRU
0.7, 0.01, 0, 0,
CU700ICRU
4.9, 5.9, 7, RADII FOR CENTRAL PART
0.0005, 0, ZTHICK, FLAG FOR LAYER 1 IN CENTRAL PART

```

0, 0, 0, 0,
CU700ICRU
0.013, 0, ZTHICK, FLAG FOR LAYER 2 IN CENTRAL PART
0, 0, 0, 0,
KAPTON700ICRU
0.33, 0, ZTHICK, FLAG FOR LAYER 3 IN CENTRAL PART
0, 0, 0, 0,
AIR700ICRU
0.005, 0, ZTHICK, FLAG FOR LAYER 4 IN CENTRAL PART
0, 0, 0, 0,
KAPTON700ICRU
0.0005, 0, ZTHICK, FLAG FOR LAYER 5 IN CENTRAL PART
0, 0, 0, 0,
CU700ICRU
0.11, 0, ZTHICK, FLAG FOR LAYER 6 IN CENTRAL PART
0, 0, 0, 0,
AIR700ICRU
0.0005, 0, ZTHICK, FLAG FOR LAYER 7 IN CENTRAL PART
0, 0, 0, 0,
CU700ICRU
0.005, 0, ZTHICK, FLAG FOR LAYER 8 IN CENTRAL PART
0, 0, 0, 0,
KAPTON700ICRU
0.38, 0, ZTHICK, FLAG FOR LAYER 9 IN CENTRAL PART
0, 0, 0, 0,
AIR700ICRU
0.013, 0, ZTHICK, FLAG FOR LAYER 10 IN CENTRAL PART
0, 0, 0, 0,
KAPTON700ICRU
0.0005, 0, ZTHICK, FLAG FOR LAYER 11 IN CENTRAL PART
0, 0, 0, 0,
CU700ICRU
0.39, 0, ZTHICK, FLAG FOR LAYER 12 IN CENTRAL PART
0, 0, 0, 0,
AIR700ICRU
0.005, 0, ZTHICK, FLAG FOR LAYER 13 IN CENTRAL PART
0, 0, 0, 0,
KAPTON700ICRU

```

0.0005, 0, ZTHICK, FLAG FOR LAYER 14 IN CENTRAL PART
0, 0, 0, 0,
CU700ICRU
0.12, 0, ZTHICK, FLAG FOR LAYER 15 IN CENTRAL PART
0, 0, 0, 0,
AIR700ICRU
0.0005, 0, ZTHICK, FLAG FOR LAYER 16 IN CENTRAL PART
0, 0, 0, 0,
CU700ICRU
0.005, 0, ZTHICK, FLAG FOR LAYER 17 IN CENTRAL PART
0, 0, 0, 0,
KAPTON700ICRU
0.32, 0, ZTHICK, FLAG FOR LAYER 18 IN CENTRAL PART
0, 0, 0, 0,
AIR700ICRU
0.013, 0, ZTHICK, FLAG FOR LAYER 19 IN CENTRAL PART
0, 0, 0, 0,
KAPTON700ICRU
0.0005, 0, ZTHICK, FLAG FOR LAYER 20 IN CENTRAL PART
0, 0, 0, 0,
CU700ICRU
0.7, 0.01, 0, 0,   chamber wall
CU700ICRU
0.7, 0.01, 0, 0,   gap
AIR700ICRU
0.7, 0.01, 0, 0,   container
CU700ICRU
0.69, 4.9, 0, ZTHICK, RCYS, FLAG FOR LAYER 1 IN BOTTOM PART
0, 0, 0, 0,
AIR700ICRU
0, 0, 0, 0,
CU700ICRU
0, MRNGE
***** start of CM SLABS with identifier phsp1 *****
30, RMAX
phsp1
1, NSLABS
17.29251, ZMIN

```

4.3374, 0, 0, 0, 0, 0

AIR700ICRU

***** start of CM MIRROR with identifier mirror *****

30, RMAX

mirror

21.63, 6.35, ZMIN, ZTHICK

4.7035, -4.3654, XFMIN, XBMIN

1, # LAYERS

0.00508, thickness of layer 1

0, 0, 0, 0,

MYLAR700ICRU

0, 0, 0, 0,

AIR700ICRU

0, 0, 0, 0,

AIR700ICRU

***** start of CM SYNCJAWS with identifier jaws *****

30, RMAX

jaws

2, 0, # PAIRED BARS OR JAWS, field type

Y

27.98, 33.54, 1.3976, 1.78571, -1.3976, -1.78571,

X

36.73, 42.32, 1.83467, 2.22428, -1.83467, -2.22428,

0.7, 0.01, 0, 0,

0, 0, 0, 0,

W700ICRU

0, 0, 0, 0,

W700ICRU

***** start of CM SYNCVMLC with identifier mlc *****

30, RMAX

mlc

0, 3, 0, ORIENT, NGROUP, MODE

48.290001, ZMIN

6.66, ZTHICK

0.535, 0.04, 0.04, 0.135, 0.369, 0.14, 48.290001, 48.570001, 51.560001, 51.770001,

0.249, 0.04, 0.04, 0.105, 0.146, 0.146, 48.390001, 48.650001, 49.550001, 49.880001,

0.25, 0.04, 0.04, 0.075, 0.14, 0.14, 48.290001, 48.410001, 51.680001, 51.710001,

10, 1

```

40, 2
10, 1
-10.35, START
0.0025, LEAFGAP
0, ENDTYPE
8, ZFOCUS or RADIUS of leaf ends
0, ZFOCUS of leaf sides
-20, 20, 28
-20, 20, 4
-20, 20, 28
0, 0, 0, 0,
AIR700ICRU
0, 0, 0, 0, 0,
W700ICRU
0, 0, 0, 0,
AL700ICRU
***** start of CM SLABS with identifier phsp2 *****
30, RMAX
phsp2
1, NSLABS
90, ZMIN
10, 0, 0, 0, 0, 0
AIR700ICRU
*****end of all CMs*****
#####
:Start MC Transport Parameter:

Global ECUT= 0.7
Global PCUT= 0.01
Global SMAX= 0.1000E+11
ESTEPE= 0.25
XIMAX= 0.5
Boundary crossing algorithm= EXACT
Skin depth for BCA= 0
Electron-step algorithm= PRESTA-II
Spin effects= On
Brems angular sampling= KM
Brems cross sections= NIST

```

```

Bound Compton scattering= On
Compton cross sections= default
Pair angular sampling= KM
Pair cross sections= BH
Photoelectron angular sampling= On
Rayleigh scattering= Off
Atomic relaxations= Off
Electron impact ionization= Off
Photon cross sections= PEGS4
Photon cross-sections output= Off

```

```
:Stop MC Transport Parameter:
```

```
#####
```

DOSXYZ Input file

The section below shows a DOSXYZnrc input file for the calculation of the dose distribution in water. It uses the previously calculated 6MV photon beam phase space file to calculate the dose distribution in DOSXYZnrc. The description of the parameters can be found in the DOSXYZnrc Users Manual [157].

```
watercalibration_FS10_6MV
```

```
1
```

```
H2O521ICRU
```

```
0.512, 0, 0, 0
```

```
5, 5, -1, 1
```

```
-15
```

```
-0.05
```

```
-0.025
```

```
0.025
```

```
0.05
```

```
15
```

```
-15
```

```
-0.05
```

```
-0.025
```

```
0.025
```

```

0.05
15
0
0.28, 125
1, 5, 1, 5, 1, 700, 1, 1
0, 0, 0, 0, 0, 0, 0, 0
0, 0, 0, 0, 0, 0, 0, 0
1, 5, 1, 5, 1, 700, 1, 0
0, 0, 0, 0, 0, 0, 0, 0
0, 2, 0, 0, 10, 180, 0, 10, 0, 0, 0, 0, 0
2, 0, 0, 0, 0, 0, 0, 0
/home/vision/egsnrc_mp/Results_rogerhead2_6.2/total_rogerhead2_6.2.egsphp1
30000000000, 0, 1000, 1247, 2721, 100.0, 0, 0, 0, 0, , 0, 0, 0, 1, 1
#####
:Start MC Transport Parameter:

Global ECUT= 0.512
Global PCUT= 0
Global SMAX= 5
ESTEPE= 0.25
XIMAX= 0.5
Boundary crossing algorithm= PRESTA-I
Skin depth for BCA= 0
Electron-step algorithm= PRESTA-II
Spin effects= On
Brems angular sampling= Simple
Brems cross sections= BH
Bound Compton scattering= Off
Compton cross sections= default
Pair angular sampling= Simple
Pair cross sections= BH
Photoelectron angular sampling= Off
Rayleigh scattering= Off
Atomic relaxations= Off
Electron impact ionization= Off
Photon cross sections= xcom
Photon cross-sections output= Off

```

:Stop MC Transport Parameter:

#####