Validating a convolution based radiotherapy treatment planning dose calculation

Dylan Mathew Cook

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VALIDATING A CONVOLUTION BASED RADIOTHERAPY TREATMENT PLANNING DOSE CALCULATION

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Thesis submitted for the degree of Master of Science (Medical Radiation Physics)

April 2018
DECLARATION

This dissertation is the result of my own work and includes nothing which is the outcome of work done in collaboration except where specifically indicated in the text. It has not been previously submitted, in part or whole, to any university of institution for any degree, diploma, or other qualification.

Signed: ____________________________________________

Date: __________________________________________________

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ABSTRACT

Dose targeting is critical in lung tumours that are treated with stereotactic ablative body radiotherapy (SABR). However, the small fields associated with these techniques in lung can cause increased lateral electron disequilibrium (LED), creating a reduction in the absorbed dose delivered to the tumour when compared with the treatment planning dose calculation.

The accuracy of the Philips Pinnacle ® collapsed cone convolution calculation in lung at densities 0.1g/cm³ to 1g/cm³ was investigated. A simple lung slab phantom was simulated for 10MV and 6MV photon beams, and for field sizes of 2cm x 2cm, 3cm x 3cm, 5cm x 5cm and 10cm x 10cm. Data from the calculation for lung depth dose and penumbral width (80% - 20%) were both investigated. Off axis dose profiles collected from these simulations were then benchmarked at 0.3g/cm³ lung density using Gaf EBT3 ® film in a CIRS® lung phantom. Depth dose measurements were also undertaken in the lung phantom with an IBA CC04 ion chamber and an Advanced Markus parallel plate ionization chamber. Results were then finally benchmarked against EGSnrc Monte Carlo calculations.

The depth dose trends in lung were as follows. The ionization chambers showed an over-response when compared with convolution, which in turn showed a slight over-response compared to Monte Carlo simulations for small fields sizes down to 2cm x 2cm. However, dose for these three methods converged as the field size increased and the amount of LED was reduced, with a close match for a 10cm x 10cm field size. The impact of lung density change on the central axis dose were estimated using the convolution calculation at three different densities. For a 3cm x 3cm field at 10MV, the percentage depth dose values in the mid lung were 55.4%, 62.6%, 66.9% for 0.2g/cm³, 0.3g/cm³ and 0.4g/cm³ densities respectively. The penumbral 80% - 20% widths were 1.12cm, 0.90cm, 0.80cm and 0.54cm for densities 0.2g/cm³, 0.3g/cm³, 0.4g/cm³, and 1.0g/cm³ respectively. The central axis deficits and penumbral flaring were also quantified at 6MV and were of lesser magnitude than at 10MV.

Lateral electron disequilibrium has a significant impact on dose coverage and needs to be considered for SABR applications. The convolution method showed good agreement with Monte Carlo and film simulations. There were only significant differences between
the convolution and Monte Carlo compared to ionization chamber measurements at very small field sizes, with the latter over-responding compared with Monte Carlo and convolution in lung.
ACKNOWLEDGEMENTS

I would like to extend my thanks to Prof Peter Metcalfe for his supervision during this project. His mentoring influence and encouragement were instrumental in helping me persist through a long and sometimes difficult research period.

I would also like to express my gratitude to all the other individuals and organisations that assisted me or enabled the pursuit of this project, A/Prof. Martin Carolan for giving up his time in order to allow use of the equipment at Illawarra Cancer Care Centre, as well as Dr. Bradley Oborn for his mentoring and Monte Carlo expertise and Trent Causer for his general troubleshooting.

Thanks also to my family and friends for your support.
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LIST OF ABBREVIATIONS AND ACRONYMS

- SCLC – Small cell lung cancer
- NSCLC – Non-small cell lung cancer
- DALY – Disability adjusted life years
- LED – Lateral electron disequilibrium
- EBRT – External beam radiotherapy treatment
- CT computed tomography
- 4D-CT 4-dimensional computed tomography
- 3D-CT 3-dimensional computed tomography
- TCP tumour control probability
- NSCLC non-small cell lung cancer
- 3D-CRT 3-dimensional conventional radiation therapy
- SBRT stereotactic body radiation therapy
- SABR stereotactic ablative body radiotherapy
- LINAC linear accelerator
- LED lateral electron disequilibrium
- TERMA total energy released per unit mass
- KERMA kinetic energy released only to charged particles per unit mass
- CPE charged particle equilibrium
- TCPE transient charged particle equilibrium
- TPS treatment planning system
- MC Monte Carlo
- MV Megavoltage
- CCC Collapsed Cone Convolution
- PDD percent depth dose
- dmax or Dmax dose at maximum depth
Chapter 1: Introduction

1 INTRODUCTION

1.1 Lung Cancer
Carcinomas of the lung represent the most common and potentially deadly forms of cancer among human beings. Of the 14.1 million new cases of cancer reported in 2012 alone, lung cancer constituted 1.8 million of these [1]. Lung cancers represent not only one of the highest incidence rates but highest rates of mortality and Disability Adjusted Life Years (DALYs). The prevalence of this disease worldwide can be attributed primarily to tobacco consumption, which was responsible for 71% of lung cancer deaths and 22% of all cancer deaths in 2012.

Lung cancers are a class of malignant growth characterized by uncontrolled cell growth occurring in the alveoli, trachea, bronchioles or bronchi. They can be divided into two primary classifications, small-cell (SCLC) and non-small-cell (NSCLC) carcinomas, with the latter constituting approximately 80% of lung cancers diagnosed [2]. SCLC and NSCLC vary considerably in factors such as doubling rate, growth fraction and their likelihood to metastasise. This in turn affects the forms of treatment that have been found to be effective. SCLC, for instance, could be described as more aggressive than NSCLC in terms of its growth and spread, but is comparatively quite receptive to treatments such as chemotherapy [3].

1.2 Treatment of Lung Cancer
Treatment methods for lung cancer include traditional surgery, chemotherapy, radiotherapy, immunotherapy and vaccine therapy. The effectiveness of each treatment
type can vary drastically depending on the type of cancer and specific patient circumstances, such as age, cancer staging and related illnesses. For instance, surgery has been found to be very effective in combating the slow growth rates of early stage NSCLC, with those having tumours removed successfully experiencing a tumour control probability of 90% [4]. However, due to the tendency for lung cancer to metastasise quickly, especially in SCLC, surgery is often ineffective and radiotherapy or chemotherapy are used. Of primary focus in this paper are the methods of radiotherapy, specifically stereotactic body radiotherapy, or SBRT.

1.3 Stereotactic Body Radiotherapy (SBRT)
Stereotactic body radiotherapy refers to an external beam radiotherapy treatment (EBRT) which aims to deliver extremely precise and biologically potent high doses of radiation to tumours in the chest, abdomen and pelvis, usually in one or only a few doses. Its origins lie in radiotherapy of the 1950s, where external markers and a stereotactic frame were placed on the head of the patient for the purposes of treating cranial cancers. With advancements in technology treatment margins can be reduced and stereotactic body radiotherapy is now the standard choice for inoperable stage I non-small cell lung cancers [5]. Varying the prescription methodology and utilizing very high doses for ablation of tumours necessitates the use of small fields. This in turn make the already vital role of accurate dose calculation even more important for use with SBRT and SABR (Stereotactic Body Ablative Radiotherapy), both because of the need to spare normal tissue from the ablative doses, and as discussed in 1.4, small fields can give rise to the effects of lateral electron disequilibrium.

1.4 Lateral Electron Disequilibrium (LED)
The density of human lung tissue can vary dramatically based on health factors and the respiratory patterns of the patient during treatment. Healthy lung tissue has a density of approximately 0.35g/cm³, which can decrease to as low as 0.1g/cm³ as a result of diseases such as emphysema [6]. Irradiation of the lung therefore is increasingly subject to the effects of secondary electron range and reduction in photon attenuation. This phenomenon, known as lateral electron disequilibrium, is expected to be especially pronounced for lower densities and smaller radiation field sizes, where secondary electrons displaced laterally are not replaced within the field area. This in turn leads to a
reduction in the dose absorbed in the lung, which is obviously very important to account for to ensure effective patient treatment.

1.5 Research Plans and Objectives

The global aims of this thesis are to further understand the effects of LED and the effects of tissue inhomogeneities on small field dosimetry in lung.

- **Aim 1:** Compare convolution calculations of dose to cylindrical and parallel plate ionisation chamber measurements in a lung phantom.

- **Aim 2:** Benchmark convolution calculations of dose off axis by using Gafchromic film in a lung phantom.

- **Aim 3:** Characterise depth dose and off axis dose in lung at different lung densities.

- **Aim 4:** Explain differences between ionisation chamber and convolution calculations for small field sizes by Monte Carlo simulation.
2 LITERATURE REVIEW

2.1 Conventional Radiation Therapy versus Surgery

For patients diagnosed with NSCLC, approximately 15% - 20% are found to have stage I disease [7]. Surgical resection can be a very effective treatment method for these patients, as early stage NSCLC has not metastasised, which affords the opportunity to remove the entire tumour from the patient with very good TCP (>90%) [8]. This is performed either via lobectomy or occasionally sublobar resection for higher risk patients [9]. However, these treatment options are not available to all patients. Long term smokers, the elderly or those with comorbidities such as heart disease or chronic pulmonary disease such as emphysema may not be able to withstand the invasive nature of open chest surgery. For these patients, conventional radiation therapy may be offered as an alternative treatment.

One such example of the conventional radiation therapy a NSCLC patient might undergo is 3D conformal radiation therapy (3D-CRT). 3DCRT involves treatment via the overlapping of several radiation beams incident at different angles, to create a highly conformed dose map which can spare skin and surrounding tissue. However, the low doses and therefore large number of treatments necessary for a treatment of this kind can be inefficient from a clinical perspective, as well as having less than optimal tumour control probabilities (overall survival rates as low as 20% - 50%) [7]. Dose escalation can result in better tumour control, but with increased risk of toxicity in surrounding healthy tissue the benefits of this option must be weighed carefully against the potential
negative health consequences for the patient. Doses of 20 Gy in the lung can result in the acute effects of radiation pneumonitis [10]. Radiation Oncologists and Medical Physicists therefore have sought out novel treatment options that could provide higher doses, to achieve better local control and overall survival rates, whilst still adhering to the absolute dose restrictions necessary.

2.2 Stereotactic Body Radiation Therapy (SBRT)

2.2.1 Advances in technology
Stereotactic radiosurgery (SRS) has its origins in the 1940s to 1960s [11]. It was originally used as method of treating brain cancers with singular high doses of radiation (e.g. 24 Gy) to ablate tumours. This set it apart from techniques which utilise standard 2 Gy per fraction radiotherapy such as 3D-CRT and IMRT. These are used to dose escalate conventional radiotherapy fractionation regimens (achieving tumour control through cell death and manipulation of repopulation rates).

Stereotactic ablative radiotherapy (SABR) techniques have been gaining popularity as a treatment for patients with inoperable early stage NSCLC. There are numerous challenges involved in translating SRS from brain to lung treatment, including the inhomogeneity of lung tissue and respiratory induced movement of the tumour. Patient movement during treatment will reduce the efficacy of the treatment plan, causing the dose distribution to move outside of the intended target margins, underdosing the tumour whilst delivering dangerous dose to surrounding healthy tissue. However, advances in real time image guidance techniques have allowed the delivery of high doses of radiation to these sensitive target volumes, whilst minimising the volume of high dose delivered to the organ at risk (OAR). Respiratory motion, for instance, can now be tracked very accurately. Instead of a conventional 3D-CRT regimen of 55-70 Gy over a 4 - 7 week period (typically 2 Gy/day), clinicians are now able to increase the dose per fraction to deliver 48-60 Gy in 2-5 fractions, which has numerous advantages [36].

The shorter treatment periods involved in SABR minimise the effect of tumour cell repopulation, but are also preferable from the perspective of running an efficient clinic. Hypoxic (oxygen deficient) tumours may experience reduced cell kill when treated with
standard fractionation and the ablative doses of SABR may also alleviate this issue. Numerous studies have demonstrated the improved tumour control probabilities possible through this treatment (>90% in some cases) [12]. A meta-analysis by Zhang et al [13] on 4850 patients showed mean survival rates for NSCLC at 1, 3 and 5 years to be 83.4%, 56.6% and 41.2% compared to 93.2%, 80.7% and 71.7% with surgical resection.

The advent of 4D planning/study sets have also assisted in reducing the risks in treating with ablative dose regimens. 4D-CT can now be used to map patient anatomy with respect to patient movement and respiration. Incorporating this information into the treatment planning system can allow for a highly customised treatment that can adjust the distribution of dose to match the position of the tumour during treatment [14].

The biological effect of tumours treated with these kinds of regimens are often > 100 Gy biological effective dose (BED), which is considered comparative to surgical removal, hence the term stereotactic ablative radiotherapy which has been coined by the radiotherapy community. The high doses involved in SABR treatment are delivered via small volume treatment fields. This in part contributes to an effect known as lateral electron disequilibrium, discussed in 2.4.

2.2.2 Dose Escalation
Numerous investigations into the potential for further dose escalation of SABR have been conducted, with Timmerman et al [15] being an example of pioneers on this topic. Lee et al [16] performed a retrospective study on 169 patients with stage I NSCLC [13 = 16]. Local control and 5-year survival rate were compared for a standard SBRT regimen of 48 Gy in 4 fractions (BED = 106 Gy) versus a regimen of 60 Gy in 4 fractions (BED = 150 Gy). It was observed that for tumours > 2cm size, the dose escalated regimen showed higher local control (76.2% vs 60.6%) and overall survival rate with no differences in treatment related toxicities.

2.3 Density Variations in Lung Tissue
Standard practice for radiotherapy treatment planning involves employing computed tomography (CT) of patient anatomy. The attenuation of X-rays produced by the CT X-
A ray generator for different tissue types will vary depending on their density and this information is transferred into the treatment planning system and converted from Hounsfield Units to mass density. The treatment of lung cancer is particularly challenging because of the inhomogeneity of lung and its proximity to many other organs at risk (OAR). Achieving an acceptable tumour control probability whilst also sparing dose to surrounding healthy tissue means it is vital to employ an accurate methodology of collecting patient specific anatomical information. The lung is subject to more density variations than any other part of the body, making it a difficult region for radiotherapy treatment planning. A lung density evaluation study by Van Dyke of 58 patients with lung disease [6] showed an almost linear decrease in density with increasing age, from an average of 0.348g/cm\(^3\) at age 5 to 0.193g/cm\(^3\) at age 80. Interestingly, though smoking can generate comorbidities that inhibit the use of surgery, there was found to be no significant statistical difference between the density of smokers and non-smokers lungs. Gender was also a non-correlated factor in lung density, however the mode of respiration was found to play a large role. Scans taken in the inspiration phase had a mean density of 0.2g/cm\(^3\) compared to 0.36g/cm\(^3\) in expiration.

**Figure 2-1:** Plot illustrating the difference in the density of lung with increasing age and expiration versus inspiration [6].
It can be seen then that the process of respiration should be carefully monitored during radiotherapy treatment planning to not introduce errors associated with the density variations seen above. At low densities and when using the small fields and high energies associated with SABR, the phenomena of LED is introduced, which is discussed more in the next section.

2.4 Lateral Electron Disequilibrium in the Lung

Early historical dose distribution modelling algorithms operated under the assumption of charged particle equilibrium, meaning that the range of secondary electrons originating via the interaction of photons and tissue is ignored [17]. The original effective depth method scaled primary beam with density and ignored photon scatter [18]. The scatter correction models, such as effective tissue air ratio method (ETAR) built into next generation algorithms calculated primary and scattered photon dose components using tissue-air ratio measurements obtained under equilibrium conditions. For beam energies in the MV range, and field sizes 5cm x 5cm or greater, this did not pose a major problem as the secondary electrons travel only a few mm before being absorbed and therefore do not significantly affect the dose modelling. However, the large doses used in SABR techniques necessitate the use of much smaller field sizes (5cm x 5cm and smaller). At these field sizes, and especially in the lung where the density of tissue can decrease lower than 0.2g/cm³, laterally displaced electrons can have ranges of up to several cm and can travel outside the treatment field. The consequence of this effect is that the secondary electrons are coming to rest and depositing dose outside the treatment volume defined by the treatment plan, and the dose in practice can be significantly lower than the modelling algorithm predicts if this LED is not accounted for. This in turn causes dose deficits in the central axis of the lung for small fields. [37]
Figure 2-2: Illustration representing the process of lateral electron disequilibrium. The secondary electron tracks for a thin field (shown right) have a great enough range to travel outside the treatment field and decrease the central axis dose [35].

The most accurate method for accounting and correcting for the effect of LED in the lung is with Monte Carlo simulations, as outlined by Disher (2013) [19]. By using Monte Carlo on a simulated lung phantom, the dose reduction in the lung caused by LED was quantified accurately and could be predicted using a relative depth dose factor (RDDF). The relevance of LED to clinical SABR treatment was also illustrated by using simulated energies from Co-60 (1.25MeV) up to 18MV, where lateral electron ranges become significantly longer. Field sizes were varied between a 15cm x 15cm field down to a 1cm x 1cm field. Density variation also played an important part in the experimental parameters as addressed in the previous section. The density of simulated lung was varied between 1g/cm³ and the extreme example of 0.001g/cm³. The results showed that for a 10cm x 10cm field the critical density at which central axis dose reduction was observed was 0.1g/cm³. Another example is evident when looking at the 3cm x 3cm field, where the critical density was seen to be only 0.2g/cm³ for a 6MV photon beam, or 0.3g/cm³ for an 18MV beam. The paper also goes on to introduce a novel technique which attempts to intentionally cause LED to utilise the steep dose gradients for dose sparing purposes.
Dose Deficits and Penumbral Flaring for Small Fields in the Lung: The Effect of Lung Density Changes

Figure 2-3: Monte Carlo simulations illustrating the effect of LED on the depth dose profiles within a lung slab phantom. Varying energies (1.25MV, 6MV and 18MV) and densities (0.001 – 1g/cm³) are represented for a 3cm x 3cm field [19].

2.5 Defining TERMA and Dose Kernels

The convolution calculation is benchmarked in this thesis hence description of the method is undertaken as follows. The content in Sections 2.5 and 2.6 are discussed extensively in Metcalfe, Kron and Hoban [20].

2.5.1 TERMA and Photon Fluence

The photons within a radiotherapy treatment beam interact with the patient’s tissue in an assortment of ways. It is useful to define a series of terms which can categorise and explain the way the dose of the initial beam is distributed within the patient. TERMA is a term referring to the total energy released per unit mass.
As a radiotherapy photon beam travels through the patient, a fluence of particles is observed. Planar fluence is defined as the amount of particles passing through a given area. As each photon possesses its own energy, then energy fluence can also be defined at any point within the patient (as MeV/cm²). The probability that any of these single photons interacts within the tissue of the patient is determined by another quantity, the linear attenuation coefficient, $\mu$ (cm⁻¹). This quantity is tissue dependent, being determined by the atomic number and density of the tissue, $\rho$. The total amount of energy released at any point within the patient can then be defined as the product of the mass attenuation and energy fluence at that point:

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

(2.1)

A subset of the total energy released is the KERMA$_c$, or the collisional kinetic energy released per unit mass. This defines the portion of the TERMA which is transferred into kinetic energy only to charged particles (such as electrons), and the subsequent amount of energy that those charged particles release as they travel along their secondary tracks. The local absorption of this KERMA$_c$ by the patient’s tissue is what ultimately contributes to the dose which is leveraged to treat cancer.

2.5.2 Dose Kernels
The second component used to calculate dose through convolution is the dose kernel (explained below). The values within the kernel are the energy deposited at a vectorial displacement from the interaction site as a fraction of the TERMA at that site. They are often separated into components used to calculated the primary dose and dose due to scatter.

2.6 Dose Calculation Algorithms
2.6.1 Convolution
The convolution model is a common method of dose calculation. One of the first physics based calculations of dose (which accounts for LED) was proposed in Mackie et al 1985 [21]. Varying algorithms exist [21, 22, 23] but all employ the same basic framework. The dose is calculated using two components, the energy deposited by the interactions of primary photons (TERMA) and the dose deposited in the area surrounding the primary photon interaction site (the kernel). The area of interest is
separated into discrete volume elements (called voxels) overlayed with a vectorial framework. The dose in each element as a result of the interaction at the primary site can now be calculated and summed together to give a measure of the total dose imparted due to the photon interaction, as well as the distribution of that dose within the kernels around the primary site.

![Diagram](image)

**Figure 2-4:** Vectorial framework representing the convolution process. A ray originating from the source travels through the surface of the patient at point $r_0$, before reaching the primary interaction site at $r'$. The dose is deposited at point $r$ and is the sum of all the TERMA from all interactions along the path $r' - r$ [20].

To derive an expression for the dose, it is said that if the TERMA at a given point $r'$ is $T(r')$, the energy deposited in a unit volume at another point $r$ due to $T(r')$ is given by $T(r')H(r - r')$, where $H(r - r')$ is the kernel value (primary plus scattered) for a displacement $r - r'$ from the kernel origin. Now the total dose at point $r$ can be found by integrating over all unit masses within the irradiated volume:

$$D(r) = \int_{r_0}^{r} T(r') [H_p(r - r') + (H_s(r - r'))] d^3r' \quad (2.2)$$
2.6.2 Inhomogeneous medium

A more rigorous method of calculation must be employed in reality, as the patient’s tissue is inhomogeneous (i.e. it consists of numerous different tissue types with their own respective densities). The true fractional energy distribution at each primary interaction site will be dependent on the position of the site, therefore the kernels will be a function of \( r' \) as well as \( r - r' \). Taking this and the variable densities of different tissues into account, the new expression for dose becomes:

\[
D(r) = \frac{1}{p(r)} \int_{r-r'}^r T(r') \rho(r') \left[ H_p(r', r - r') + H_s(r', r - r') \right] d^3r'
\]  

(2.3)

The calculation of these kernel values needs to strike a balance between accuracy and computation time. It can therefore be assumed the energy lost by secondary electrons along the path of \( r' \) to \( r \) is dependent on their effective path length and hence average density, given as:

\[
\rho_{av} = \frac{1}{|r-r'|} \int_{r-r'}^r \rho(r - r') d(r - r')
\]  

(2.4)

This quantity is found using ray tracing through the voxels in between \( r \) and \( r' \). Even though the path between two points in the medium may consist of numerous different densities, this method still provides a good estimation for the primary kernel since the dose is also dependent on the density distribution and not only the average density. The assumption becomes even more accurate for the scatter kernel, as it can be seen that the scattered photon fluence along the path \( r' \) to \( r \) is exactly proportional to the average density.

Using the above assumptions, the kernel values can now be obtained using the following equation:

\[
H(r', r - r') = H(\rho_{av}, r - r') \frac{\rho(r)}{\rho_{av}}
\]  

(2.5)

And substituting this into equation (2.3), the final expression for the convolution when considering an inhomogeneous medium becomes:
Dose Deficits and Penumbral Flaring for Small Fields in the Lung: The Effect of Lung Density Changes

\[ D(r) = \int_{r'} T(r') \frac{\rho(r')}{\rho_{av}} [H_p(\rho_{av}, r - r') + H_s(\rho_{av}, r - r')] d^3 r' \]  

(2.6)

This equation assumes that the mass density is proportional to the energy distribution whereas this is not strictly correct. A more complete version of this expression for convolution would use the electron density, as most energy loss occurs to the deposition due to secondary charged particles and Compton scattering by photons. In the example of a photon travelling between bone and soft tissue the ratio of mass density to electron density is not proportional as it is in water and the density cancellation made in (2.6) cannot take place. Using electron density \( \rho_e^w(r) \) instead of mass density, the expression becomes:

\[ D(r) = \frac{1}{\rho(r)} \int_{r'} T(r') \rho_e^w [H_p(\rho_{av}, r - r') + H_s(\rho_{av}, r - r')] \frac{\rho_e^w(r')}{\rho_{av}} d^3 r' \]  

(2.7)

2.6.3 Collapsed Cone Convolution

A refined method of the convolution model emerged in 1987 through the work of Ahnesjo et al [12 = 23]. Called the collapsed cone convolution algorithm (CCC), it utilises polyenergetic TERMA and kernels and combines the primary and scattered contributions of dose. The kernel components are characterised using the expression:

\[ H(r, \theta) = \frac{A_\theta e^{-a\theta r} + B_\theta e^{-b\theta r}}{r^2} \]  

(2.8)

Where \( \theta \) is a finite polar angle with respect to the primary beam. The main difference is in the definition of the primary interaction site, which is now considered to be the apex of a cone, constructed out of a series of radially directed lines of polar angle \( \theta \). The kernel functions are calculated along these lines in the same way that they were calculated along the path \( r' \) - \( r \) in the previous section. However now the kernel values are the amount of energy deposited within the cone at radius \( r \), collapsed onto the line.

CCC has been shown to have vastly improved computation speed when compared to standard convolution, the main reason it is used widely in clinical practice today. The calculation time for CCC is proportional to \( MN^3 \), where \( M \) is the number of cones and \( N \) is the number of voxels along one dimension of the calculation volume. By contrast,
standard convolution as introduced by Mackie et al [11] in 1985 is proportional to $N^6$, meaning CCC offers not only faster computation by also greater accuracy when heterogeneous tissue is present.

The CCC method employed in the Pinnacle\(^3\) treatment planning system (TPS) was implemented by Papanikolaou et al 1993 [23].

2.6.4 Monte Carlo

Monte Carlo is an alternative method of dose distribution mapping. It uses a first principles calculation to stochastically simulate particle transport from a primary interaction site. As opposed to convolution, which averages the effects of large groups of particles into kernels, in Monte Carlo the history of each individual particle is simulated and tracked within a dose spread array and convolved with the kinetic energy along the particle’s path to calculate the primary dose distribution. The probability of different interactions such as Compton scattering, photoelectric effect and pair production are weighted based on the energy and randomly assigned to each particle. The energy deposition in each voxel is then scored and combined to produce a final dose distribution [21].

![Figure 2-5: Monte Carlo results compared to physical measurement using 0.6cm\(^3\) Baldwin Framer ion chamber for a 5cm x 5cm 10MV beam in lung [20].](image)

This method of dose calculation has been shown to have a high degree of accuracy [20] over effective depth and ETAR algorithms, especially in predicting the large reduction in central lung dose caused by LED. However, the amount of particle histories required to produce a good estimate of the dose distribution can result in computational times
which may exclude its use in a clinical environment. As processing power made available to clinics improves, this method is finding more common use in commercially available treatment planning systems, but at this stage other calculation methods are still more widely used. A study by Guckenberger et al 2015 [24] showed that for 582 patients treated with SBRT across 13 institutions, 36% of the plans were generated using pencil beam, 31% CCC, 15% Antistropic Analytical Algorithm (AAA) and only 5% using Monte Carlo.

Fast Monte Carlo systems such as VMC++ are starting to be used in commercial systems [25]. Access to such a code was not available for this thesis however some benchmark calculations using GEANT4 Monte Carlo are reported in chapter 4.
3 METHODS

3.1 CT Numbers
As discussed previously, the main parameters governing the magnitude of the lateral electron disequilibrium effect are the tissue density, beam energy and field size. CT images used for treatment planning distinguish between tissue types using the Hounsfield unit scale, which transforms the linear attenuation of each tissue type into a scale where the radiodensity of water at STP is 0 HU whilst air is -1000HU. Subsequent Hounsfield values (i.e. CT numbers) can then be derived via the following equation:

\[
H_U = 1000 \times \frac{\mu - \mu_{\text{water}}}{\mu_{\text{water}} - \mu_{\text{air}}}
\] (3.1)
Figure 3-1: Replication of a Phillips Pinnacle treatment planning calibration curve of CT number + 1000 (x-axis) versus density (y-axis).

However, Pinnacle uses CT numbers +1000 rather than CT number directly as they go from 0 to 4000, which saves memory that would otherwise be spent on negative integers. To this end a calibration curve is required to convert the values from CT# +1000 to mass density (shown in Figure 3.1(a)).

Assuming Compton interactions dominate then the density of electrons will have the greatest effect on the deposited dose so ideally this quantity should be used in the TPS. However, a CT directly measures physical attenuation, which necessitates the use of a correction curve in Pinnacle to convert physical density to electron density when calculating dose. The gradient change 1000 to 2200 is designed to account for bone.

Using experimental data, the relationship between CT number and electron density can be expressed as:

\[ \rho_e^w = R_\gamma \left( \frac{1}{1000} N_{CT} \right) + 1 \]  

(3.1)

Where \( R_\gamma \) represents a regression line gradient for different substances. \( R_\gamma \) equals approximately 1 for air, lung and adipose tissue, and 0.5 for bone. Hence, for lung, a CT number of -700 equates to a density of 0.3g/cm\(^3\).
The opacity of tissue to ionising radiation is based on its density and therefore in a Pinnacle phantom we can create different tissue types by simply varying the density within the desired region of interest (ROI). The density parameters of the normal tissue were therefore assigned a value of 1g/cm³ for solid water. The density of the lung was assigned a value of 0.3g/cm³ initially and was varied for later experimentation (see 3.5). Furthermore, the mean energy output of the simulated beam was set to either 6MV or 10MV, whilst the field size was manipulated by varying the dimensions of the x and y jaws between 2cm x 2cm, 3cm x 3cm, 5cm x 5cm or 10cm x 10cm. A standard single beam with perpendicular entry to phantom surface was employed and the resolution of the dose grid was assigned a value of 0.1cm x 0.1cm x 0.1cm, to ensure the highest degree of dose gradient sensitivity whilst balancing out with acceptable calculation times. Collapsed cone convolution was chosen as the calculation method.

3.2 Virtual and Real Solid Water Phantoms
A virtual lung phantom was generated using Phillips’ Pinnacle³ treatment planning system contour tools, which mimicked the experimental setup of a lung slab consisting of solid water and lung analogue. The simulated phantom shown in 3-2 consisted of; 5cm depth of water (simulating chest wall ignoring ribs). This is referred to in the thesis as “tissue”, followed by 10cm depth of lung tissue equivalent space, followed by another 10cm depth of water (simulating beam exit). For comparison of measured and calculated results, a real phantom was used consisting of several interchangeable layers of CIRS solid water of density 0.3g/cm³. The 0.3g/cm³ lung was CT scanned on a Siemens CT scanner and ROI samples CT number \( \bar{X} = -700 \text{ HU} \) with standard deviation \( \bar{\sigma} = \pm 30 \text{ HU} \). This places them on the Pinnacle calibration line shown in Figure 3-1.

3.3 Script to Extract Dose
A custom script (not supplied by the manufacturer) was implemented into the treatment planning system. It functioned using the Point of Interest (POI) feature of Pinnacle. The Source to Surface Distance (SSD) of the phantom setup was defined using a point of interest placed along the central axis and at 0cm depth. When the script was run, the program generated a line between the second and third POIs within the plan and
collected the dose data at every point according to the dose grid resolution. The raw data file was then transferred into the Pinnacle filing directory where it was accessed as a text file and processed into a more legible form as seen in section 4. This scripting acquisition method was found to be versatile as POIs could simply be moved from a vertical to horizontal orientation to collect off axis dose measurements. Both off axis and percentage depth dose measurement data was acquired for all possible variations of the given parameters, 64 dose profiles in total.

![Figure 3-2: Screen capture of the Pinnacle treatment planning system lung phantom setup. POI 1 and 2 (green and blue respectively) are placed to acquire an off axis ratio measurement.](image)

### 3.4 Benchmarking of depth dose data using ionisation chambers

The percentage depth dose portion of the Pinnacle phantom data was benchmarked against the same CIRS phantom setup as in section 3.2. However rather than Gaf-chromic film, an IBA CC04 ionisation chamber with cavity length 3.6mm, radius 2.0mm and wall thickness 0.7g/cm², together with an Advanced Markus parallel plate chamber were employed. The electrometer based current readings were then collected and normalised to the charge readings at $d_{max}$. By manipulating the order of the phantom and solid water layers, the thimble chamber was placed at depths of 1.2cm or 2.1cm ($d_{max}$, depending on the beam energy used), 4, 6, 9, 11, 14, 15, 16, and 20cm. The secondary build up gradient present at the lung-tissue interface (15cm depth) is a good indication of the degree to which lateral electron disequilibrium has occurred. To obtain a more sensitive measurement at this interface an Advanced Markus parallel plate...
ionisation chamber was employed. Markus measurements were taken at both polarities and the charge measurements were averaged. An identical experimental setup was used, with the chamber placed at depths immediately following the interface, at 15, 15.1, 15.3, 15.6, 16, 16.5 and 17.2cm. All measurements were conducted in an inplane orientation and radial axis. Air cavity and chamber wall backscatter effects can lead to perturbation of the beam, but the large guard ring around the cavity of the Markus chamber acts to minimise this.

3.5 Benchmarking off-axis data using EBT3 Gaf-chromic film

The Pinnacle lung phantom data was benchmarked against EBT3 Gaf-Chromic film using a setup consisting of a combination of CIRS lung phantom slabs and solid water explained in 3.2. The geometry of the phantom was maintained in this physical version of the experiment, with tissue-lung interface placed at 5cm depth and lung-tissue interface placed at 15cm depth. A region of solid water of 10cm depth was placed at the bottom of the phantom geometry to generate the effects of backscatter and dose buildup. The film samples were measured to be slightly larger than the field size, to account for field divergence. For the 3cm x 3cm field, a 6cm x 6cm piece of film was prepared. 8cm x 8cm was used for the 5cm x 5cm field and 13cm x 13cm for the 10cm x 10cm fields.

![Figure 3.3: CIRS lung and solid water phantom.](image)

The samples were then pre-scanned on Epson 10000xl scanner with a resolution of 72dpi to obtain an un-irradiated reference value for comparison. Each film was scanned 6 times, and the last data from the 3 scans of each were recorded and averaged out during analysis. The film samples were then separately placed between layers of the...
CIRS phantom slab, at interfaces representing both the middle and bottom of the lung. This was done to compare the off-axis profiles against one another and different depths and account for the effect of field divergence. For each film sample, the phantom was irradiated with 200MU on a Varian 21EX linear accelerator. Each sample corresponds to one of the various parameter combinations (field size and energy) as described in the previous section, for both middle and bottom lung. Dose at different lung densities were not investigated with ion chamber or film using this method as the only lung density of the CIRS phantom was 0.3g/cm³. 20 control film samples were also prepared and used to create a calibration curve (seen in figures 3-4 and 3-5) specific to the accelerator used for this experiment.

3.6 Film Analysis

Post exposure, all film samples were re-scanned and the images were analysed using ImageJ software. Registration between pre-scan and post-scan images was necessary to account for any movement in either the scanner or phantom between setups. The raw exposure values were plotted and applied to the calibration curve as seen in figure 3-4 in order to obtain the off-axis dose profiles seen in section 4.

Figure 3-4: Calibration curve based on film data.
3.7 Comparison of data with Monte Carlo simulation

As discussed in Chapter 2, Monte Carlo is the most rigorous method for dose distribution calculation available. In order to further validate the results of the previous experiments, a modified GEANT4 Monte Carlo based software dubbed BradCalc was used to simulate the lung phantom setup. A dose voxel resolution of 2mm x 2mm x 2mm was used. The Pinnacle dose cube was exported into the program and used to generate simulations mirroring the convolution. The depth dose profiles were then extracted and the data normalised to $D_{\text{max}}$.

GEANT4 version 10.00 was used to model the dose deposition inside the virtual phantom, and the beam was modelled off a Varian 2100C linear accelerator using the EGSnrc user code BEAMnrc. This was also used for the physical ion chamber and film measurements. The component modules used for modelling of the treatment head included x-ray target (SLABS), primary collimator (CONS3R), exit window (SLABS), flattening filter (FLATFILT), ion chamber (CHAMBER, CHAMBER), mirror (MIRROR), Y and X-jaws (JAWS, JAWS), Millenium 120 leaf MLC (DYNVMLC), and the reticle (SLABS). For all BEAMnrc simulations the global ECUT= 0.521 MeV, and global PCUT= 0.01 MeV. The low ECUT value is important for tracking secondary electrons in the monitor chamber accurately [26].
The physics processes modelled by this simulation included multiple scattering, photoelectric effect, Compton scattering, gamma conversion, ionization, Bremsstrahlung, and positron annihilation.

3.8 Density Variation

After acquiring data at the standard density of 0.3g/cm$^3$ (for comparison with other methods such as film and ion chamber), the density of the lung was varied within Pinnacle’s density override toolkit to values of 0.2g/cm$^3$, 0.4g/cm$^3$ and 1g/cm$^3$. Results for both depth dose and off-axis in the mid lung (10cm) and at the lung-tissue interface (15cm) were re-acquired and plotted together. This served to illustrate the effect of changing lung density on the amount of dose reduction and penumbral flaring due to LED.
4 RESULTS

4.1 Percentage Depth Dose Comparison of Three Modalities; Monte Carlo, Collapsed Cone Convolution and Ion Chamber

The figures shown in section 4.1 are the percentage depth dose curves calculated using collapsed cone convolution in Pinnacle and the virtual phantom geometry described in section 3. An SSD of 100cm and a 1mm x 1mm x 1mm voxel size was used. The beam energy was varied between 6MV and 10MV, the field size between 3cm x 3cm, 5cm x 5cm and 10cm x 10cm, and the lung density was varied between 0.2, 0.3, 0.4 and 1g/cm³. A custom script was used to extract the depth dose data. This was compared with an equivalent simulated Monte Carlo PDD and a data series of combined (0.4cc) ionization chamber results and Advanced Markus ionization chamber results.
Figure 4-1 illustrates best the effect of combining a relatively high beam energy and small field size on the dose reduction associated with lateral electron disequilibrium. There is a steep dose gradient present starting from the tissue-lung interface at 5cm depth and the PDD for the CCC acquisition in the mid lung is approximately 20% lower than the equivalent data set taken for a 10cm x 10cm field. Following the lung-tissue interface at 15cm depth there is a sharp secondary build up region, which the high spatial resolution of the Advanced Marcus parallel plate chamber was more suited to capturing accurately. The secondary build-up region beyond the lung continues for approximately 1cm before peaking and returning to what could be considered a normal depth dose profile, free from the effects of LED.

There can also be seen to be a difference between the CCC and ion chamber for these parameters. The first three data points for the pin point (0.4cc) ion chamber are in good agreement with the convolution but as the profile penetrates deeper into the lung and is subject to LED there is a noticeable discrepancy of approximately 5% for the ion chamber data points at 9cm, 11cm and 14cm. This disagreement continues to a lesser extent (approximately 1% difference) for the Markus chamber buildup before returning to a good agreement outside the lung.
As expected in the steep dose gradient region, the Monte Carlo simulation sits 6% lower than the convolution. In the mid lung, the inaccuracy of the convolution decreases to approximately 3%. The physics behind this effect will be discussed in section 5. The inaccuracy of the ion chamber is also quite large here (6.7%).

Shown in figure 4-2 is the percentage depth dose calculated by the collapsed cone convolution (CCC) calculation, but for a lower energy, namely, 6 MV with a 2cm x 2cm field size. This is compared with Monte Carlo simulated depth dose, (0.4cc) ionization chamber results and advanced Markus ionization chamber results. All data has been normalized to 100% at $d_{\text{max}}$.

**Figure 4-2:** Percentage depth dose curve comparing Monte Carlo, Pinnacle convolution and ion chamber in lung phantom for a 2cm x 2cm field and beam energy of 6MV.

Figure 4-2 also exhibits significant LED dose reduction along the central axis. The magnitude of dose reduction compared to the previous figure is lesser, illustrating that higher beam energy increases the amount of LED occurring. The ion chamber over-response discussed in the previous figure is also present but the magnitude of the difference between ion chamber and convolution is less, approximately 1% PDD. The ion chamber is also more accurate at this lower energy in comparison to the Monte Carlo (4.3% difference).
Figure 4-3: Percentage depth dose curve comparing Pinnacle convolution and ion chamber in lung phantom for a 3cm x 3cm field and beam energy of 10MV.

Figure 4-3 exhibits a similar amount of LED as the previous figure, which had a smaller field size but lower energy, highlighting the optimal combination of parameters required to manipulate the LED effect. The disagreement between ion chamber and CCC here is still present, and slightly higher than the preceding figure. There is a similar over-response of approximately 2% between the convolution and the Monte Carlo simulation (see Table 4-3 for more accurate numerical analysis).
Chapter 4: Results

Figure 4-4: Percentage depth dose curve comparing Pinnacle convolution and ion chamber in lung phantom for a 3cm x 3cm field and beam energy of 6MV.

The dose deficit continues to decrease with decreasing beam energy in Figure 4-4 as expected. The ion chamber/convolution discrepancy is still present in the mid lung but has decreased to a difference of approximately 1%. The steep dose gradient region in the beginning of the lung is also less noteworthy now, and the convolution dose compared to Monte Carlo is consistently 1-2% higher throughout the lung.
Figure 4-5: Percentage depth dose curve comparing Pinnacle convolution and ion chamber in lung phantom for a 5cm x 5cm field and beam energy of 10MV.

Figure 4-5 is very similar to the preceding figure, suggesting that specifically the variation between 6MV and 10MV beam energy and 3cm x 3cm to 5cm x 5cm field size produce a similar amount of dose reduction despite accomplishing this in different ways.
Figure 4-6: Percentage depth dose curve comparing Pinnacle convolution and ion chamber in lung phantom for a 5cm x 5cm field and beam energy of 6MV.

Figure 4-6 is noteworthy for appearing visually at least to be the first of this section which does not exhibit a noticeable amount of dose reduction through the lung phantom, indicating that field sizes less than 5cm x 5cm with energies greater than 6MV show the largest dose deficits due to LED. The disagreement between the ion chamber and CCC curves are also within close agreement for the first time.
Figure 4-7: Percentage depth dose curve comparing Pinnacle convolution and ion chamber in lung phantom for a 10cm x 10cm field and beam energy of 10MV.

Figure 4-8: Percentage depth dose curve comparing Pinnacle convolution and ion chamber in lung phantom for a 10cm x 10cm field and beam energy of 6MV.

Figure 4-7 and 4-8 are control measurements. At these field sizes, the laterally displaced secondary electrons cannot reach outside the edge of the field in large enough numbers to cause a reduction in dose at the central axis. Therefore, LED is not present in these
data sets and they can be considered standard depth dose profiles, which makes them useful in quantifying LED in the other figures. Note the very good agreement between both ion chamber and CCC data sets in these figures. The error of the convolution with respect to the Monte Carlo has also reduced below 2% at 10cm x 10cm for a 10MV beam.

Table 4-1: Difference (as a % of Dmax) between convolution and CC04 ion chamber depth dose in mid-lung (depth 11cm).

<table>
<thead>
<tr>
<th>Field size</th>
<th>% Difference (6MV)</th>
<th>% Difference (10MV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>2cm x 2cm</td>
<td>1.96</td>
<td>4.32</td>
</tr>
<tr>
<td>3cm x 3cm</td>
<td>1.82</td>
<td>3.44</td>
</tr>
<tr>
<td>5cm x 5cm</td>
<td>0.77</td>
<td>2.05</td>
</tr>
<tr>
<td>10cm x 10cm</td>
<td>0.21</td>
<td>0.45</td>
</tr>
</tbody>
</table>

Table 4-1 illustrates that there was a good agreement between ion chamber and convolution depth dose curves for most combinations of energy and field size parameters. The largest disagreement was approximately 4% (see section 5.5 for discussion about limits of accuracy) and occurred for the setup with the largest amount of LED (2cm x 2cm, 10MV), indicating that perhaps the convolution is underestimating the dose, hence overcompensating for the LED effect. In general, the higher energy of 10MV seemed to have a greater effect on the agreement between the two data sets than the field size.

Table 4-2: Difference (as a % of Dmax) between convolution and Advanced Marcus ion chamber depth dose at lung-tissue interface (depth 15cm).

<table>
<thead>
<tr>
<th>Field size</th>
<th>% Difference (6MV)</th>
<th>% Difference (10MV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>2cm x 2cm</td>
<td>2.43</td>
<td>3.75</td>
</tr>
<tr>
<td>3cm x 3cm</td>
<td>1.58</td>
<td>3.26</td>
</tr>
<tr>
<td>5cm x 5cm</td>
<td>0.58</td>
<td>2.19</td>
</tr>
<tr>
<td>10cm x 10cm</td>
<td>0.49</td>
<td>0.59</td>
</tr>
</tbody>
</table>
Similar trends are observed when the difference between the convolution and Advanced Markus chamber are measured at the lung-tissue interface at 15cm.

**Table 4-3:** Difference (as a % of Dmax) between Monte Carlo depth dose and both ion chamber and convolution depth dose in the mid lung (11cm depth).

<table>
<thead>
<tr>
<th>Field size</th>
<th>IC-MC % Difference (6MV)</th>
<th>CCC-MC % Difference (6MV)</th>
<th>IC-MC % Difference (10MV)</th>
<th>CCC-MC % Difference (10MV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>2cm x 2cm</td>
<td>4.39</td>
<td>2.43</td>
<td>6.79</td>
<td>2.47</td>
</tr>
<tr>
<td>3cm x 3cm</td>
<td>4.19</td>
<td>2.38</td>
<td>6.03</td>
<td>2.59</td>
</tr>
<tr>
<td>5cm x 5cm</td>
<td>3.18</td>
<td>2.41</td>
<td>4.42</td>
<td>2.28</td>
</tr>
<tr>
<td>10cm x 10cm</td>
<td>2.72</td>
<td>2.51</td>
<td>2.23</td>
<td>1.77</td>
</tr>
</tbody>
</table>

The results of Table 4-3 show the CC04 ion chamber possesses a surprising lack of accuracy when compared to the ‘true value’ of the Monte Carlo simulation. The difference as a percentage of $D_{max}$ between the ion chamber and Monte Carlo peaks at 6.79% (at 11cm depth) and decreases for lower energies and larger field sizes as the effects of LED decrease. The rest of the results, however, fall within the 2-5% range of difference which is being considered the limits of accuracy in this thesis. In general, the trends of this table indicate that accuracy of the ion chamber is a function of energy and field size, and that the discrepancy between both depth dose curves decreases along with the decreasing effect of LED. The ion chamber also appears to overrespond significantly for small fields (2cm x 2cm and 3cm x 3cm) at 10MV, with inaccuracies of approximately 6% recorded when compared to the Monte Carlo. The accuracy of the convolution appears to remain unchanged appears with respect to beam energy, but decreases slightly for the 10MV beam when the field size is increased, becoming statistically very accurate (less than 2% difference) for a 10cm x 10cm field.

4.2 Off-axis dose measurements, comparison of convolution and Gafchromic film
Whilst the depth dose helps characterise inverse plan calculations, if dose delivered matches dose prescribed inverse plan calculations may rely on target coverage, hence validating dose profile accuracy is also critical to ensure predicted target coverage is occurring due to valid estimation of the penumbral profile shape and width.

Shown in figure 4-9 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 6 MV for a 3cm x 3cm field size in the mid lung (depth 11cm). This is compared with EBT3 Gafchromic film results. All data has been normalized to 100% at $d_{\text{max}}$.

![Figure 4-9: Off-axis dose profile, comparison of convolution and gafchromic film for a 3cm x 3cm field at 6MV energy and in the mid lung (11cm depth).](image)

Figure 4-9 illustrates an excellent agreement between the off-axis dose profiles acquired using CCC and Gafchromic film for a 3cm x 3cm field at 6MV and in the mid lung. Though the pre-scan and post-scan images have been registered to improve accuracy, there is still some noticeable noise present in the film profile compared to the smooth curve of the CCC profile.
Figure 4-10: Off-axis dose profile, comparison of convolution and gafchromic film for a 3cm x 3cm field at 10MV energy and in the mid lung (11cm depth).

Figure 4-10 exhibits slightly more noise along the dose plateau and therefore sits slightly below the CCC curve. The negative distance side of the profile is slightly wider than the positive though this may be a due to misalignment (see discussion). Very good agreement is still observed in the penumbra, which is marginally more spread out than the 6MV graph at this field size seen in 4-9, as expected.

Figure 4-11: Off-axis dose profile, comparison of convolution and gafchromic film for a 3cm x 3cm field at 6MV energy and at the lung-tissue interface lung (15cm depth).
Shown in figure 4-12 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 10 MV for a 3cm x 3cm field size at the lung-tissue interface (15cm depth). This is compared with EBT3 Gafchromic film results. All data has been normalized to 100% at $d_{\text{max}}$.

![Graph of normalized dose vs. off-axis distance](image)

**Figure 4-12:** Off-axis dose profile, comparison of convolution and gafchromic film for a 3cm x 3cm field at 10MV energy and at the lung-tissue interface lung (15cm depth).

Figure 4-11 and 4-12 are very similar to their mid lung equivalents in Figure 4-9 and 4-10 as expected. There is still very good agreement between the film and CCC however as the beam has travelled a larger distance field divergence is observed and the profile is slightly more spread out. The major discrepancies occurring in the plateau region due to noise within the film results affecting the normalization.
Figure 4-13: Off-axis dose profile, comparison of convolution and gafchromic film for a 5cm x 5cm field at 6MV energy and in the mid lung (11cm depth).

Figure 4-13 illustrates well the increase in penumbral flaring based only on beam energy variation when compared to 4-14. Despite having the same field size, the 6MV profile is clearly observed to possess a flatter and wider dose plateau, with less penumbral flaring. This is indicative with less LED as is expected.

Figure 4-14: Off-axis dose profile, comparison of convolution and gafchromic film for a 5cm x 5cm field at 10MV energy and in the mid lung (11cm depth).
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Moving to a 5cm x 5cm field, a clearly wider beam profile can be observed. There is still very good agreement between film and CCC measurements.

**Figure 4-15:** Off-axis dose profile, comparison of convolution and gafchromic film for a 5cm x 5cm field at 6MV energy and at the lung-tissue interface (15cm depth).

Shown in figure 4-16 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on pinnacle at 100cm SSD for 10 MV for a 5cm x 5cm field size at the lung-tissue interface (15cm depth). This is compared with EBT3 Gafchromic film results. All data has been normalized to 100% at d$_{max}$.

**Figure 4-16:** Off-axis dose profile, comparison of convolution and gafchromic film for a 5cm x 5cm field at 10MV energy at the lung-tissue interface (15cm depth).
Again, the bottom lung profiles offer little new information except to reaffirm the conclusions of their mid-lung equivalent and introduce more field divergence as the beam travels further away from the source.

**Figure 4-17:** Off-axis dose profile, comparison of convolution and gafchromic film for a 10cm x 10cm field at 6MV energy in the mid lung (10cm depth).

**Figure 4-18:** Off-axis dose profile, comparison of convolution and gafchromic film for a 10cm x 10cm field at 10MV energy in the mid lung (10cm depth).
**Figure 4-19:** Off-axis dose profile, comparison of convolution and gafchromic film for a 10cm x 10cm field at 6MV energy at the lung-tissue interface (15cm depth).

**Figure 4-20:** Off-axis dose profile, comparison of convolution and gafchromic film for a 10cm x 10cm field at 10MV energy at the lung-tissue interface (15cm depth).

The 10cm x 10cm graphs are again used as control measurement as it has been established that LED is not expected to occur for field sizes larger than 5cm x 5cm. One potential error in the experimental setup observed in these graphs is that the 13cm x 13cm size of the film chosen for measurements at this field size was most likely not appropriate. The divergence of the field extends past the edge of the film, and the exposure on the film was not enough to represent anything lower than 20% of the maximum dose.
Figure 4-21: Off-axis dose profile, comparison of convolution and gafchromic film for a 5cm x 5cm MLC field at 6MV energy in the mid lung (10cm depth).

Figure 4-22: Off-axis dose profile, comparison of convolution and gafchromic film for a 5cm x 5cm MLC field at 10MV energy in the mid lung (10cm depth).
Figure 4-23: Off-axis dose profile, comparison of convolution and gafchromic film for a 5cm x 5cm MLC field at 6MV energy at the lung-tissue interface (15cm depth).

Figure 4-24: Off-axis dose profile, comparison of convolution and gafchromic film for a 5cm x 5cm MLC field at 10MV energy at the lung-tissue interface (15cm depth).

Figures 4-23-24 all still exhibit good agreement between film and convolution. The penumbra has been noticeably flattened out by the introduction of the multi-leaf collimator as opposed to the jaws used in the rest of the experiment.

Table 4-4 – Comparison of 80%-20% penumbral distances off axis for film and convolution off axis dose profiles. 6MV beam, both left (negative distances) and right (positive distances), as well as the difference between each expressed as a percentage.
Dose Deficits and Penumbral Flaring for Small Fields in the Lung: The Effect of Lung Density Changes

<table>
<thead>
<tr>
<th>Field Size (cm x cm)</th>
<th>d(80/20)_{film, left (cm)}</th>
<th>d(80/20)_{film, right (cm)}</th>
<th>d(80/20)_{CCC, left (cm)}</th>
<th>d(80/20)_{CCC, right (cm)}</th>
<th>Diff. left (cm)</th>
<th>Diff. right (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>3</td>
<td>0.66</td>
<td>0.64</td>
<td>0.71</td>
<td>0.69</td>
<td>0.03</td>
<td>0.05</td>
</tr>
<tr>
<td>5</td>
<td>0.80</td>
<td>0.75</td>
<td>0.83</td>
<td>0.82</td>
<td>0.03</td>
<td>0.07</td>
</tr>
<tr>
<td>10</td>
<td>0.94</td>
<td>0.9</td>
<td>0.96</td>
<td>0.96</td>
<td>0.02</td>
<td>0.06</td>
</tr>
<tr>
<td>5 (MLC)</td>
<td>0.80</td>
<td>0.81</td>
<td>0.88</td>
<td>0.88</td>
<td>0.03</td>
<td>0.08</td>
</tr>
</tbody>
</table>

Table 4-5 - Comparison of 80%-20% penumbral distances off axis for film and convolution off axis dose profiles. 10MV beam, both left (negative distances) and right (positive distances), as well as the difference between each expressed as a percentage.

From the results above it can be seen that the difference in off axis penumbras for convolution and Gafchromic film is minimal, averaging less than 0.05cm. Comparing both sides of each profile also yielded little difference, with only some variation for the 10cm x 10cm and 5cm x 5cm at 6MV cases. Upon closer inspection these could be contributed to registration errors or misalignment of the film within the phantom (see section 5.1). The highest discrepancies occurred in the case of larger field sizes and higher energy. Additionally, a treatment head setup utilizing MLC defined field did not offer much benefit over the same field size (5cm x 5cm) defined by jaws only. Penumbral difference with laterality may have also been due to the dose calculation grid not being completely symmetrical about the central axis.

4.3 Pinnacle depth dose, effects of changing lung density

Shown in figure 4-25 is the depth dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 6 MV for a 2cm x 2cm
field size. The depth dose was acquired for lung densities 0.2g/cm$^3$, 0.3g/cm$^3$, 0.4g/cm$^3$ and 1g/cm$^3$ and the results compared to one another.

![Convolution percentage depth dose curve showing the effects of changing lung density on LED effect for a 2cm x 2cm field and 6MV beam energy.](image)

**Figure 4-25:** Convolution percentage depth dose curve showing the effects of changing lung density on LED effect for a 2cm x 2cm field and 6MV beam energy.

Figure 4-25 highlights the difference in LED dose reduction caused by varying only the lung density. The drop off from the standard depth dose curve can be observed beginning at 5cm depth when the beam enters the lung. As expected, dose in the lung is least for the lowest value of density, 0.2g/cm$^3$. Closer analysis from Table 4-25 below also highlights that the change in dose in the mid-lung is not linear with respect to the change in density, the only parameter manipulated in this set of data. Though the density (excluding the 1g/cm$^3$ measurement) vary by a consistent 0.1g/cm$^3$, the difference between 0.2g/cm$^3$ and 0.3g/cm$^3$ is 7.34 % PDD whilst the difference between 0.3g/cm$^3$ and 0.4g/cm$^3$ is only 3.6%. The PDD difference between any two densities also appears to stay consistent through the depth of the lung, reaching its lowest point at the lung-tissue interface, before a secondary buildup region.

**Table 4-6:** Comparison of PDD values acquired via convolution in pinnacle lung phantom. A 2cm x 2cm field at 6MV beam energy and at mid lung (10cm depth). PDD difference expressed relative to D0.2g/cm$^3$.

<table>
<thead>
<tr>
<th>Density (g/cm$^3$)</th>
<th>PDD (%)</th>
<th>PDD Diff. (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2</td>
<td>47.88</td>
<td>0</td>
</tr>
<tr>
<td>0.3</td>
<td>55.22</td>
<td>7.34</td>
</tr>
<tr>
<td>0.4</td>
<td>58.84</td>
<td>10.96</td>
</tr>
</tbody>
</table>
Shown in figure 4-26 is the depth dose profile calculated by the collapsed cone convolution (CCC) calculation on pinnacle at 100cm SSD for 10 MV for a 2cm x 2cm field size. The depth dose was acquired for lung densities 0.2g/cm³, 0.3g/cm³, 0.4g/cm³ and 1g/cm³ and the results compared to one another.

![Convolution percentage depth dose curve showing the effects of changing lung density on LED effect for a 2cm x 2cm field and 10MV beam energy.](image)

**Figure 4-26:** Convolution percentage depth dose curve showing the effects of changing lung density on LED effect for a 2cm x 2cm field and 10MV beam energy.

The points addressed above regarding Figure 4-25 are all relevant for Figure 4-26 as well, though obviously as the energy has increased the dose reduction and differences between densities have also scaled up as expected. The amount of dose lost outside of a 2cm x 2cm field in the lung for a 0.2g/cm³ compared to 0.4g/cm³ is a significant 14.26%, which highlights the importance of accurate lung density measurements in treatment planning.

**Table 4-7:** Comparison of PDD values acquired via convolution in pinnacle lung phantom. A 2cm x 2cm field at 10MV beam energy and at mid lung (10cm depth). PDD difference expressed as a percentage of Dmax and relative to D0.2g/cm³.

<table>
<thead>
<tr>
<th>Density (g/cm³)</th>
<th>PDD (%)</th>
<th>PDD Diff. (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2</td>
<td>44.93</td>
<td>0</td>
</tr>
<tr>
<td>0.3</td>
<td>53.54</td>
<td>8.61</td>
</tr>
<tr>
<td>0.4</td>
<td>59.19</td>
<td>14.26</td>
</tr>
<tr>
<td>1</td>
<td>68.13</td>
<td>23.20</td>
</tr>
</tbody>
</table>
Shown in figure 4-26 is the depth dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 6 MV for a 3cm x 3cm field size. The depth dose was acquired for lung densities 0.2g/cm³, 0.3g/cm³, 0.4g/cm³ and 1g/cm³ and the results compared to one another.

![Figure 4-26](image)

**Figure 4-26:** Convolution percentage depth dose curve showing the effects of changing lung density on LED effect for a 3cm x 3cm field and 6MV beam energy.

Figure 4-27 shows a reduction of approximately 2% PDD for 0.2, 0.3 and 0.4g/cm³ when moving from a 2cm x 2cm field to a 3cm x 3cm field. The control measurement also dips below the lower densities in the mid lung.

**Table 4-8:** Comparison of PDD values acquired via convolution in pinnacle lung phantom. A 3cm x 3cm field at 6MV beam energy and at mid lung (10cm depth). PDD difference expressed as a percentage of Dmax and relative to D(0.2g/cm³).

<table>
<thead>
<tr>
<th>Density (g/cm³)</th>
<th>PDD (%)</th>
<th>PDD Diff. (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2</td>
<td>56.07</td>
<td>0</td>
</tr>
<tr>
<td>0.3</td>
<td>61.51</td>
<td>5.44</td>
</tr>
<tr>
<td>0.4</td>
<td>63.57</td>
<td>7.50</td>
</tr>
<tr>
<td>1</td>
<td>60.18</td>
<td>4.11</td>
</tr>
</tbody>
</table>

Shown in figure 4-27 is the depth dose profile calculated by the collapsed cone convolution (CCC) calculation on pinnacle at 100cm SSD for 10 MV for a 3cm x 3cm
field size. The depth dose was acquired for lung densities 0.2g/cm$^3$, 0.3g/cm$^3$, 0.4g/cm$^3$ and 1g/cm$^3$ and the results compared to one another.

![Image of depth dose profile](image)

**Figure 4-28:** Convolution percentage depth dose curve showing the effects of changing lung density on LED effect for a 3cm x 3cm field and 10MV beam energy.

**Table 4-9:** Comparison of PDD values acquired via convolution in pinnacle lung phantom. A 3cm x 3cm field at 10MV beam energy and at mid lung (10cm depth). PDD difference expressed as a percentage of Dmax and relative to D(0.2g/cm$^3$).

<table>
<thead>
<tr>
<th>Density (g/cm$^3$)</th>
<th>PDD (%)</th>
<th>PDD Diff. (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2</td>
<td>54.45</td>
<td>0</td>
</tr>
<tr>
<td>0.3</td>
<td>61.86</td>
<td>7.41</td>
</tr>
<tr>
<td>0.4</td>
<td>66.21</td>
<td>11.76</td>
</tr>
<tr>
<td>1</td>
<td>69.56</td>
<td>15.11</td>
</tr>
</tbody>
</table>

Shown in figure 4-28 is the depth dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 6 MV for a 5cm x 5cm field size. The depth dose was acquired for lung densities 0.2g/cm$^3$, 0.3g/cm$^3$, 0.4g/cm$^3$ and 1g/cm$^3$ and the results compared to one another.
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Figure 4-29: Convolution percentage depth dose curve showing the effects of changing lung density on LED effect for a 5cm x 5cm field and 6MV beam energy.

Moving to a 5cm x 5cm field the amount of dose reduction due to LED is observed to be much smaller. The discrepancy between 0.3g/cm$^3$ and 0.4g/cm$^3$ is not visually noticeable in Figure 4.3(e) and analysing the numerical data there is only 0.44% PDD between the two curves.

Table 4-10: Comparison of PDD values acquired via convolution in pinnacle lung phantom. A 5cm x 5cm field at 6MV beam energy and at mid lung (10cm depth). PDD difference expressed as a percentage of Dmax and relative to D(0.2g/cm$^3$).

<table>
<thead>
<tr>
<th>Density (g/cm$^3$)</th>
<th>PDD (%)</th>
<th>PDD Diff. (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2</td>
<td>64.55</td>
<td>0</td>
</tr>
<tr>
<td>0.3</td>
<td>67.10</td>
<td>2.55</td>
</tr>
<tr>
<td>0.4</td>
<td>67.44</td>
<td>2.89</td>
</tr>
<tr>
<td>1</td>
<td>62.61</td>
<td>-1.94</td>
</tr>
</tbody>
</table>

Shown in figure 4-29 is the depth dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 10 MV for a 5cm x 5cm field size. The depth dose was acquired for lung densities 0.2g/cm$^3$, 0.3g/cm$^3$, 0.4g/cm$^3$ and 1g/cm$^3$ and the results compared to one another.
Figure 4-30: Convolution percentage depth dose curve showing the effects of changing lung density on LED effect for a 5cm x 5cm field and 10MV beam energy.

Figure 4-30 further highlights the effect of changing field size.

Table 4-11: Comparison of PDD values acquired via convolution in pinnacle lung phantom. A 5cm x 5cm field at 10MV beam energy and at mid lung (10cm depth). PDD difference expressed as a percentage of Dmax and relative to D(0.2g/cm³).

<table>
<thead>
<tr>
<th>Density (g/cm³)</th>
<th>PDD (%)</th>
<th>PDD Diff. (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2</td>
<td>66.82</td>
<td>0</td>
</tr>
<tr>
<td>0.3</td>
<td>71.40</td>
<td>4.58</td>
</tr>
<tr>
<td>0.4</td>
<td>73.32</td>
<td>6.50</td>
</tr>
<tr>
<td>1</td>
<td>71.28</td>
<td>4.46</td>
</tr>
</tbody>
</table>

Shown in figure 4-31 is the depth dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 6 MV for a 10cm x 10cm field size. The depth dose was acquired for lung densities 0.2g/cm³, 0.3g/cm³, 0.4g/cm³ and 1g/cm³ and the results compared to one another.
Figure 4-31: Convolution percentage depth dose curve showing the effects of changing lung density on LED effect for a 10cm x 10cm field and 6MV beam energy.

In Figure 4-31 and Figure 4-32 it can be observed that for fields larger than 5cm x 5cm the LED effect cannot occur to a significant extent and in fact the dose here is higher than just a standard depth dose curve in regular tissue of density 1g/cm$^3$. The PDD difference between densities in the lung is less than 1%.

Table 4-12: Comparison of PDD values acquired via convolution in pinnacle lung phantom. A 10cm x 10cm field at 6MV beam energy and at mid lung (10cm depth). PDD difference expressed as a percentage of Dmax and relative to D(0.2g/cm$^3$).

<table>
<thead>
<tr>
<th>Density (g/cm$^3$)</th>
<th>PDD (%)</th>
<th>PDD Diff. (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2</td>
<td>70.19</td>
<td>0</td>
</tr>
<tr>
<td>0.3</td>
<td>70.82</td>
<td>0.63</td>
</tr>
<tr>
<td>0.4</td>
<td>70.50</td>
<td>0.31</td>
</tr>
<tr>
<td>1</td>
<td>66.35</td>
<td>-3.84</td>
</tr>
</tbody>
</table>

Shown in figure 4-32 is the depth dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 10 MV for a 10cm x 10cm field size. The depth dose was acquired for lung densities 0.2g/cm$^3$, 0.3g/cm$^3$, 0.4g/cm$^3$ and 1g/cm$^3$ and the results compared to one another.
Dose Deficits and Penumbral Flaring for Small Fields in the Lung: The Effect of Lung Density Changes

**Figure 4-32:** Convolution percentage depth dose curve showing the effects of changing lung density on LED effect for a 10cm x 10cm field and 10MV beam energy.

**Table 4-13:** Comparison of PDD values acquired via convolution in pinnacle lung phantom. A 10cm x 10cm field at 10MV beam energy and at mid lung (10cm depth). PDD difference expressed as a percentage of Dmax and relative to D0.2g/cm3.

<table>
<thead>
<tr>
<th>Density (g/cm³)</th>
<th>PDD (%)</th>
<th>PDD Diff. (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2</td>
<td>76.23</td>
<td>0</td>
</tr>
<tr>
<td>0.3</td>
<td>77.36</td>
<td>1.13</td>
</tr>
<tr>
<td>0.4</td>
<td>77.15</td>
<td>0.92</td>
</tr>
<tr>
<td>1</td>
<td>73.41</td>
<td>-2.82</td>
</tr>
</tbody>
</table>

4.4 Pinnacle off-axis dose, effects of changing density

Shown in figure 4-33 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 6 MV for a 3cm x 3cm field size calculated with a 1mm x 1mm x 1mm dose voxel grid. A script was written to extract the off-axis dose data. The depth dose was acquired for lung densities 0.2g/cm³, 0.3g/cm³, 0.4g/cm³ and 1g/cm³ and the results compared to one another.
Figure 4-33: Convolution off-axis dose profile showing the effects of changing lung density on penumbral flaring for a 3cm x 3cm field and 6MV beam energy in mid lung (10cm depth).

Figure 4-33 quantifies the LED effect in a different way, as it can be observed that as the density of the lung decreases there is an increase in the penumbral flaring. In a physical sense, the range of secondary electrons in less dense lungs are greater and so more dose is being deposited near the edge or outside of the field. This is represented graphically above.

Table 4-14: Off-axis distance between 80% and 20% of max dose for varying lung densities. A 3cm x 3cm field at 6MV beam energy and in the mid lung (10cm depth).

<table>
<thead>
<tr>
<th>Density (g/cm³)</th>
<th>OAD (80%) (cm)</th>
<th>OAD (20%) (cm)</th>
<th>d_{80-20} (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2</td>
<td>1.41</td>
<td>2.26</td>
<td>0.85</td>
</tr>
<tr>
<td>0.3</td>
<td>1.43</td>
<td>2.13</td>
<td>0.70</td>
</tr>
<tr>
<td>0.4</td>
<td>1.46</td>
<td>2.06</td>
<td>0.6</td>
</tr>
<tr>
<td>1</td>
<td>1.53</td>
<td>1.95</td>
<td>0.42</td>
</tr>
</tbody>
</table>

Shown in figure 4-34 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 10 MV for a 3cm x 3cm field size in the mid lung (10cm depth). The depth dose was acquired for lung densities 0.2g/cm³, 0.3g/cm³, 0.4g/cm³ and 1g/cm³ and the results compared to one another.
Figure 4-34: Convolution off-axis dose profile showing the effects of changing lung density on penumbral flaring for a 3cm x 3cm field and 10MV beam energy in mid lung (10cm depth).

Moving to a 10MV beam energy in figure 4-34, it is observed that at higher energies there is a general spreading out of the off-axis profile, as more flaring is observed in the penumbra for all four densities. The relationship between the densities remains the same though, with 0.2g/cm³ showing the highest amount of flaring and therefore LED effect, decreasing almost linearly (as seen in Table 4.4(b) below) with 0.3g/cm³ and 0.4g/cm³ respectively.

Table 4-15: Off-axis distance between 80% and 20% of max dose for varying lung densities. A 3cm x 3cm field at 10MV beam energy and in the mid lung (10cm depth).

<table>
<thead>
<tr>
<th>Density (g/cm³)</th>
<th>OAD (80%)</th>
<th>OAD (20%)</th>
<th>d_{80-20} (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2</td>
<td>1.42</td>
<td>2.54</td>
<td>1.12</td>
</tr>
<tr>
<td>0.3</td>
<td>1.42</td>
<td>2.32</td>
<td>0.90</td>
</tr>
<tr>
<td>0.4</td>
<td>1.42</td>
<td>2.22</td>
<td>0.80</td>
</tr>
<tr>
<td>1</td>
<td>1.48</td>
<td>2.02</td>
<td>0.54</td>
</tr>
</tbody>
</table>

Shown in figure 4-35 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 6 MV for a 3cm x 3cm field size at the lung-tissue interface (15cm depth). The depth dose was acquired for
lung densities 0.2g/cm\(^3\), 0.3g/cm\(^3\), 0.4g/cm\(^3\) and 1g/cm\(^3\) and the results compared to one another.

![Graph showing off-axis dose profile](image)

**Figure 4-35:** Convolution off-axis dose profile showing the effects of changing lung density on penumbral flaring for a 3cm x 3cm field and 6MV beam energy at lung-tissue interface (15cm depth).

**Table 4-16:** Off-axis distance between 80% and 20% of max dose for varying lung densities. A 3cm x 3cm field at 6MV beam energy and at lung-tissue interface (15cm depth).

<table>
<thead>
<tr>
<th>Density (g/cm(^3))</th>
<th>OAD (80%)</th>
<th>OAD (20%)</th>
<th>(d_{80-20}) (cm)</th>
</tr>
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<tr>
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<td>2.10</td>
<td>0.85</td>
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<tr>
<td>0.3</td>
<td>1.27</td>
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<td>0.71</td>
</tr>
<tr>
<td>0.4</td>
<td>1.29</td>
<td>1.92</td>
<td>0.63</td>
</tr>
<tr>
<td>1</td>
<td>1.37</td>
<td>1.81</td>
<td>0.44</td>
</tr>
</tbody>
</table>

Shown in figure 4-36 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 10 MV for a 3cm x 3cm field size at the lung-tissue interface (15cm depth). The depth dose was acquired for lung densities 0.2g/cm\(^3\), 0.3g/cm\(^3\), 0.4g/cm\(^3\) and 1g/cm\(^3\) and the results compared to one another.
Dose Deficits and Penumbral Flaring for Small Fields in the Lung: The Effect of Lung Density Changes

**Figure 4-36:** Convolution off-axis dose profile showing the effects of changing lung density on penumbral flaring for a 3cm x 3cm field and 10MV beam energy at lung-tissue interface (15cm depth).

**Table 4-17:** Off-axis distance between 80% and 20% of max dose for varying lung densities. A 3cm x 3cm field at 10MV beam energy and at lung-tissue interface (15cm depth).

<table>
<thead>
<tr>
<th>Density (g/cm³)</th>
<th>OAD (80%)</th>
<th>OAD (20%)</th>
<th>d₈₀₋₂₀ (cm)</th>
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<tr>
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<td>1.25</td>
<td>2.37</td>
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<td>0.3</td>
<td>1.25</td>
<td>2.17</td>
<td>0.92</td>
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<td>0.4</td>
<td>1.25</td>
<td>2.06</td>
<td>0.81</td>
</tr>
<tr>
<td>1</td>
<td>1.33</td>
<td>1.85</td>
<td>0.52</td>
</tr>
</tbody>
</table>

Shown in figure 4-37 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 6 MV for a 5cm x 5cm field size in the mid lung (10cm depth). The depth dose was acquired for lung densities 0.2g/cm³, 0.3g/cm³, 0.4g/cm³ and 1g/cm³ and the results compared to one another.
Chapter 4: Results

Figure 4-37: Convolution off-axis dose profile showing the effects of changing lung density on penumbral flaring for a 5cm x 5cm field and 6MV beam energy in mid lung (10cm depth).

Table 4-18: Off-axis distance between 80% and 20% of max dose for varying lung densities. A 5cm x 5cm field at 6MV beam energy and in the mid lung (10cm depth).

<table>
<thead>
<tr>
<th>Density (g/cm³)</th>
<th>OAD (80%)</th>
<th>OAD (20%)</th>
<th>d_{80-20} (cm)</th>
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<tr>
<td>0.2</td>
<td>2.27</td>
<td>3.32</td>
<td>1.05</td>
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<td>0.3</td>
<td>2.36</td>
<td>3.18</td>
<td>0.82</td>
</tr>
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<td>0.4</td>
<td>2.41</td>
<td>3.09</td>
<td>0.68</td>
</tr>
<tr>
<td>1</td>
<td>2.51</td>
<td>2.97</td>
<td>0.46</td>
</tr>
</tbody>
</table>

Shown in figure 4-38 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 10 MV for a 5cm x 5cm field size in the mid lung (10cm depth). The depth dose was acquired for lung densities 0.2g/cm³, 0.3g/cm³, 0.4g/cm³ and 1g/cm³ and the results compared to one another.
Figure 4-38: Convolution off-axis dose profile showing the effects of changing lung density on penumbral flaring for a 3cm x 3cm field and 10MV beam energy in mid lung (10cm depth).

Table 4-19: Off-axis distance between 80% and 20% of max dose for varying lung densities. A 5cm x 5cm field at 10MV beam energy and in the mid lung (10cm depth).

<table>
<thead>
<tr>
<th>Density (g/cm³)</th>
<th>OAD (80%)</th>
<th>OAD (20%)</th>
<th>d_{80-20} (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2</td>
<td>2.18</td>
<td>3.54</td>
<td>1.36</td>
</tr>
<tr>
<td>0.3</td>
<td>2.24</td>
<td>3.37</td>
<td>1.13</td>
</tr>
<tr>
<td>0.4</td>
<td>2.30</td>
<td>3.26</td>
<td>0.96</td>
</tr>
<tr>
<td>1</td>
<td>2.45</td>
<td>3.05</td>
<td>0.60</td>
</tr>
</tbody>
</table>

Shown in figure 4-39 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 6 MV for a 5cm x 5cm field size at the lung-tissue interface (15cm depth). The depth dose was acquired for lung densities 0.2g/cm³, 0.3g/cm³, 0.4g/cm³ and 1g/cm³ and the results compared to one another.
Figure 4-39: Convolution off-axis dose profile showing the effects of changing lung density on penumbral flaring for a 5cm x 5cm field and 6MV beam energy at lung-tissue interface (15cm depth).

Table 4-20: Off-axis distance between 80% and 20% of max dose for varying lung densities. A 5cm x 5cm field at 6MV beam energy and at lung-tissue interface (15cm depth).

<table>
<thead>
<tr>
<th>Density (g/cm³)</th>
<th>OAD (80%)</th>
<th>OAD (20%)</th>
<th>d_{80-20} (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2</td>
<td>2.66</td>
<td>3.69</td>
<td>1.03</td>
</tr>
<tr>
<td>0.3</td>
<td>2.72</td>
<td>3.60</td>
<td>0.88</td>
</tr>
<tr>
<td>0.4</td>
<td>2.78</td>
<td>3.47</td>
<td>0.69</td>
</tr>
<tr>
<td>1</td>
<td>2.87</td>
<td>3.36</td>
<td>0.49</td>
</tr>
</tbody>
</table>

Shown in figure 4-40 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 10 MV for a 5cm x 5cm field size at the lung-tissue interface (15cm depth). The depth dose was acquired for lung densities 0.2g/cm³, 0.3g/cm³, 0.4g/cm³ and 1g/cm³ and the results compared to one another.
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Figure 4-40: Convolution off-axis dose profile showing the effects of changing lung density on penumbral flaring for a 5cm x 5cm field and 10MV beam energy at lung-tissue interface (15cm depth).

Table 4-21: Off-axis distance between 80% and 20% of max dose for varying lung densities. A 5cm x 5cm field at 10MV beam energy and at lung-tissue interface (15cm depth).

<table>
<thead>
<tr>
<th>Density (g/cm³)</th>
<th>OAD (80%)</th>
<th>OAD (20%)</th>
<th>d₈₀₋₂₀ (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2</td>
<td>2.59</td>
<td>3.98</td>
<td>1.39</td>
</tr>
<tr>
<td>0.3</td>
<td>2.61</td>
<td>3.75</td>
<td>1.14</td>
</tr>
<tr>
<td>0.4</td>
<td>2.65</td>
<td>3.63</td>
<td>0.98</td>
</tr>
<tr>
<td>1</td>
<td>2.81</td>
<td>3.44</td>
<td>0.63</td>
</tr>
</tbody>
</table>

Shown in figure 4-41 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 6 MV for a 10cm x 10cm field size in the mid lung (10cm depth). The depth dose was acquired for lung densities 0.2g/cm³, 0.3g/cm³, 0.4g/cm³ and 1g/cm³ and the results compared to one another.
Figure 4-41: Convolution off-axis dose profile showing the effects of changing lung density on penumbral flaring for a 10cm x 10cm field and 6MV beam energy in the mid lung (10cm depth).

Shown in figure 4-42 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 10 MV for a 10cm x 10cm field size in the mid lung (10cm depth). The depth dose was acquired for lung densities 0.2g/cm$^3$, 0.3g/cm$^3$, 0.4g/cm$^3$ and 1g/cm$^3$ and the results compared to one another.

Figure 4-42: Convolution off-axis dose profile showing the effects of changing lung density on penumbral flaring for a 10cm x 10cm field and 10MV beam energy in the mid lung (10cm depth).

Shown in figure 4-43 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 6 MV for a 10cm x 10cm field size at the lung-tissue interface (15cm depth). The depth dose was acquired for lung densities 0.2g/cm$^3$, 0.3g/cm$^3$, 0.4g/cm$^3$ and 1g/cm$^3$ and the results compared to one another.
Figure 4-43: Convolution off-axis dose profile showing the effects of changing lung density on penumbral flaring for a 10cm x 10cm field and 6MV beam energy at the lung-tissue interface (15cm depth).

Shown in figure 4-44 is the off-axis dose profile calculated by the collapsed cone convolution (CCC) calculation on Pinnacle at 100cm SSD for 10 MV for a 10cm x 10cm field size at the lung-tissue interface (15cm depth). The depth dose was acquired for lung densities 0.2g/cm$^3$, 0.3g/cm$^3$, 0.4g/cm$^3$ and 1g/cm$^3$ and the results compared to one another.

Figure 4-44: Convolution off-axis dose profile showing the effects of changing lung density on penumbral flaring for a 10cm x 10cm field and 10MV beam energy at the lung-tissue interface (15cm depth).
5 DISCUSSION

5.1 Phantom based comparison of Monte Carlo simulation, Pinnacle convolution and ion chamber percentage depth doses

The figures in section 4.1 largely confirmed the assumptions held regarding the effects of LED in the lung from previous research [19]. Regardless of the combination of field size, energy and density, the dose reduction effect associated with LED appeared to begin at the tissue-lung interface at 5cm depth and continued until the lung-tissue interface at 15cm depth. At this point, a secondary buildup region was observed and the dose peaked at a point approximately 1cm past the lung-tissue interface, before decreasing in line with standard depth dose curve not affected by LED.

The steepness of the dose gradient at 5cm depth and the overall loss of dose along the central axis was largely observed to be due to the combination of beam energy and field size. As seen in figure 4-2, the dose reduction effect was largest for the optimal combination of small field and high energy, namely for the 10MV and 2cm x 2cm setup. In general, both increasing energy and decreasing field size were seen to increase the effect of LED on dose reduction. The accuracy of the electrometer measurements using field sizes smaller than 3cm x 3cm is questionable, due to the physical size of the chamber, however the trend remains clear.

The Pinnacle convolution measurements showed good agreement with the ion chamber measurements in most cases. Assuming that the ion chamber results were obtained
Dose Deficits and Penumbral Flaring for Small Fields in the Lung: The Effect of Lung Density Changes

accurately and are a good physical model of what is actually happening inside the lung, the CCC can be said to have modeled the LED effect accurately down to a 5cm x 5cm field with a 6MV beam. Anything below this in terms of field size or above in terms of beam energy produced at least some level of discrepancy between CCC and ion chamber.

The largest deviation between the two methods occurred for the small field, high energy case of a 10MV beam and 2cm x 2cm field. Here an over-response of the ion chamber can be seen in the mid-lung, of approximately 4% PDD. More curious is the behaviour of the ion chamber, which overresponded even more compared to Monte Carlo for small fields, in some cases as much as 7.9% from the Monte Carlo value in the steep dose gradient region of the lung. This is quite a significant discrepancy in terms of practical clinical applications, and is not due to the accuracy of the tool but rather is a systematic error resulting from some unknown physical process. One potential explanation for this trend is that ion chamber may be in disequilibrium for small fields hence the over response, however, more experimentation is required to quantify this.

The overestimation by the ion chamber may also be due to some equilibrium being re-established by the chamber wall. The effect seen in this thesis is affirmed by Carrasco et al, 2004 [27], which show the same underestimation of the central axis dose presented in this work. The clinical consequence of this result is that the overestimation of the LED effect could lead to an underestimation of the dose required by calculation algorithm and hence reduced tumour control probability. Publications by Maucerie [28], Rice [29], and Araki [30], predict this over-response but the exact quantification was not established in this thesis.

The Monte Carlo percentage depth dose curves shown in section 4.1 show a clear deficit of between 2-7% in the amount of dose in the region of the lung affected by LED, when compared with both convolution and ion chamber methods of dose calculation. This level of accuracy of the convolution in comparison to the designated ‘true value’ of dose distribution assigned to the Monte Carlo is within the limits of that which is expected, as discussed in section 5.4. The Monte Carlo was always lower than the convolution, with the discrepancy increasing for higher energies and thinner fields in line with the increase in the amount of disequilibrium occurring. This trend is expected as it agrees
with previous Monte Carlo simulations by Keall and Hoban [31], who also found a disagreement in the lung. More specifically, they attribute this to the rectilinear scaling of the dose spread arrays in convolution models slightly overestimating the dose compared with real scaling of curved electron paths, which is accounted for in most Monte Carlo methods.

A study of perturbation factors and inhomogeneity corrections by Araki in 2012 [30] used an inhomogeneous lung phantom (density 0.3g/cm³) to compare PDDs between Monte Carlo and a PTW 31010 ion chamber for field sizes of 10cm x 10cm, 5cm x 5cm and 3cm x 3cm at 6MV and 15MV energies. The study observed an overestimation of the ion chamber result for lung depth dose when LED was present, however the magnitude of overestimation is only about 2%. This is in part due to the perturbation of the electron fluence through the thin chamber wall, where the roughly water equivalent density does not match the lower density of the lung within the phantom. This creates a build-up of dose on the chamber wall, an effect also described by Disher et al in 2013 [19]. This in part explains why the ion chamber measurements in lung were higher than convolution and Monte Carlo.

Only a Monte Carlo simulation including the CC04 geometry would quantify the difference but due to time constraints this was beyond the scope of this thesis. This is going to be attempted as future work. Another potential reason for the differences is that all data was normalised at dmax. It is possible volume averaging of the ionization chamber at dmax in the sharper profile may have induced an over correction of the normalised ionization data in the flatter lung profiles, a hypothesis which also falls into the category of future investigation.

In retrospect, the detectors used should have been characterised if they were to be used to compare measurements against calculations as part of planning system validation. This could have allowed the overresponse of the CC04 IC to be predicted before small field measurements in non-water density media rather than be a source of confusion. For further extensions of this work, some characterisation measurements conducted in water or water like phantom media would better assist with analysis of results.
Figure 5-1: PDD curves for irradiation of a CIRS lung phantom compared to several commercial calculation algorithms shows the same underestimation of central axis dose by the CCC as presented in this work. The left column shows 6MV photons and the right 18MV, whilst the rows show results for 10cm x 10cm, 5cm x 5cm and 2cm x 2cm from top to bottom. The empty circles represent TLDs, the triangles ionization chamber and the black line Monte Carlo. [27].

5.2 Off axis dose measurements, comparison of convolution and EBT film
The Gafchromic film measurements showed good agreement with Pinnacle CCC for all cases. Penumbral flaring was observed to some extent for all combinations of parameters, but more so for higher energies and smaller field sizes, as expected. Here
the range of secondary electrons is greater and more dose is deposited at the edge of the field. There also seemed to be comparatively less agreement surrounding the central axis (dose plateau) for smaller field sizes. This effect seems to be caused by an increased deviation in the position of the maximum dose when off axis profiles were collected using ImageJ software. This results in a dose plateau region that appears noisier when compared with the smooth curve of the convolution. Smoothing of the data may have alleviated this, but at the cost of spatial resolution. However, an error of this nature is expected when considering the small sensitive volume of the CC04 thimble chamber (0.04cm$^3$) combined with the small field size used in these measurements.

![Image of EBT Gafchromic film](image.png)

**Figure 5-2:** Final post registration image of EBT Gafchromic film. Note the off centre exposure.

Further, some small discrepancies between real and simulated data can be seen in the beginning and end of the penumbra. This may be due to difficulties in centering the film within the treatment field, resulting in an exposure such as the one shown in figure 5-2. The resulting dose profile was therefore off centre and had to be adjusted horizontally in some cases to give an accurate comparison with Pinnacle convolution. More care taken to create a Pinnacle dose grid centred around the virtual phantom could have alleviated this lateral error.

### 5.3 Pinnacle depth dose, effects of changing lung density

The results shown in the figures in section 4.3 affirmed the expected trend relating to LED in the lung. The dose reduction effect increased as lung density decreased. As
above, dose reduction was most pronounced for the case of 10MV beam energy and 2cm x 2cm field size, as the LED effect becomes more pronounced for higher energy and smaller field as observed in figure 5-1.

The PDD difference between 0.4g/cm$^3$ and 0.3g/cm$^3$ for this setup was seen to be approximately 4% of max dose, whereas decreasing lung density from 0.3g/cm$^3$ to the minimum value of 0.2g/cm$^3$ produced a difference of approximately 6%. This suggests a non-linear depreciation of dose as lung density is reduced. This trend is also seen for other beam and field setups, albeit on a smaller scale. The boundaries of the LED effect were seen to be consistent across all setups, with the dose drop-off beginning at 5cm depth and the secondary build up occurring at 15cm depth unanimously.

5.4 Pinnacle off-axis dose, effects of changing density
The effects of changing lung density of the magnitude of penumbral flaring was observed in section 4.4. The magnitude of flaring was seen to increase the lower the density. This was quantified using the 80/20 ratio, the off-axis distance along the profile between the points of 80% and 20% of the profile’s maximum dose. These results are seen in tables 4-5 and 4-6. As in previous sections, the magnitude of this effect is seen to increase with higher energies and smaller field sizes. This was to be expected, as for higher energies more secondary electrons are being generated, which are in turn more likely to deposit dose near the edge or outside of what is a relatively thin field. The range of the secondary electrons in-field for lower densities have a greater range, and are as such more likely to deposit dose towards the edge of the field. This more spread out dose distribution causes the penumbras to be wider in comparison to the steep decline observed for higher densities.

5.5 Acceptable Error Limits
Any validation of radiotherapy treatment planning dose calculation methods must include a discussion about the accuracy of these methods when compared with some gold standard. In this thesis, Monte Carlo based GEANT4 dose calculations should provide a high level of accuracy in estimating the dose delivered to the lung under LED conditions, as they simulate the interaction of photons and matter.
Previously, studies such as Fragoso et al (2010) [32] have conducted validation of a MC-based dose calculation method, comparing it with film and ionisation chambers in both water phantoms and heterogenous solid water slabs containing lung equivalent material. They found that the agreement between the calculated and measured dose distributions was 2% in water and 4% in the solid water phantom. However, this range increased slightly when measured for small lesions in the lung where LED is present, such as in this thesis.

This range of tolerance for differences between measured and calculated dose is also affirmed by Van Dyk et al (2013) [33], who propose 2 – 3% of dose delivered as an ideal level of accuracy for optimising patient treatment. However, the authors acknowledge that in reality, with the increasing complexity of modern dose calculation algorithms and treatments that this figure is extremely difficult to adhere to and that any discussion of dose distribution accuracy must take into account what is reasonably achievable. Thwaites (2013) [34] expands on this by explaining that clinical dose accuracy requirements are dependent on dose response curves for TCP and NTCP. These accuracy recommendations need to be based on the steepest dose-effect relationships, where a change of 5% in dose can cause an unacceptable shift in TCP of 10-20% and 20-30% change in NTCP. To this end, the author refers to a general figure of 3% (relative sd) agreed upon by several supporting studies.

The results of this thesis show an accuracy, (that is to say the comparison of both ion chamber and convolution to the true value provided by the Monte Carlo simulations) which for the most part lie within the range of accuracy discussed above. As expected the worst case situation was the setup which included the highest energy and thinnest field, namely the 2cm x 2cm field at 10MV energy. Here the difference between the convolution and Monte Carlo dose peaked at approximately 5% in the steep dose gradient region immediately following the tissue-lung interface. Upon stabilising in the mid lung LED region, the difference between the two curves decreased to approximately 2%.
6 CONCLUSIONS

The goals of this thesis were to:

1) Compare convolution calculations of dose by cylindrical and parallel plate ionisation chamber measurements in a lung phantom. These results are presented in section 4.1.

2) Compare convolution calculations of dose off axis by Gafchromic film measurements in a lung phantom. This comparison is presented in section 4.2.

3) Characterise depth dose and off axis dose in lung at different lung densities. See section 4.3.

4) Explain differences between ionisation chamber and convolution calculations for small field sizes by Monte Carlo simulation. See discussion in section 5.1.

In broader terms, the aim of this work was to determine the effect of lateral electron disequilibrium in the lung for small treatment fields (down to 2cm x 2cm) as sometimes used in stereotactic body radiation therapy. The LED effect was quantified in terms of the reduction in percentage depth dose observed along the central axis of the lung as well as the increased penumbral flaring observed in off-axis dose profiles. Pinnacle convolution was performed on a simulated lung phantom and benchmarked against
EBT3 Gafchromic film, a CC04 thimble chamber, an Advanced Marcus ionisation chamber and Monte Carlo simulations to assess the relative accuracy of each method.

It was observed through this experimentation that all of the parameters manipulate in the experiment affect LED to some degree, and an optimal combination of them is required to see the greatest magnitude of dose reduction and penumbral flaring occur. LED was seen to increase for:

- Increasing beam energy (10MV compared to 6MV)
- Decreasing field size (2cm x 2cm produced the greatest effect, but anything less than 10cm x 10cm was observed to produce LED)
- Decreasing density (Lung densities 0.2g/cm$^3$ – 0.4g/cm$^3$ all produced LED but it was observed that the lower the density the greater the dose deficit effect).

Therefore, the results of this experiment largely reaffirm the conclusions of previous research into the topic down to fields 5cm x 5cm. However, this research extends the comparison to smaller (2cm x 2cm) fields and up to a 7.9% difference between convolution and ion chamber is observed for a 10MV, 2cm x 2cm field at 0.3g/cm$^3$.

LED can be seen to cause a significant deviation from the prescribed dose when considering the high energies and small treatment fields associated with stereotactic body radiotherapy. Whilst this could certainly have negative repercussions for the efficacy of patient treatment if the lung density is incorrectly estimated using CBCT for instance, there is also some potential for clinicians to manipulate the high dose gradients generated in the lung to produce highly localized treatment.

This study only summarises single field effects and multiple fields may ameliorate the error. This study does however provide solid benchmark data to help guide clinical medical physicists with the advice they provide about the accuracy of this algorithm in lung when used for small field SABR treatment planning.
7 REFERENCES


[34] D Thwaites, “Accuracy required and achievable in radiotherapy dosimetry: have modern technology and techniques changed our views?”, 2013 Journal of Physics: Conference Series 44401200

