Field matching of 6 MeV X-Ray asymmetric collimated fields for radiotherapy patients

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FIELD MATCHING OF 6 MeV X-RAY ASYMMETRIC
COLLIMATED FIELDS FOR RADIOTHERAPY PATIENTS

A Thesis submitted for the partial fulfilment of the
requirements for the award of the degree of

Master of Science

from

The University of Wollongong

by

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Department of Physics 1993
This thesis is submitted to
THE UNIVERSITY OF WOLLONGONG
and has not been submitted for a higher
degree to any other University or institution

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Abstract

Asymmetric X-ray fields are used in cancer therapy to eliminate the problem of beam divergence with depth. A study of the dose distribution resulting from asymmetric fields being matched with both symmetric and asymmetric fields is presented.

Forming functions are used by computer planning systems to model the way X-ray beams interact with the body of the patient. The properties of two forming functions are compared with a possible alternative function.
Acknowledgements

I would like to thank Dr Peter Metcalfe for the opportunity to use the linear accelerator at the Illawarra Cancer Care Centre for my thesis work and for his help and encouragement during the course of my work.

I would like to thank Dr Tomas Kron who was always full of ideas and who was a great encouragement with the forming function section of this thesis.

Thanks to Dr Jagdish Mathur for all his help and advice, not only during the work on this thesis but through out my last four years at the University of Wollongong

Thankyou also to Mr Tony Wong and all the staff at the ICCC who put up with me for a year.

I would like to thank my parents for all that they have done to encourage me over the years and I would like to thank the Lord for whom all things are possible.
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Chapter 1

Introduction
1.1 Introduction

For the treatment of cancer one of the most commonly used methods is the use of high energy photons (X-rays) which are capable of penetrating to great depths within the body.

In order to cause maximum damage to cancer tissue while doing the least damage to the surrounding healthy tissue the X-ray beam alignment and positioning is critical.

To achieve this, beam alignment techniques such as angling of the treatment couch and collimators and also the use of beam blocks have been developed.

This thesis describes the results of using another technique involving the use of the linear accelerators own collimators. Collimators which are built into the Linear Accelerator and which act to shape the X-ray beam as it emerges from the accelerator.

The collimators on modern accelerators can be driven independently and are referred to as independent jaws.

The technique developed in this research work, which is discussed in detail in this thesis involves using the asymmetric collimators to block off one half of the X-ray beam by driving one of them to the centre of the beam which is indicated by a cross hair maker on the transparent mylar exit plate of the accelerator.

As explained in chapter one, this blocking of half the beam gives, a straight edge to what is normally a diverging beam, which is helpful in matching of the beams for multibeam treatments.

The way in which this works is that the nondivergent edge of one of the asymmetric beams is matched to the non divergent edge of a second beam coming from a different angle. The nondiverging edge of these two beams can then be matched with that of a third asymmetric beam and the dose at the junction of these two beams should be uniform because of their nondivergence.

In this thesis the initial measurements for asymmetric profiles were done using a Scanditronix RFA300 water tank and a red (photon measuring) diode.
A second set of measurements were then done using a Rando phantom which had a 2mm thick slice of solid water cut to shape and placed between the sections of the phantom. This solid water slice had small TLD recesses cut into its surfaces. The positioning of these recesses was made to coincide with isodose positions derived from a CT plan of the Rando phantom.

The second part of the thesis is concerned with forming functions. These are mathematical functions which model the shape of the dose profile and are of importance in computer planning systems.

Two such forming functions are presented in this thesis. They consist of an inverse square root function to reproduce the S-shaped curve of the penumbral region.

The first function is continuous and attempts to model an asymmetric dose profile.

The second function is an adjusted version of the first and attempts to model an asymmetric dose profile. Both functions were compared with measured data and also with the Cosine forming function\textsuperscript{18} and the exponential Function\textsuperscript{19}. All three functions are tested against measured data taken from the water tank for both single fields as well as for matching of both symmetric/symmetric and asymmetric/asymmetric fields.
1.2 Breast Cancer

Breast cancer is the most common type of cancer suffered by women in western countries. About seven percent of women will develop breast cancer and it accounts for between 20% and 25% of cancer related deaths in women. Of all the women who develop breast cancer the number for whom it proves fatal has remained constant for over thirty years. In New South Wales it is estimated that about one in fifteen women will develop breast cancer before the age of 75. The statistics seem to indicate that a woman is more at risk if a) she is over 40, b) her mother or sister had breast cancer and c) if she has already had cancer in either of her breasts. She is also at risk if she has never had children or if her first child was born after the age of thirty. Men can also develop breast cancer but when compared to women the chances of this occurring are very slight. What actually triggers a breast cancer is still a mystery, there may well be several factors involved. These could include such things as hormonal factors and age.

1.2.1 Hormonal Factors

Unlike the other tissues of a woman's body (with the exception of the uterus) the tissue of the breast is undergoing monthly changes with a woman's hormone cycle. The result of this hormone stimulation is the temporary growth in the milk secreting tissue in preparation for possible pregnancy. The fact that the probability of breast cancer occurring is greater for single women than for married ones, for women who have low fertility or who have difficulty in breast feeding, and in women who have experienced early menstruation or late menopause, points to hormonal involvement. The actual relationship between hormonal activity and breast cancer is still not well understood and a lot more research has to be done in this area.
The behaviour of breast cancer can differ enormously from one patient to another. It can differ in both the growth rate of the tumour itself and also its ability to metastasis. **Metastasis:** This is the process by which, as a tumour grows in size, individual or bunches of tumour cells can break off from the side of the tumour and proceed to float around the body to possibly start a second tumour somewhere else.

In some patients, tumour growth and its spread is so rapid that death can occur less than twelve months after the initial diagnosis, despite the most rigorous treatments. The opposite extreme, this is most noticeably true for elderly patients, is that the primary tumour grows at an almost unnoticeable rate for a large number of years before giving any trouble. In the growth of breast cancer, there is generally a period during which no lump is perceivable, this period is dangerous as it allows for metastasis to take place even before the primary tumour is initially diagnosed.

1.2.2 Treatment of Breast Cancer

Before the 1950's, the standard method for dealing with breast cancer was to perform a radical mastectomy. This is an operation which involved not only the surgical removal of all the patients' breast tissue but also involved the removal of the pectoral muscles to expose the axilla and the surgical removal of many of the lymph nodes. This sort of radical operation has a number of drawbacks. The first is the cosmetic outcome which is undesirable, and a large scar is left on the chest as a result. The second drawback comes from the fact that some of the lymph nodes that are removed during the procedure form part of the lymphatic system which acts to drain excess body fluid from whichever of the patients' arms is closest to that breast. The removal of these lymph nodes can seriously impede the ability of the lymph system to drain the arm. Another problem caused by this surgery is impairment of muscular control of the arm which may require physiotherapy to regain full functionality. This is because the pectoralis can be effected.
Earlier diagnosis is leading to less radical mastectomy surgery for breast cancers. Other forms of surgery which tend to conserve more of the healthy tissue of the breast area are becoming more common. These types of surgery include the simple mastectomy (also called a local mastectomy) in which only the breast itself is removed without effecting the axilla and also local extraction of the tumour (sometimes called a "lumpectomy") which removes the primary lump but leaves the breast itself intact. The advantages in using these more conservative forms of surgery include the positive psychological benefits to the patient as a result of the improved cosmetic appearance, the fact that the lymph nodes and associated muscles are not touched, means that the patient will not loose any mobility in her arm.

1.3 The Lymphatic System

All the tissues of the body are bathed in a tissue fluid, this fluid contains the diffusible products from the blood and also the waste products produced by cells. Some of this fluid is drained back into the blood capillaries, and the rest is drained in to the lymph capillaries and becomes lymph. The lymph capillaries have walls that are more permeable than those of blood capillaries and so they allow larger particles to enter than blood capillaries do. The lymphatic system is a drainage system that is responsible for carrying away cell waste or things that are no longer required by the body such as particles from the body that are dead or damaged from disease. The lymph system starts as the sort of blind capillary that is shown in the diagram in fig 1.1.
The walls of the lymph capillaries are similar to those of blood vessels being composed of a cell layer of endothelial cells. As the waste tissue fluid flows along the length of the lymph capillaries, the capillaries join into lymph vessels which are similar to veins and possess a similar tissue structure. The lymph vessels contain large numbers of cup shaped valves along their length, these valves act only in one way and prevent the lymph from flowing back along the lymph vessel. The lymphatic system does not possess a pumping system to drive the lymph along through the vessel in the way that the heart drives blood around the body, instead there are several factors which are believed to influence the flow of lymph. These include tissue fluid pressure, contraction of the surrounding muscles, nerve stimulation of the muscles in the lymph vessel walls and pressure due to nearby arteries. The lymph vessel system drains fluid from all areas of the body and directs this flow towards two large ducts, the thoracic duct and the lymphatic duct, both of which empty into the subclavian veins in the root of the neck.

1.3.1 Lymph Nodes

The small and medium size lymph vessels drain through lymph nodes, the lymph nodes are structures that occur at the junctions of lymph vessels, they vary in size, the smallest being about the size of a pin head and the largest being about the size of an
A diagram of a lymph node is shown below.

fig 1.2 Lymph node

The lymph will drain through a number of lymph nodes before entering the blood. A single lymph node may have as many as four or five lymph vessels entering it (called afferent lymph vessels) but they only have one vessel leaving the node (efferent vessel). The lymph node itself consists of a capsule of fibrous tissue that forms an external wall for the lymph node. This fibrous tissue also penetrates down into the node itself forming the partitions inside the node called trabeculae. The rest of the matter inside the node consists of reticular and lymphatic tissue that contain lymphocytes and macrophages.

Because of the larger size of the openings through which material can drain into the lymphatic capillaries and vessels, the lymph system acts to drain away the dead or damaged cell material from damaged parts of the body, this also however enables the lymphatic system to carry infected or cancerous material from one part of the body to another. So because the tumours are themselves drained by the lymphatic system it means that a metastasised tumour cell or a group of such cells which have broken off from the original tumour will be transported around the body by the lymphatic system.
If these cells are capable of multiplying, then if they get caught up in the channels of a lymph node they will settle and begin to do so.

1.4 Accelerators

The linear accelerators that are used in radiotherapy are different in design and construction to those used for high energy physics. The basic components of these machines are shown in the diagram below.

![Diagram of Linear Accelerator Components](image)

*fig 1.3 Linear accelerator (components)*

The linac is composed of two distinct sections, the **stand** and the rotatable **gantry**. The stand contains those components that are needed to operate the accelerator, these are the klystron, the circulator, microwave waveguides, oil tank and the water cooling system.

The klystron is the power source for the accelerator and acts as a microwave **amplifier**, a simplified two cavity klystron is shown below.
The way in which the klystron amplifies the microwaves is that a low power beam of microwaves is injected into the first or "buncher cavity" of the klystron and this sets up an electric field \( E \) inside the cavity that alternates at the frequency of the introduced microwaves. The cathode is heated to produce a continuous stream of electrons that then flow into the buncher cavity, once inside the buncher cavity the electrons are either slowed down or speeded up by the alternating \( E \) field of the cavity, depending on whether they enter the cavity during the positive or negative part of the cycle. After passing through this first cavity the electrons enter the drift tube, the fact that some of the electrons were slowed down while some were speeded up in the buncher cavity means that as the electrons move through the drift cavity the beam breaks up into discrete bunches and the length of the drift tube means that these bunches are well defined by the time they enter the second or "catcher cavity" of the accelerator.

The catcher cavity resonates at the arrival frequency of the incoming electron bunches, ie. an electron bunch enters the cavity just as the alternating \( E \) field of the cavity is in that part of the cycle that will retard or slow down the electron bunch. As the electron bunch is slowed down the energy it loses is taken up by the \( E \) field inside the cavity,
this effect acts to amplify the alternating E field and produces an amplified microwave beam that then leaves the catcher cavity via the waveguide. Any residue of the electron beam that gets through the catcher cavity is taken up by the electron beam collector, its remaining energy is converted into heat which is taken up by the water cooling system. For the high energy linacs used in radiotherapy that require energies of 18MeV or more for photons and electrons, klystrons with up to five cavities are used, the additional cavities give higher current bunching as well as increasing the overall amplification (eg microwave amplifications of 100,000:1 are possible)\textsuperscript{4}.

The circulator is a device that is placed in between the klystron and the accelerator guide structure, it acts as a one way gate for microwaves, it will allow the microwaves to pass from the klystron to the accelerator structure but any microwave that is reflected back from the accelerator structure will be taken inside the circulator and will be channelled for absorption to a dummy load. If the circulator were not present the klystron could suffer permanent damage from microwaves being reflected back from the accelerator and into the klystron.

The waveguides are hollow pipes that are used to transport the microwave energy from the klystron to the accelerator structure. The linac possesses two types of waveguides, the rectangular type is employed to transport the microwaves around inside the stand and also inside the gantry and the circular type with a special connecting seal is used in the connecting section between the stand and the gantry which allows the gantry to rotate. The microwaves are contained within the waveguides by being reflected from the sides and the guides themselves are pressurised with sulphur hexafluoride so that they can transmit high power microwaves without suffering dielectric breakdown. The waveguides are equipped with ceramic windows that are transparent to microwaves, these windows act to separate the pressurised waveguides from the evacuated klystron and accelerator.

The type of accelerator used for the data collection of this thesis was a Varian Clinac
2100C. It is capable of producing electrons of various energies between 6MeV and 20MeV.

It is also capable of producing X-rays of two different energies, these are 18MV (megavolt) X-rays and 6MV (megavolt) X-rays, the 6MV X-rays are produced by electrons that were accelerated across an effective potential of six million volts. So when these electrons reach the end of the accelerator guide, they have a kinetic energy of 6MeV.

When these electrons enter the X-ray target they will then produce Bremsstrahlung X-rays with a continuous energy spectrum from 6MeV down to the keV range. This beam is however referred to as the 6MV beam for the purpose of simplicity. The same argument holds for the 18MV beam.

The X-rays then pass through the beam flattening filter. The beam flattener is a device located in the treatment head of the linac, its purpose is to flatten out the X-ray beam coming from the X-ray target to produce a flat, uniform dose distribution when the beam strikes the phantom medium. The beam flattener works as shown in fig 1.5.
fig 1.5 Beam flattener

The X-ray beam emerges from the target in the shape of a droplet. The shape of the beam flattener is such that it will alter the beam shape to produce the uniform dose distribution seen below the flattener as seen in fig 1.5. As the X-ray beam passes through the beam flattener the high Z material of the flattener will tend to remove the lower energy photons from the beam in preference to the higher energy ones. This means that when the photons exit the beam flattener, those in the middle of the beam will be on average, higher in energy than those on the edge of the beam that went through the thinnest part of the flattener material. When the beam enters the water medium the low energy photons at the side of the beam will tend to interact more with the water and so produce a higher dose than those photons in the beam centre. The resulting dose distribution is seen in fig 1.6. This is the horn effect.
Because the lower energy photons produce shorter range electrons and less forward peaked scatter than the higher energy photons, the low energy photons will deposit their dose more locally in the upper regions of the water tank, so the horn effect is diminished with increasing depth. Most linacs are designed so that the horns will diminish and a uniform dose distribution will be achieved at a depth of 10 cm in the phantom.

1.4.1 Gantry and Alignment Equipment

The accelerator gantry is set so that it can rotate in a circle. The centre point of this circle which the beam will pass through is known as the accelerator’s isocentre.

On clinical accelerators this isocentre point is set to be exactly 100 cm from the X-ray target position (the source as it is known). This isocentre is shown in fig 1.7 over the page.
The isocentre position is marked in space by three positioning lasers. These lasers are set in the accelerator bunker as shown in fig 1.8. The beams of the three individual lasers intersect at the accelerator's isocentre. They are used for positioning of patients. They can also be used by a physicist for research work, for positioning of the water tank or solid water phantom.

The distance from the X-ray source to the skin of the patient is called the source skin distance (SSD). If the patient's skin surface is positioned at the isocentre, the patient is said to be at 100 SSD (100 centimeters from the source of the X-rays).
1.5 X-ray Interaction Process

When a high energy photon such as the 6MV X-ray photons produced by the Clinac 2100C accelerator interact with matter they produce a cascade or shower of particles. These include electrons, primary scattered and multiscattered photons. The complex nature of these showers is very difficult to model by transport equations, therefore Monte Carlo methods are used to model them.

Photons interact with matter by three main processes:

1) Photoelectric Effect
2) Compton Scattering
3) Pair Production
1.5.1 Photoelectric Effect

Photoelectric effect involves a photon being completely absorbed by an atom and this results in the ejection of an electron. The photon interaction must be said to take place with the atom as a whole, and not simply with an electron as this would violate the conservation of linear momentum. This is so because when the photon strikes the atom and the electron is ejected the nucleus recoils slightly. This recoil is not large due to the relatively large mass of the nucleus as opposed to the electron but it will ensure that energy and momentum are conserved.

It is clear that for the photoelectric effect to take place then the incoming photon must have an energy which is greater than the binding energy of the electron which it will eject. For this reason the absorption coefficient curve for the photoelectric effect can be easily distinguished from that of the Compton and pair production curves by the fact that it contains jumps which correspond to the binding energy for different shells. The energies for which these jumps occur for any particular material can be given approximately by Moseley's Law.

\[
E = 13.6 \frac{(Z - \sigma)^2}{n^2} \quad (1.1)
\]

Where \( E \) is the binding energy

\( Z \) is the atomic number

\( \sigma \) is the screening factor

\( n \) is the energy quantum number

It can be seen that this equation is simply the equation for the energy of the Bohr orbit with a modified screening factor added. The values for the screening factor is approximately 1 for the K shell and 3 for the L shell.

Because a third object, eg. an atom or a nucleus is required to fulfil the conservation of momentum the probability of the photoelectric effect taking place will increase rapidly
with increasing binding energy of the electrons. For the loosely bound electrons Compton scattering is more probable than photoelectric effect. Experiments have shown that about 80% of all photoelectric interactions occur in the K shell electrons. As an approximation for high energy photons the photoelectric absorption coefficient, \( \tau_{\text{photo}} \) has a magnitude of

\[
\tau_{\text{photo}} \sim \frac{Z^5}{hv} \quad (1.2)
\]

Where \( \tau_{\text{photo}} \) is the photoelectric absorption coefficient

\( v \) is the photon frequency
\( h \) is Planck's constant

This differs from the corresponding low energy approximation given by

\[
\tau_{\text{photo}} \sim \frac{Z^4}{(hv)^3} \quad (1.3)
\]

and shows that as the photon energy increases the photoelectric cross section decreases but at a slower rate for the high energy photons than for the low energy ones. When the incident photon ejects an electron from the inner energy shells a space is created within this shell for an electron in one of the higher energy shells to fall into, this de-excitation of the atom is accompanied by either the emission of electromagnetic radiation or the emission of an Auger electron. As seen in the chart in fig 1.9 (showing K fluorescence yield vs Atomic number) the probability of emission of a characteristic K X-ray is nearly unity for high Z number elements and is almost zero for low Z number elements which means that the probability of Auger electron emission is high for low Z number elements. Therefore the Auger emission would be the most probable for biological situations.
1.5.2 Compton Effect

Compton scattering is incoherent scattering where some of the photons energy is given to the electron in the form of kinetic energy and the rest appears in the form of scattered photon energy.

The Compton scattering of an electron is shown in the diagram below.

![Compton scattering diagram](image-url)
where \( v_0 \) is the frequency of the incident photon
\( v \) is the frequency of the scattered photon
\( \theta \) is the angle of the scattered photon
\( \phi \) is the angle of the scattered electron
\( p \) is the momentum of the scattered electron
\( T \) is the kinetic energy of the scattered electron

When dealing with scattering it is traditional to consider the electron to be at rest and unbound. In actual experiment these assumptions limit the theory to those cases where the binding energy of the electron to the atom is small compared to the energy of the incident photon, \( h v_0 \). High energy X-rays are used in cancer therapy (i.e. up to 18 MeV). The electrons contained in human tissue are lightly bound. If the energy of the photon is low enough to be comparable with the binding energy of the electron then the photoelectric cross section becomes greater than the Compton cross section and tends to dominate the interaction processes.

Returning to fig 1.10 we can determine the energy and momentum relation between the photons and the scattered electron. The directions of the incident and scattered photons together define the scattering plane. There is no component of momentum which is normal to this plane.

Since the momentum of the photon is given as \( \frac{h v}{c} \) then from fig 1.10 the conservation of momentum yields.

\[
\frac{h v_0}{c} = \frac{h v}{c} \cos \theta + p \cos \phi \quad (1.4)
\]

where \( p \) is the electrons momentum
\( c \) is the speed of light
0 = \frac{h}{c} \sin \theta - p \sin \phi \quad (1.5)

Conservation of energy gives

\[ h\nu_0 = h\nu + T \quad (1.6) \]

Where \( T \) is the relativistic kinetic energy of the electron.

Now it can be shown that

\[ pc = \sqrt{T(T + 2m_0c^2)} \quad (1.7) \]

This relativistic relation maybe used to eliminate some of the parameters from equations 1.4, 1.5 and 1.6 in order to derive various relationships.

These relations can be used to give such information as the change in wavelength which is suffered by the photon, known as the Compton shift.

\[ \lambda - \lambda_0 = \frac{h}{m_0c} (1 - \cos \theta) \quad (1.8) \]

Where \( \lambda_0 \) is the initial photon wavelength
\( \lambda \) is the scattered photon wavelength
\( m_0 \) is the electrons rest mass
\( \theta \) is the angle of the scattered photon
Then the energy of the scattered photon is given by

\[ hv = \frac{m_0c^2}{1 - \cos \theta + \left( \frac{m_0c^2}{hv_o} \right)} \]  

(1.9)

Because it is the electrons which are responsible for depositing a large amount of dose within the medium it is useful to know how energy will be transferred from the photon to the electron. The energy which is gained by the struck electron is given as:

\[ T = hv_o - hv \]  

(1.10)

\[ T = hv_o \frac{\alpha (1 - \cos \theta)}{1 + \alpha (1 - \cos \theta)} \]  

(1.11)

Where \( \alpha = \frac{hv_o}{m_0c^2} \)

The above formula shows that the kinetic energy which is transferred to the electron during the scattering process can range from zero in the forward direction (\( \theta=0 \)) to a maximum of

\[ T_{\text{max}} = hv_o \left( \frac{2\alpha}{1 + 2\alpha} \right) \]  

(1.12)

for an angle of \( \theta \) equal to 180°.

This implies that for maximum energy transfer from the photon to the electron to take place then the scattered photon must be sent back along its own incident path, ie scattered backwards. It can also be noted that the shift in photon wavelength for any particular direction is independent of the energy of the incident photon. However the shift in energy, the equation for which is equation 1.13 is dependent on the energy of the incident photon.
A quantum mechanical description of scattering was given in 1928 by Klein and Nishina when they applied Dirac's relativistic electron theory to the problem.

For the problem of Compton scattering of unpolarised radiation Klein and Nishina gave a differential scattering cross section as

$$d\sigma_s = r_0^2 d\Omega \left\{ \frac{1}{1 + \alpha(1 - \cos \theta)} \right\}^3 \left\{ 1 + \frac{\alpha^2(1 - \cos \theta)^2}{2} \right\} \left( 1 + \frac{\alpha^2(1 - \cos \theta)^2}{1 + \cos^2 \theta(1 + \alpha(1 - \cos \theta))} \right\}$$

(1.14)

Where $r_0$ is the classical electron radius

d$\Omega$ is the differential solid angle

In the above equation the term in the braces is the quantum mechanical correction which was introduced by Klein and Nishina.

The angular distribution of the scattered photons per unit solid angle is given by the equation below:

$$\frac{d\sigma_p}{d\Omega} = \frac{r_0^2}{2} \left[ \frac{v}{v_o} + \frac{v}{v_o - \sin^2 \theta} \right]$$

(1.15)

Where $\sigma_p$ is the photon crossection

If what is required is information about the photon scattered per unit angle, $\theta$ (where $\theta$ is the scattering angle between the incident and scattered photons) instead of photon
scattered per unit solid angle then this is given as

\[
\frac{d\sigma_p}{d\theta} = \frac{d\sigma_p}{d\Omega} 2\pi \sin \theta \quad (1.16)
\]

This gives a very different result from that obtained from \( \frac{d\sigma_p}{d\Omega} \)

The angular distribution of electrons is of particular importance because the ionisation reactions which deposit energy in many radiation detectors is due primarily to the electrons which result either from Compton interactions within the detector material itself or are produced in the material of the detector walls. As a result the directional distribution of Compton electrons coming from outside the detector can have a considerable influence on the response characteristics of a detector ( eg. effects of backscattered electrons ). The angular distribution of electrons coming from the outside of the material can be calculated from the angular distribution of the scattered photons combined with an expression relating the scattered photon angle \( \theta \) to the scattered electron angle \( \phi \), ie. for each photon which is scattered through a solid angle of between \( \theta \) and \( \theta + d\theta \) there will be a corresponding electron scattered with an angle between \( \phi \) and \( \phi + d\phi \) into a solid angle of \( d\Omega' = 2\pi \sin \phi \ d\phi \)

Therefore

\[
\frac{d\sigma_e}{d\Omega} = \frac{d\sigma_e}{d\Omega'} 2\pi \sin \theta \ d\theta = \frac{d\sigma_e}{d\Omega} 2\pi \sin \phi \ d\phi \quad (1.17)^6
\]

The directional distribution of electrons is given as

\[
\frac{d\sigma_e}{d\Omega'} = \frac{d\sigma_e}{d\Omega} \sin \theta \ d\theta \quad \sin \phi \ d\phi \quad (1.18)
\]
The number of scattered electrons per angle can be shown to be

\[ \frac{d\sigma}{d\phi} = \frac{d\sigma_e}{d\Omega} \cdot 2\pi \sin \phi \quad (1.19) \]

So even though \( \frac{d\sigma_e}{d\Omega} \) is peaked in the forward direction equation 1.19 shows that \( \frac{d\sigma}{d\phi} \) is zero in the forward direction (i.e., at \( \phi = 0 \)).

The angle \( \phi \) at which \( \frac{d\sigma}{d\phi} \) shows a maxima is dependent on the energy of the incident photon. The lower the energy of the incident photons, the greater the angular distribution of the scattered electrons.

1.5.3 Pair Production

Pair production is a process where a photon of energy greater than 1.022 MeV interacts with matter to produce a negatron-positron pair. This is an example of energy to mass conversion. This type of interaction is characterised by the fact that a photon is completely absorbed and a negatron-positron pair is produced, the total energy of which is equal to the energy of the original photon.

The process of pair production can only take place within the field of some charged particle, e.g., the nucleus and to a lesser extent the electron cloud surrounding the nucleus. The presence of these other particles ensures that momentum is conserved.

The process is shown in fig 1.11.
When the positron-negatron pair leave the atom they do so with kinetic energies which can be given by

\[ hv > KE_{neg} + KE_{pos} + 2m_0c^2 \]  \hspace{1cm} (1.20)

Where \( KE_{neg} \) is the kinetic energy of the negatron

\( KE_{pos} \) is the kinetic energy of the positron

\( hv \) is the original photon energy

\( m_0c^2 \) is the electrons rest mass

The actual theory which is used to explain pair production is the Dirac electron theory, a simplified way of looking at this theory is illustrated in fig 1.12
The Dirac theory states that empty space is normally populated by a large number of positive energy states which are usually empty and an equal number of negative energy states which are full. In order to lift an electron up out of a negative energy state into a positive energy state requires an energy of at least $2m_0c^2$ joules (in S.I units). If more than this amount of energy is supplied then an electron is lifted up out of the negative energy state to a positive one and a positively charged hole is left in the negative energy state. This positively charged hole has a mass equal to that of the electron and is observed as a positron.

The angular distribution of the positron-negatron pair generated during pair production is usually in the direction forward of the incident photon for high energies. Also for high energies (i.e. $KE \gg m_0c^2$) the angle between the incident photon and the positron and negatron is of the order of $\frac{m_0c^2}{KE}$.

For incident photons of lower energy comparable to $2m_0c^2$ there is less tendency for
the positron-negatron pair to be directed in the forward direction and the distribution is more complicated.

1.5.4 Macroscopic Interactions

When the X-ray penetration of matter is studied on a macroscopic scale, the attenuation of the radiation involves much more than calculating the attenuation using cross-section. The attenuation of a wide beam of applied radiation is also influenced by the so called 'scattering in' effect which causes the amount of radiation observed to exceed the calculated value. To account for this the 'build up factor' has been studied both experimentally and theoretically. This is important as it is always broad beams that are used in radiotherapy. The scattering in factor becomes more important as the size of the incident radiation beam increases. This can be seen from the figure below which shows gamma-ray isodose curves (lines of constant dose) for a particular medium, the attenuation of the beam is less for a larger size fields than for smaller ones.

![Figure 1.13 Beam attenuation for different field sizes](image)

The concept of the 'build up factor' has become widely used as it is important for making calculations. The build up factor is the amount by which the quantity of radiation being studied exceeds the quantity that would be seen if a very narrow beam...
of the same flux were to be used.

The build up factor can be taken into account in the attenuation formula,

\[ D = D_0 B(\mu x) e^{-\mu x} \]

Where \( D \) is Dose

\( D_0 \) is the initial dose

\( \mu \) is the linear attenuation coefficient

\( B \) is the build up factor which is itself a function of \( \mu x \).

The value of the build up factor is also influenced by the nature of the radiation source. One of the possible analytical formulations for the build up factor is expressed in terms of the mean free path and is shown below.

\[ B = A e^{-\alpha_1 \mu x} + (1 + A) e^{-\alpha_2 \mu x} \]

Where \( A \), \( \alpha_1 \) and \( \alpha_2 \) are adjustable parameters and \( \mu \) is the linear attenuation coefficient.

As the value of \( \mu x \) increases so does the value of \( B \), this is because radiation that is scattered in from the peripheral part of the beam is being observed as well as the radiation from the primary beam itself.

As the X-ray beam enters the water phantom it causes electrons to deposit in the way shown in fig 1.14 and 1.15.
Figure 1.14 shows the relative dose deposited in the water phantom as a function of depth, three regions of the dose curve are significant. First the 'build up region' mentioned above can be clearly seen, this is the region in which a growing number of electrons is produced with increasing depth. The second part of the depth dose curve is the equilibrium part at the top, this is the region in which the number of incident photons being absorbed comes to an equilibrium with the number of electrons being created. When the number of photons is compared with the number of electrons, the number of electrons is larger than the number of incident photons.

The final part of the curve is the near exponential drop off in dose that occurs with increasing depth and indicates the absorption of radiation by the phantom material.

Figure 1.15 also shows a dose distribution, this time in two dimensions showing isodose lines. The two straight lines which enter the water at the top and which are seen to continue down past the isodose lines form the geometrical edge of the beam, in between these two lines is the X-ray fluence or photon beam. Once in the phantom
material itself there is a mixture of X-rays and electrons of varying energy. The region immediately outside the geometric edge is occupied by low energy electrons and some low energy photons which have been scattered at wide angles out of the beam.

The isodose lines seen in fig1.15 are lines of equal relative dose, they are normalised to 100% of $d_{\text{max}}$, central axis dose. This position is usually calibrated to give a 100 cGy dose in a water phantom when using a 6MV X-ray beam which is 10 X10 cm at the surface of the phantom.

The way of insuring that the correct dose is delivered at this depth is by instructing the accelerator to deliver the appropriate number of monitor units. The accelerator delivers X-rays in units called monitor units (mu), when the accelerator is calibrated it will be set so that 100 mu will deliver 100cGy at a depth of 1.5 cm below the water phantom surface for a 10x10 cm 6MV X-ray beam, see fig 1.16.

![Diagram of 10x10 X-ray beam in water phantom](image)

fig 1.16 10x10 X-ray beam in water phantom

This is only one example of how to set up a dose calibration. For example calculations can be done in air with the use of a buildup cap. Also the dose can be calculated so that 100mu equals 100cGy at some point other than 1.5 cm deep for 6MV.
The mu system is used because the mechanical and electrical complexity of linacs means that each individual machine will deliver a different dose at a different rate. By adjusting the monitor unit sensitivity all linacs can be made to give a standard dose to within a reproducibility of about 2%.
Fig. 1.17 and fig. 1.18 show the relationship between depth dose curves and the dose profiles. This is important because all of the dose profiles that appear in chapter four are the results of measured profiles that are analysed. What the dose profile shows is the relative dose as a function of beam width (this is the dose to 50% width of the beam not the geometrical width). These beam profiles can be divided into three regions as shown in fig. 1.19.

![Diagram of beam profiles]

**fig. 1.19** A dose profile

The umbral region is the flat uniform distribution in the centre of the beam, the umbral height also decreases with increasing depth as seen in fig. 1.17 and fig. 1.18. The penumbral region is where dose drop off occurs, it is the region which lies very close to the geometrical edge of the beam. If the beam steering is aligned correctly the geometrical edge of the beam will in general lie at the point that is marked as 50% of the maximum dose profile height as shown on fig. 1.19. At the lower part of the curve the penumbra has a low dose penumbral tail, this tail lies outside the primary photon fluence of the beam and is a result of the electrons and low energy photons that have been scattered out of the beam. When higher energy X-ray beams are used for clinical treatment it is most often the case that more than one field will be used to treat a certain
area of the patient. This technique is known as multiple field treatment and may involve the use of two, three, four or more fields. A treatment technique for breast cancer patients is shown in the diagram below.

fig 1.20 Beam set up for breast cancer treatment

This is a three field treatment involving two tangential fields which come from either side of the patient to treat the actual breast tissue and the third beam, called the supraclavicular beam, is directed down to a point above the breast to treat any cancer cell metastases that maybe present in the lymph nodes which exist in this region.

The common line along which the edges of these three fields match is known as the matching line and it is the dose distribution along this matching line which is the subject of this thesis. An accurate knowledge of this of the dose along this matching line is critical because, as is demonstrated in chapters three and four, any misalignment of the treatment fields will lead to underdosing or overdosing of the patient.

In order to produce each of the field dose matching profiles that are seen in chapter four an experimental dose profile was taken in a water phantom. This profile data was then transferred to a computer where one half of the profile was matched with its own
mirror image. The mirror image of each half profile was produced by effectively rotating the experimental data points through $180^\circ$ about a vertical axis which passes through the 50% of maximum dose point seen in fig 1.19. The respective dose values from the experimental profile and its mirror image were then added to produce the total curve which is seen at the top of the two profiles shown in fig 1.21.

![fig 1.21 Summation curve for two matched profiles](image1)

This will simulate the effect of matching two individuals beams. The isodose and geometrical edge of two such matched beams are shown below.

![fig 1.22 Isodoses for two matched profiles](image2)
Fig 1.22 shows two matched asymmetric profiles. The purpose of this study was to look at the matching of asymmetric profiles that are produced using asymmetric collimators or independent jaws as they are also known.

A typical profile shape for an asymmetric field is shown in the schematic below.

![Asymmetric Profile Schematic](image)

The profile in fig 1.23 is an asymmetric profile produced by driving an asymmetric collimator to the centre of the beam and then blocking off half of the beam.

1.6 Advantages of Asymmetric Collimators

Because of the popularity of conservative breast cancer treatments, radiotherapy treatments have evolved which involve a complex arrangement of treatment fields. This has been necessitated by the need to apply treatment fields to irregular target volumes which can be located close to critical organs and also by the need to treat the lymphatic nodes around the breast area in order to prevent the cancer from spreading around the body.

One of the major problems in the use of multiple field treatments is that the diverging
field edges tend to overlap at depth and lead to the formation of what are known as dose islands (regions of overdose in the patient's body) which can produce organ damage, fibrosis and cosmetic damage. Overdosing due to the matching of diverging radiation fields is shown in the diagram below.

![Diagram of overdosing for divergent beams](image)

**fig 1.24 Overdosing for divergent beams**

At the other extreme mismatching of the treatment fields can lead to underdosing in the matching region. This means that only a fraction of the prescribed dose is delivered to the patient along the line where the fields are matched. This will lead to a lack of tumour control at this point.
Asymmetric Collimators are essentially large mobile tungsten blocks which are situated in the head of the linac as shown below. They are mechanically driven and unlike conventional blocks they don't have to be placed and fixed to a block tray holder.

![Accelerator head diagram](image)

**fig. 1.25 Accelerator head**

The asymmetric collimator or independent jaws as they are also known can be driven independently to block off a portion of the X-ray beam. The advantage in this is that the asymmetric collimator can be driven to the centre of the treatment field, thus blocking off half of the X-ray beam. This will produce a non-diverging edge on the beam as shown below.

![Driven Collimator diagram](image)

**fig 1.26 Asymmetric Beam**
The non diverging field edge seen in fig 1.26 can be used as the matching line for multiple field treatments. The use of asymmetric collimators also means that in some cases when treatment fields are matched neither the couch nor the collimators need to be rotated, this further decreases patient setup time, the reason for this is that the asymmetric edge of the beam is projected straight down onto the surface of the patient where as the diverging edge comes down at an angle onto the curved surface of the patient as shown below. As the straight edge of the treatment field allows for more accurate setting up of the patient a more uniform treatment of the patient should result.

As can be seen in fig 1.23, the dose in the umbral region falls off steadily as one moves from the area of the midfield across to the penumbral region. This dose drop off is characteristic of the asymmetric fields but its cause is not well understood. It is possible that the effect is due to the relationship between the flattening filter and the jaw at central axis. Because the beam is harder at central axis the scatter
distribution is more forwardly peaked\textsuperscript{21}.

Therefore the chance of an isotropic scatter distribution which would compensate for
dose loss due to the collimators is reduced near the central axis.
Chapter 2

Radiotherapy Dosimetry Methods
2.1 Introduction

In the experimental work that was done with asymmetric collimators four different types of detectors were used to achieve the results. These detectors included two ion chambers, a diode detector and TLD detectors.

The Farmer chamber which has been used as a standard means of measuring clinical radiation dose since the 1950's and the RK chamber which is also an ionisation chamber of the thimble type and which is used to measure radiation in a water tank phantom.

As well as ionisation chambers there were also solid state type detectors that were employed. These detectors consisted of a p-type diode that could be used in the water tank phantom in the same way as the RK chamber. The other solid state type detector used were Thermoluminescent detectors (TLD's) which were used in both the solid water experiments and in the Rando phantom experiments.

2.2 The General Properties of Radiation Detectors\textsuperscript{11} (except TLD's and Film)

If radiation entering a detector is to make that detector respond, then the radiation must undergo some type of physical interaction within the detector. The types of interactions which are generally suffered by radiation are the Photoelectric effect, Compton effect, and Pair production as mentioned in chapter 1. Although other types of interactions are possible they are rare when compared to these three for energies in the megavoltage range. All detectors have an interaction or stopping time, the time it takes for an ionising particle to liberate all its energy in ionising reactions, this is typically of the order of picoseconds for gas detectors and nanoseconds for solid state detectors. This time is so short that for normal counting situations it can be considered instantaneous.
All these detectors work on the principle that the incoming radiation will generate a charge $Q$ within the specific detectors active volume, the active volume being that part of the detector which actually detects the radiation. An electric field applied across this active volume will make the negative and positive charges that are generated there travel in opposite directions to be collected at the edge of the active volume, the collected charge $Q$ will then form an electrical signal which is recorded.

As the charge $Q$ is collected it flows as a current through the electrical circuits of the detector. This current must then be integrated over time for a duration equal to the collection time in order to give the original ionisation charge $Q$, therefore:

$$\int_{0}^{t_c} i(t) \, dt = Q.$$ 

The time interval between the arrival of specific current pulses is random as is the arrival of the incident photons, the arrival of which is governed by Poisson statistics and the height and duration of each current pulse is dependent on the type of interaction which produced it. As a further complication real detectors have to deal with a large amount of radiation producing ionisation within the active volume in rapid succession, hence situations may arise when current from more than one interaction may flow at any one time.

There are three main ways in which a detector collects information, pulse mode, current mode and mean square voltage mode. In pulse mode single events are recorded, that is they are theoretically designed to register each individual quanta of radiation. Since the energy of the incident radiation is proportional to the charge $Q$ created within the active volume then each burst of current that is generated by the incident quanta is time integrated to give back the charge $Q$. All detectors which are used in spectroscopy analysis are operated in pulse mode.
When very large numbers of radiation quanta are incident on the detector the rate at which pulses are formed may be too fast for the pulse mode method to work effectively, that is individual events will overlap. When this happens other means of recording must be employed. The mean square method mode is a method similar to current mode in that it looks at averages. However the mean square mode acts to block out the average current and only look at the fluctuation in the current for a fixed time $T$. The mean square fluctuation signal can be interpreted as shown below.

$$\sigma^2 = \frac{rQ^2}{T}$$

Where $T$ is the response time

$r$ is the event rate

$Q$ is the charge produced per event

From the above formula the charge per event can be calculated. This method can also be used to distinguish between different types of radiation entering the detector.

2.3 Ionisation Chambers

The types of ionisation chambers which were used in this thesis work were of the thimble type design. A typical thimble chamber is shown below.

![Thimble Chamber Diagram](image)

fig 2.1 Thimble chamber
The thimble wall of the chamber is made to be as air equivalent as possible, i.e. it is made so that its effective atomic number \( Z \) is as close to that of air as possible. The wall is also made thick enough so that electronic equilibrium can be achieved within the active volume (cavity). The interior surface of the thimble wall is coated with a conducting film and so comprises one electrode. A rod made up of low atomic number such as graphite or aluminium, is located along the centre line of the thimble chamber and is isolated from the thimble wall, this forms the second electrode. A voltage is applied between these two electrodes in order to collect the ions. One disadvantage with the use of thimble chambers is that they cannot measure the exact exposure, this is because the thimble wall is not quite air equivalent, its cavity volume is not accurately known and its wall thickness may not be sufficient to set up electron equilibrium\(^1\)\(^2\).

In practical situations thimble chambers are calibrated against either free air ionisation chambers or against other standard thimble chambers, a calibration factor can then be used to correct the thimble chamber so that the reading from the chamber can be given as a dose estimate.

For medium to high energies a buildup cap is placed over the chamber to produce the necessary electron equilibrium for 'in air measurements'.

At high X-ray energies the radius required for the buildup caps is so large that it restricts the minimum field size for measurements. Examples of high energy buildup caps are 1.5 cm radius of solid water for 6 MV X-rays and 3.3 cm radius of solid water for 18 MV X-rays.

For the purpose of this thesis, measurements were carried out using two different types of thimble detectors, called the Farmer chamber and the RK chamber.

The Farmer chamber was invented by F.T. Farmer in 1955, its purpose was to give a reliable and stable secondary standard for both \( \gamma \)-rays and X-rays in the therapeutic
range. The original chamber was modified to give better energy response and consistency of construction by Aird and Farmer. A diagram of the Farmer chamber is shown in fig 2.2.  

![Diagram of Farmer chamber]

fig 2.2 Farmer chamber

The central electrode in the above diagram is pure aluminium and the thimble wall is graphite, the insulation material is polytrichlorofluoroethylene and the active volume of the chamber used for this thesis is nominally about 0.6 cc. Farmer style chambers are commercially available, these follow the design of the one seen in the above diagram, however they differ in the material used in the construction of the thimble wall and the central electrode.

The second thimble type chamber used for this thesis was the RK chamber. The RK is cylindrical in shape and is designed for use in the water tank phantom. The air inside the cavity is free to communicate with the air outside the cavity by means of an air tube that travels along the electronic signal cable. The RK is designed to measure both electron and photon beams of energy typically greater than 1MV. It also comes with a perspex cap which has a density of $0.45 \pm 0.05$ g/cm$^3$ which can be placed over the RK so that measurements can be performed in free air.

The active volume of the air cavity of the RK has a volume of 0.12cc and the inner cavity wall is coated with a mixture of 50% epoxy resin and 50% graphite to make up the polarising electrode.
2.4 Solid State Detectors

Solid state (semiconductor) detectors have a number of advantages over the gas filled variety. They have superior energy resolution to the gas detectors, they are more compact in size and their increased density means that they are capable of stopping higher energy radiation quanta than gas detectors. One negative aspect of semiconductor operation is their susceptibility to radiation damage.

2.4.1 The Effect of Ionising Radiation on Semiconductors

When ionising radiation enters a semiconductor it generates electron-hole pairs on its way through. The radiation can create these pairs either by direct ionisation of the material or by an indirect method where the electron that was ionised by the incoming radiation (these electrons are called delta rays) go off and cause ionisation interactions of their own. A quantity of interest in the use of solid state detectors is the energy that is required to create one electron-hole pair, this is termed the ionisation energy and is given the symbol $\varepsilon$. The number of electron-hole pairs that are produced can be interpreted in terms of the incident radiation.

An advantage of semiconductors is that their $\varepsilon$ value is very small ($\varepsilon \approx 3$ eV for silicon/germanium as opposed to $\varepsilon \approx 30$ eV for a typical gas filled container). This means that the number of electron-hole pairs that are produced in the semiconductor will be ten times the amount of ion pairs that are produced within the gas filled detectors, this fact will both lower the statistical fluctuations and offer an increased signal to noise ratio for the detectors.

As well as the mean number of charge carriers produced by incident radiation the fluctuation or variance in the number of carriers is also important because it relates to the energy resolution which is achievable with a particular detector. The fluctuations in the number of electron-hole charge carriers in semiconductors is less than that predicted by the Poisson distribution. In order to relate the observed variance of the
charge carrier (electron plus holes) numbers to that predicted by the Poisson distribution the Fano factor \( (F) \) is introduced\(^{11} \).

\[
F = \frac{\text{observed statistical variance}}{\frac{E}{\epsilon}}
\]

where \( E \) is the total energy of all the charge carriers produced

\( \frac{E}{\epsilon} \) is the total number of charge carriers produced

In order to achieve good energy resolution the Fano factor should be as small as possible.

All radiation detectors record radiation by means of their active volume, the active volume of a semi conducting detector is its depletion region. A diagram of the depletion region is shown below\(^{14} \).

The depletion region is the area in between the n-type and the p-type parts of the diode, any electrons which happen to drift into the depletion region from the n-type donor
material are neutralised by the holes which drift across from the acceptor region. When this happens an electric field is built up in the depletion region which is in the direction from the n-type to the p-type region. This field then acts to prevent further drift of charge carriers into the depletion region and in this way the depletion region acts like the spacing between two capacitor plates. Achieving good detector characteristics requires a small capacitance and the ability to move the charge carriers that were created in the depletion region by the incident radiation out of the depletion region rapidly. A diode which is being used as a solid state detector of radiation can be used in either a biased or unbiased mode.

As the diode detectors used in this study were not reverse biased they work in the same way as a solar cell. The photon energy which is incident on the detector serves only to generate free electron hole pairs. These charge carriers then migrate around in the semiconductor due to the influence of localised fields there. They then tend to congregate in certain regions and build up a photovoltage. The local fields which act to drive the charge carriers are the results of discontinuity's within the crystal lattice or to nonuniform doping, most of these fields occur in random directions resulting in zero total effect.

When a photon interacts with a semiconductor to produce an electron-hole pair, the electric field which exists across the depletion region of the p-n junction tends to drive the electrons and the holes in opposite directions. The electron is sent towards the n-type region and the hole is sent towards the p-type region. This separation of the charges will result in a potential difference being set up across the depletion region. This is shown in fig. 2.4.
The potential which is set up by the photon interaction is given as $V$ and acts to forward bias the p-n junction. This leads to a lower potential barrier ($\phi - qV$). If the p-n junction is now connected to an external load resistance which is low a photocurrent can be read. If $I$ is the current which does flow through the junction region under the influence of $V$ and if $I_o$ represents the total current which could flow through the junction under thermal excitation, then the current $I$ is given by the formula,

$$I = I_{sc} - I_o \left( \exp \frac{qV}{kT} - 1 \right).$$

$I_{sc}$ is the short circuit current that is produced when the irradiated semiconductor is short circuited and the only current flowing in the circuit is that which is produced by the electric field in the depletion region$^{15}$.  

fig 2.4 Diode potential barrier
2.5 Thermoluminescence Dosimeters (TLD's)\textsuperscript{16}

These consist of inorganic crystals which work much like solid state detectors, the main difference is that once the electron-hole pairs have been created they do not immediately recombine nor are they carried off by an electric field to be read. Instead of producing a continuous energy level within the forbidden region like semiconductors do, TLD's have discrete electron trapping centres and discrete hole trapping centres within the forbidden band gap. These traps occur in the band gap at sites in the crystal lattice where there is either a defect in the normal crystal structure or an impurity atom has been doped into the lattice. The resulting electron-hole traps in the band gap are shown below.

![Diagram of electron-hole traps](image)

fig 2.5 Electron/hole traps

When radiation is incident on TLD material, electrons in the valance band are promoted up into the conduction band from where most of them fall straight back down into the conduction band, but a few are caught in the trapping sites. If these electron trapping sites are very close to the top of the band gap then the electrons contained within them may have enough thermal energy at room temperature to escape from the trap to the conduction band. From there they drop down and recombine with a hole in the valance band, thus releasing energy. This process of emptying these high level (thermally unstable) traps can be done deliberately by the process known as preannealing where
the TLD's are heated to a temperature of 100 degrees Celsius for one hour. When this is done before the actual reading process is carried out this leads to less variance in the dose read. The number of electrons and holes that are found in trapping centres is proportional to the exposure that the TLD has received, TLD's therefore act as integrating detectors for radiation dose.

For the purposes of reading the TLD's they are placed in a TLD reader which heats them to a very high temperature (eg 400 degrees Celsius). This supplies sufficient thermal energy necessary to free the remaining trapped electrons. Electrons released from the traps then recombine with the holes and give off photons which are read by a photomultiplier placed within the TLD reader. The intensity of the photons released from the TLD during the heating process is then recorded as a function of temperature of the TLD to produce what is known as a "glow curve". The radiation exposure is related to the total number of photons emitted or the area under the glow curve. The type of TLD used in the experimental work for this thesis were LiF TLD (100) Harshaw chips although other types (CaSO₄ or CaF₂) are available. LiF TLD's have a low average atomic number (Z = 8.31)¹⁹ which does not differ greatly from that of water (Z = 7.4) and the actual energy deposited in LiF TLD's corresponds to a wide range of X-ray energies. LiF is sensitive to doses as low as 100 μGy and its sensitivity remains linear for dose up to about 1Gy. From 1Gy to about 10Gy the sensitivity starts to become nonlinear²⁹ and at higher doses the material will display a nonlinear increase in the dose response. This behaviour is termed "supralinearity"¹⁶.

2.6 Film¹¹,¹⁷

Photographic film has been used to record ionising radiation since X-rays were discovered in the late 1890's.

The films are made up of an emulsion of silver halide (mostly silver bromide) grains that are suspended in gelatine and are supported by a cellulose acetate film backing.
The film works by grains becoming sensitised by interaction with radiation. These sensitised grains then form a latent image on the film. The development process involves the whole sensitised grain being converted into metallic silver, this increases the number of effected molecules until the whole grain becomes visible. When the image has appeared on the film it is washed in a fixer solution which dissolves all the remaining silver halide away, the film is then worked again to remove the developing chemical residue.

Although film can be used to study individual particle tracks, in radiotherapy it is used to record the cumulative darkening caused by multiple particle interactions.

In order to be recorded the photon must interact with an electron in the emulsion of the film, the direct interaction probability is small and so the film is relatively insensitive for high energy photons, however it is more sensitive to lower energy photons.

To increase the sensitivity of the film it can be sandwiched between two metal plates of high atomic number, the electrons that are produced by either photoelectric effect or Compton effect within the metal plates produce secondary interactions within the emulsion, also light emitting phosphorus of high atomic number can be placed near the film to produce a greater amount of film darkening when exposed. These sensitising agents can be needed because at low exposures there are two few grains exposed to produce any form of measurable image, at high exposures dense concentrations of developed grains can occur this will reduce the measurable image quality.

Exposure in between these extremes results in a film response that is approximately linear and this is the normal operating region.

For quantitative X-ray dosimetry to be carried out the development of the film must be standardised and constantly controlled, also the use of a blank (unexposed) control film helps correct for background effects which are the result of storage and shipping of that particular batch of films. For most films the minimum level of exposure which can be observed above the background (fog) level is about 0.4mGy. A films
sensitivity to photons of different energies depends on its grain size and grain spacing, with the larger and more sensitive grains detecting the higher energy photons.
Chapter 3

Forming Functions
3.1 Introduction

The concept of a dose profile was introduced in chapter one, the usual method of obtaining such a profile is to scan a small radiation detector such as a diode or a small ion chamber laterally across the X-ray beam at a specific depth and so produce the type of dose profile curve seen in fig 1.19 in chapter one. Fig 1.18 in chapter one shows that the height of the dose profile decreases due to increased attenuation of the X-ray beam at depth in the phantom. It also shows how the dose profile widens with increasing depth due to the diverging X-ray beam.

Dose profile curves are used in radiotherapy planning computers to estimate the dose that will be given to a patient.

For this reason any mathematical function capable of reproducing the type of curve seen in fig 1.19 and which can be adjusted for different phantom depths would be of considerable use in radiotherapy. The types of functions which do this are known as forming functions and two such functions which are currently used are the cosine function and the Exponential function.

3.2 The Cosine function

The form of the cosine function is shown below (ref. GE planning system manual)

\[ P = \frac{\cos \left( \frac{1}{s} \times 180 \right) + 1}{2} \times (1 - F) + F \quad \text{----------3.1} \]

Where \( t \) represents the distance from the unblocked edge across the penumbra

\( s \) is the effective diameter of the source

\( F \) is the block transmission factor

The Cosine Function is currently used as the forming function on the GE planning system.
The weakness of the cosine function is that it is not tunable to closely match the penumbral region of the dose profile as shown in fig 3.1 of a half profile. (For more details see fig 3.8)

fig 3.1 Region of Cosine fit

In order to deal with the whole half profile seen in fig 3.1 the profile must be divided into three regions.

3.3 The Exponential Function

The Exponential function shown in equations 3.2a and 3.2b is taken from the book, 'The Physics of Radiology' by Johns and Cunningham 4th edition, page 371
\[ F(d,x) = 1 + \frac{1}{2} \exp \left( - \frac{\alpha_1}{p} (|x|) - |x| \right) \] for \(|x| < \frac{W_d}{2}\) \hspace{1cm} 3.2a

And

\[ F(d,x) = t + \left( \frac{1}{2} - t \right) \exp \left( - \frac{\alpha_2}{p} \left( |x| - \frac{W_d}{2} \right) \right) \] for \(|x| > \frac{W_d}{2}\) \hspace{1cm} 3.2b

where \(\alpha_1, \alpha_2\) are arbitrary constants

\(W_d\) is the width of the beam at a depth \(d\)

\(p\) is the geometrical penumbra of the beam

\(t\) is the X-ray transmission through the collimators

The exponential function unlike the Cosine function does not describe the penumbral dose profile curve in continuous fashion, but in two distinct regions (upper and lower), hence the need for two separate equations 3.2a and 3.2b. The method is also known as the double exponential forming function.

These regions are shown in fig 3.2

In addition, the exponential function also possesses two arbitrary parameters \(\alpha_1\) and \(\alpha_2\), the values of which must be selected to match the experimental data. These
parameters alter the sharpness of curvature of the penumbral profile at the top ($\alpha_1$) and at the tail ($\alpha_2$) respectively.

3.4 The Symmetric Square Root Function

For this thesis an attempt was made to find an alternative function which could accurately describe the dose profile curve. It had to be well behaved (was everywhere continuous) and possess the least number of arbitrary parameters possible.

One function which looked promising was found in the book 'Biophysics'. This function was adjusted to fit a symmetric profile at a given depth, resulting in equation 3.3.

\[
\text{Dose} = D_0 \left(1 + \frac{x_0 - |x|}{\sqrt{(x_0 - x)^2 + n^2}}\right) + D_1 \quad 3.3
\]

Where $\text{Dose} =$ The relative dose of the profile

$D_0 =$ The half maximum relative dose of the umbral region (ie. half the height of the umbra)

$x_0 =$ The location of the 50% of maximum dose position

$x =$ Distance across the profile from zero

$D_1 =$ Sum of the transmission dose through the blocks (assumed to be constant) and the dose appearing in the penumbral tail due to photons scattered out of the beam.

$n^2 =$ An arbitrary parameter which must remain small compared to the value of $(x_0 - x)^2$ in order to make the penumbra the correct shape.

Equation 3.3 will model the profile data for a given depth, ie the depth at which the dose
given to the medium is $2D_o$.

The only arbitrary constant in this formula is the constant $n^2$, the value of $n^2$ is very important as it determines the slope of the penumbra. This can be seen in fig 3.3 which shows the half profiles of equation 3.3 for different values of $n^2$.

![Half Profile](image)

fig. 3.3 Square root function for different $n^2$'s

The reason why the value of $n$ is squared and not simply $n$ is for dimensional reasons. The value of $(x_0 - x)^2$ in the bottom line of equation 3.3 has the dimensions of length squared and so any number added to it must also have these dimensions.

In order to achieve a practical penumbral fit the value of $n^2$ must remain small when compared to the value of $(x_0 - x)^2$ in the bottom line of equation 3.3. The value of $n^2$ would be a function of the width of the X-ray beam and also of the amount of lateral electron scattering out of the beam. The second constant $D_1$ is a function of the dose
deposited within the medium by those X-rays that are transmitted through the collimators, this is taken as a constant. \( D_1 \) is also a function of the X-rays which are scattered out of the beam and lose their energy in the surrounding medium.

A standard way of characterising penumbra is to study their widths. The usual way of doing this involves measurement of the lateral distance between the 80\% of maximum dose and the 20\% of maximum dose points. This distance is referred to as the 80-20 (eighty-twenty).

Another measure of the penumbra is the lateral distance between the 90\% of maximum dose and the 10\% of maximum dose. This is known as the 90-10 (ninety-ten).

Examples of both these measurements are shown by various authors\(^2^1\).

Both the 80-20 and 90-10 are shown on the penumbra in fig 3.4 below.

For equation 3.3 the calculation of \((x_1)\) is straightforward.
\[
D = D_0 \left(1 + \frac{(x_0 - |x|)}{\sqrt{(x_0 - x)^2 + n^2}}\right) + D_1
\]

\[
\frac{D - D_1}{D_0} = 1 + \frac{(x_0 - x_1)}{\sqrt{(x_0 - x_1)^2 + n^2}}
\]

\[
\left(\frac{D - D_1}{D_0} - 1\right)^2 = \frac{(x_0 - x_1)^2}{(x_0 - x_1)^2 + n^2}
\]

\[
n^2\left(\frac{D - D_1}{D_0} - 1\right)^2 = (x_0 - x_1)^2 \left(1 - \left(\frac{D - D_1}{D_0} - 1\right)^2\right)
\]

\[
\text{LET } k^2 = \left(\frac{D - D_1}{D_0} - 1\right)^2
\]

\[
(x_0 - x_1) = \frac{k\sqrt{\frac{1}{1 - k^2}}}{n}
\]

This will give the value of x1 shown in fig 3.4

The calculation for (x2) for the 20% of maximum dose follows in the same fashion.

As seen in equation 3.4 the value of the 80-20 (or 90-10) is dependent only on the value of n, the arbitrary constant of the function.
3.5 The Asymmetric Square Root Function

Modelling of the asymmetric profiles presents a different problem, the profile is no longer symmetric about the y-(dose) axis but as explained in chapter 1 the dose decreases across the umbral region of the blocked side of the beam as seen in fig 3.5.

fig. 3.5 Asymmetric profile with block

Forming functions that are currently used have difficulty in modelling this dose drop off, for example the exponential function is shown along with actual measured data in fig 3.6.
As seen in fig 3.6 the exponential function (the solid line) does model well the symmetric data shown by the diamond shapes. But it does not model well the asymmetric data shown by the circles.

This data fit was obtained using a program called Sigma Plot\textsuperscript{30} which employs the Marquardt-Levenberge\textsuperscript{31} fit.

In order to make the square root function model asymmetric profiles, an adjustment had to be made to equation 3.3. Because of the asymmetry of the profile, the absolute value sign in equation 3.3 had to be removed so now the adjusted square root function would only model that half of the profile which is blocked by the collimator.

The adjusted equation is the equation 3.4.
Dose = \( (D_0 - kx) \left( 1 + \frac{(x_0 - x)}{\sqrt{(x_0 - x)^2 + n^2}} \right) \) + D_1 \quad 3.4

Where the \((-kx)\) term will produce the downward slope in the umbral dose seen in fig. 3.5.

If eq. 3.4 were allowed to include negative values of \(x\) then the dose profile produced will not be flat on the negative side of the \(x\) axis as is shown in fig. 3.5. For this reason eq. 3.4 can only handle the asymmetric half of the penumbra.

This factor will also take account of the lower dose in the penumbral tail region as the \((-kx)\) term continues to alter the profile in the region beyond the penumbra. Figure 3.7 shows how this adjusted square root function fits both symmetrical and asymmetrical data. Both the symmetric and the asymmetric data for this fit were taken from actual data produced using the water phantom and the red diode detector at the ICCC.

![Adjusted square root function and measured data](fig. 3.7 Adjusted square root function and measured data)
Figure 3.7 as well as showing how the square root function matches the experimental data also suggests a possible physical meaning for the value of k. Since k is a positive number for the asymmetric profile (when one of the independent jaws has been driven to the centre of the field) and has a value of zero for symmetric fields. This indicates that k is a function of the distance that the collimator is driven into the field, however further analysis would be required to validate this assumption.

Figures 3.8 and 3.9 which show how this function compared with the Cosine fit, the exponential function, and how all three functions are compared with measured data.

**Cosine and Square Root Fit matched to Data**

![Graph showing Cosine and Square Root Fit matched to Data](image)

Fig 3.8 Cosine function, square root function and measured data
The square root function appears to be a good option as a forming function based on fig 3.8. Because the square root function is continuous this means that it is more symmetric which would be a disadvantage if the head and tail of the penumbra mismatch.

However being continuous also gives it advantages for the reasons already mentioned in the chapter.

**Exponential and Square Root Fit Matched To Data**

![Graph showing Exponential, square root function and measured data](image)

fig 3.9 Exponential, square root function and measured data

In the clinical application of high energy X-ray beams one of the most common forms of treatments involves multiple field techniques. As explained in chapter 1 this involves the use of a number of separate X-ray fields to treat one patient, in practice some of the edges of these fields lie on a common line called the matching line. This matching line
is a sensitive region as it will receive dose contributions from more than one field and as a result is prone to overdosing or underdosing depending on how well the different fields are aligned. The results of matching two fields which are only slightly misaligned are shown in chapter 4.

As a result of this any function which is designed to model the behaviour of the dose profile must also accurately model the profile behaviour in a matching region.

The remainder of this chapter shows the results of matching two profiles of the cosine function, exponential function, the square root function and comparing the results of the summation curves produced with those of actual data taken from matched fields.

In order to see the ability of each of the forming functions to produce a reliable result for the region where two X-ray beams were to be matched, half profiles of all three forming functions were taken and added to their respective mirror image half profile. The results of these summations were then compared with the results obtained by performing the same procedure on half profiles obtained from performing actual experiments.

The graphs shown below show the results of comparing the measured symmetric data for the exponential forming function, the Cosine function and the square root function. The graphs show summation curves generated by adding the half profiles which are produced by the different forming functions and the measured data to their respective mirror images. The measured data shown below is taken for data obtained using the Scanditronix RFA 300 (see section 4.11), the data was taken for 6MV beams at 1.5cm deep with a 10x10 field size and using a diode detector device.
fig 3.10 Summation curves for Cosine, square root and measured data

fig 3.11 Summation curve for Exponential, square root and measured data
As seen in these figures for the matching of symmetric profiles the exponential function tends to coincide best with the measured data.

The next two figures show the summation curves of two asymmetric profiles. Like the symmetric summation curves in figures 3.10 and 3.11 they show the correlation between actual measured data, exponential function, the Cosine function and the square root function. Figure 3.12 shows the matched summation curves for the measured data, the Cosine function and the square root function. Figure 3.13 shows the relation between the exponential function, the measured data, and the square root function.

fig 3.12 Summation curve for cosine, adjusted square root function and measured data
fig 3.13 Summation curve for the exponential, adjusted square root function and measured data

As seen in figures 3.12 and 3.13 the square root function tends to match the measured data better than both the Cosine and exponential function.

3.6 Further Adjustments

Equation 3.4 still has only been experimentally examined at a single depth. In order to describe the way the dose profile changes with increasing depth two adjustments, would be necessary to the formula.

The first would be the adjustment for varying depth. This would involve multiplying the first term on the right hand side of equation 3.4 by another function, call it I(d) which models the depth dose curve seen in fig. 3.14.

This would adjust the height of the profile to take account of the depth in the phantom in which the profile was taken.
Fig. 3.14 also shows in schematic the ionisation fall off with increasing depth for increasing field size. This is due to the scattering in factor discussed in chapter 1. This means that the function I(d) would also be a function of the field size, ie I(d,s), it would also be a function of energy.

The second adjustment to the function is required because as shown in fig 1.16 the beam diverges with increasing depth, which means that the profile will get wider as well as shorter as the phantom depth increases. This second adjustment is simpler than the one just mentioned for the depth dose, because it only depends on the profile geometry. The calculation to allow for beam divergence with increasing depth is shown in fig 3.15.
fig 3.15 Adjustment for divergence with depth

\[
\frac{x_0 - x_c}{d_0} = \tan \theta
\]

\[
\frac{x_1 - x_0}{d + d_0} = \tan \theta
\]

Therefore

\[
\frac{x_1 - x_0}{d + d_0} = \frac{x_0 - x_c}{d_0}
\]

Thus

\[
x_1 - x_0 = \left(\frac{d + d_0}{d_0}\right)(x_0 - x_c)
\]

This could then be substituted into equation 3.3 or 3.4 to adjust for beam divergence at depth.
Chapter 4

Methods and Results
4.1 Introduction

When treating patients with high energy X-rays, quantitative assessment of the dose distribution of the radiation throughout the patients body is required. To obtain this information phantoms are used, these are devices which simulate normal human tissue. The design of phantoms can range in complexity from a square tank filled with water ( which simulates human tissue because of the large water content of human tissue ), through so called solid water which consists of blocks of solidified resin made to have the same electron density as water, up to anthropomorphic phantoms such as the Rando phantom which consists of an actual human skeleton which is encased in a synthetic rubber-like compound which has the same electron density as tissue.

The types of phantoms used in this study were the water tank phantom ( for preliminary results ), blocks and sheets of solid water for doing film work and a Rando phantom for quantitative dose distribution work simulating a real patient set up.

4.1.1 The Water Tank Phantom

The water tank phantom used was an RFA 300 which consists of a cubic perspex tank which sits on a hydraulic platform that can be raised or lowered to the desired position. Inside the tank are a pair of parallel aluminium rails which have two degrees of freedom in the way they move, left-right ( x ) and up-down ( z ). Situated on this pair of rails is a small seat on which a radiation detector ( a diode or a Farmer chamber ) is placed. This seat then moves backwards and forwards on the two aluminium rails providing the third degree of movement in the ( y ) direction.

The tank is set up by wheeling it into position under the treatment head of the accelerator. The tank is then filled with distilled water up to a certain level, it is then raised up to the desired height using the hydraulic lifting device. The tank is raised to a height where the positioning lasers described in chapter one just skim the surface of the water, in this way the radiation probe can be positioned at the
isocentre (at an SSD of 100 cm) when it is at the water surface. In this way the probe can gather data whilst it is moving straight down to produce a dose curve showing how the dose changes with depth (a depth dose curve) or it can be placed at a specific depth and scan in the lateral direction to produce a profile scan.

4.1.2 The Solid Water Phantom

Solid water consists of a solidified resin-like substance which comes in the form of square blocks and sheets of varying thickness (from 1mm to 4cm) and it possesses a similar electron density to water made by RMI. The solid water phantom is set up by stacking the individual blocks and sheets on top of each other on the treatment couch. Provision is made in one of the thicker blocks of solid water for the insertion of a Farmer chamber and another block has space for a PTW (Markus type) chamber. Other than this TLD or X-ray film can be inserted between the sheets of solid water. The solid water set up with TLD's is shown below.

![fig 4.1 TLD's in solid water](image)
4.1.3 The Rando Phantom

The Rando phantom is a tissue equivalent phantom that is constructed in the shape of a person. It consists of an actual human skeleton which is contained inside a synthetic rubber material which has the same electron density as human tissue. The rubber is cast in the same shape as a human being and the skeleton is enclosed inside it. Inside the chest of the Rando phantom there are two regions where the material used is of a less dense type than the surrounding rubber and these regions of lower electron density simulate the lungs, there is also provision made for air cavities which exist inside the human body such as the nasal passages, the larynx and trachea etc. The phantom is sectioned into a number of axial slices which may be taken apart individually to allow for access to the phantom interior.

Small holes exist within each slice of the Rando phantom, the purpose of these holes is to hold TLD’s for dosimetry experiments. Because these holes extend across the width of the slice the placement of TLD’s in these positions for field matching studies was not deemed accurate or reproducible. Instead sheets of 2mm thick solid water phantom where cut to fit exactly between the slices of the Rando phantom. Three such TLD holder were manufactured . One to be positioned at the centre of the supraclavicular beam, one to be positioned in the lower chest region along the centre line of the two tangential beams and the third to lie along the line where the three fields would be matched. To hold the Rando’s slices in place a number of perspex pins are placed along two lines in the phantom and holes corresponding to the positions of these pins were drilled into the solid water sheets so they could also be anchored by the pins. The Rando phantom was CT scanned in the same way that a normal patient would be and then a treatment plan was drawn up for the phantom using a GE treatment planning system. The computer plan consisted of a cross sectional view of the Rando phantom, superimposed on which were the isodose patterns as predicted by the computer for all three proposed treatment fields. Following the dose lines on this
distribution a series of 42 recesses were cut into the face of the solid water slice which would be placed at the junction line of all three matching fields, recesses were also cut into the other two TLD holders as well. The TLD's were positioned to give information about depth dose as well as isodose distribution. In this way it was hoped that the maximum amount of experimental information would be obtained. The TLD’s were placed in their respective positions and the solid water slices were then placed within the Rando phantom and the whole phantom was then clamped down with a special clamping mechanism to prevent the individual slices from moving around. Using this phantom setup a number of measurements using various field arrangements were done. The results of these measurements are shown in section 4.7 and their interpretation is contained in the results section of this chapter.

4.2 Detector Comparisons

The actual measured shape that the penumbra takes is dependent on the detector used. This is because any physical object that is put into the beam will alter the characteristics of the beam. So when any type of detector is used the dose measurement which results is the dose from the beam as effected by the detector and not the actual dose itself. As mentioned in chapter two the types of dose measuring instruments used were Red (shielded against electron backscatter) diode, the grey (unshielded) diode, the Farmer and the RK ionisation chamber and LiF TLD’s. As all these detectors give different responses, profile scans were done with all of them so they could be compared with each other. The first half profile scans shown below in fig 4.2 is of the grey diode and the red diode.
It is shown in fig 4.2 that the two diode profiles are the same basic shape, the only difference is that the grey diode has a slightly broader penumbral region. In figure 4.3 two more dose half profiles are shown, this time the profiles being compared are those of the red diode and RK chamber. Virtually no difference in dose is shown in the tails of the penumbras. At greater depths however they would be a significant difference in the two penumbras due to low energy scattering out of the beam²⁶.
In the top or umbral part of the curve it can be seen that the response of the RK chamber is not as smooth as that of the diode. This effect may just be a statistical phenomenon due to less signal collection brought about by the fact that the RK chamber is air filled and has a much lower electron density than the diode. The second major difference which can be seen between the two half profiles is that the RK chamber profile is wider as one goes down the penumbra. Because of the size of its measuring chamber the Farmer chamber also produces a penumbra which is wider than that of the diode.\textsuperscript{26}
The effective volume of the Farmer is greater than that of the diode and as a result the
dose reading which it gives is lower than that of the diode because the larger volume
can't resolve the steep dose gradient as the smaller diode can.

As is seen in fig 4.4 when a small volume detector is driven at a constant speed through
the water tank it will cross the penumbral edge almost instantaneously, going from a
region of high dose ( inside the penumbra ) to a region of low dose ( outside the
penumbra ). When a detector of large volume crosses the water tank at the same speed
there will be a point ( as seen in fig 4.4 ) when a fraction of the volume will be inside
the penumbra and a fraction of the active volume will be outside the penumbra, when
this happens the dose that is recorded by the detector will be an average of the dose inside the penumbral region and the dose outside this region (which will of course be very much lower). This average will be less than the dose recorded by a small volume detector. The dose that is recorded by a small volume detector will also be an average, however looking at fig 4.4 it can be seen that because of its smaller size the average dose of the small volume detector will be closer to the true value of the dose at the penumbral edge.

Figure 4.5 shows the results of comparing the red (shielded diode with an active area approximately 2mm in diameter), film and TLD (3mm X 3mm in area) data\textsuperscript{26}. The data shows that the TLD's produce a sharper penumbral edge than the diodes and the film, this is due to the higher resolving power of the TLD rods used for this measurement.

The film response and the diode response is seen to be the same. This is because the sensitive part of the diode has the same resolution as the film. The film was scanned using a infra-red scanner with a 2mm diameter detector window.

![Graph showing comparison of red diode, TLD's, and film](image-url)

fig 4.5 Red diode v.s. TLD's and film
4.3 Beam Blocks and Collimators

Before the advent of independent jaws to create asymmetric profiles those portions of the beam that were not required for use in treatment could be blocked off for a treatment by the use of beam blocks. These blocks were positioned under the treatment head of the accelerator, they mounted on a flat, clear perspex sheet of about 0.5 cm in thickness called the block tray. This block tray can be slotted into and out of a metal frame which is attached to the accelerator treatment head. Although large geometrically shaped lead blocks can be situated on the block tray to produce similar effects to those produced by asymmetric collimators. Other types of blocks however are made of a low melting point alloy so they can be easily shaped, these blocks can be used for one patient and then can be remelted and used again for another patient. The purpose of these blocks is to prevent certain regions of the patients body from receiving dose, these regions can include such areas as the lungs or the larynx. The way these blocks work is shown in fig 4.6 below.

fig 4.6 Use of beam blocks
As is seen in fig 4.6 a head/neck treatment is being performed on a patient and a low melting point alloy block has been made and positioned on the block tray so it will cast an X-ray shadow over the patient’s larynx protecting it from the X-rays, the same sort of thing can be done to protect the lungs.

To determine whether these blocks would affect dose differently to the collimators, two half profiles were taken at a depth of 1.5cm in the water tank, the test was carried out with a rectangular, rectangular blocks and the results are shown in fig 4.7. The source to upper collimator distance was 28cm and the source to block distance was 65cm.

**10X10 field match Block v.s Collimator at 1.5cm deep**

Fig 4.7 shows two half profiles produced using the red diode in the water tank at 1.5cm
for both the collimated beam and the blocked beam. The only major difference between the two profiles at this depth is the slightly wider penumbral shoulder of the blocked beam. This corresponds to a slightly higher dose being given when the blocks are used as opposed to the collimators.

4.4 Symmetric Field Matching

Profile scans where done at different depths in the water phantom. The first graph which is shown in fig 4.8 is the result of matching a dose profile of a 10 X 10 cm² field at a depth of 1.5 cm (d_max) with that of its "mirror image". The mirror image is produced by rotating the original half profile 180° about a vertical axis that passes through the 50% dose value of the profile.

Field Match 10X10 Symmetric at 1.5cm depth

![Field Match 10X10 Symmetric at 1.5cm depth](image)

fig 4.8 Symmetric profile match at 1.5cm
Both of these symmetric profiles have been normalised to a value of 100% (100 cGy) at this depth as a standard. The third curve at the top of fig 4.8 shows the total dose administered, which is the sum of the two symmetric profile dose values at each point. As can be seen on the graph the slope of the curve representing the total dose increases linearly from point A to point B on the curve. This increase is due to the relatively flat umbral dose profile being added to the penumbral tail section of the second profile which trails off slowly. This is shown by the fact that this rise in dose of the total is the same as the slope of the profile tail. The drop in dose which can be seen in the middle of the total is due to the actual matching of the penumbral curves. The reason that the dose drop occurs is that in this matching region the dose from the penumbral tail is not added to the normalised umbral dose. The trough which occurs as a result of the drop in dose is seen to have a minimum dose level of 103% (3% above the simple profile but 5% below the maximum of the combined profiles, point B). The next figure (fig 4.9) shows the same beam profiles as fig 4.8 but at a depth in the water phantom of 5cm.
The first feature to note in fig 4.9 is that each symmetric profile only obtains a percentage dose value of 88% (88 cGy) this is due to the attenuation of the radiation beam as a function of depth, a percentage dose of 100% would still occur at a depth of 1.5 cm in the water phantom. Looking at fig 4.9 the total percentage dose curve is seen to rise as one moves from the umbral region to the matching region, this is once again due to the low dose of the penumbral tail being added to the flat umbral region of the opposing profile. In fig 4.9 the two profiles intersect at a value of 70% and, as can be seen from the dose total curve in fig 4.9, doses combine to give a total percentage dose
of 140%. This translates to an actual overdose of 59% when it is normalised to the 88 cGy level of the umbra i.e normalised to the central axis at this depth.

This large overdose of 59% is brought about by a combination of two reasons. The first reason is the one which has already been stated, the dose from the penumbral tail of one profile adds to the umbral dose of the other. However the second and main cause of this overdose is due to the divergence of the two beams and this leads to a substantial increase in the dose level at the matching region.

Figure 4.10 shows the situation for the same X-ray beam at a depth of 10 cm in the water phantom.

Field Match 10X10 Symmetric at 10 cm depth

![Graph showing percentage depth dose for symmetric profile match at 10 cm depth.]

fig 4.10 Symmetric profile match at 10 cm deep
The first thing to note is that the beam attenuation is greater at this depth, this fact is shown by the height of the beam profile. As can be seen from the figure the maximum depth dose of the two beam profiles is 68% as opposed to 100% at a depth of 1.5 cm.

A second feature of these profiles occurs in the low dose penumbral tails, here the percentage dose outside the actual beam increases with increasing depth. The reason for this is that more photons will be scattered out of the beam with increasing depth due to the Compton effect (ref. BJR supplement 17). The final feature which is seen in the symmetric field match at 10 cm deep is the very large overdose (almost double the dose of the actual profile), this high degree of overlap between the two dose profiles at this depth is completely due to the divergence of the symmetric beams which becomes more pronounced with increasing depth. Fig 4.10 shows a maximum over dose of 82% due to the divergence of the symmetric beams when normalised to the 68% maximum central axis dose shown in fig 4.10.

4.5 Asymmetric / Asymmetric Field Matching

Using the water tank a series of asymmetric field profiles were taken, these profile scans were performed at depths of 1.5 cm, 5 cm and 10 cm and each of these profiles was then matched with its own mirror image.
Field Match 10 X 10 Asymmetric at 1.5cm depth

The above graph of fig 4.11 shows the matching of an asymmetric profile of a 6 MV X-rays with its mirror image at dmax (1.5 cm) in the water phantom. (This once again is the depth in water where the transient longitudinal electron equilibrium is established for a 6 MV photon beam.) The dose delivered to the water at this depth is set to be 100 cGy and so the dose values which are obtained here are normalised to 100%. As with the symmetric profile matching there are three curves present in the graph shown in fig 4.11. The first of these curves which is labelled as asymmetric profile 1 in fig 4.11 shows the actual data which was collected when the profile scan was taken. The second
curve on the graph labelled as asymmetric profile 2 is the mirror image of the first and the third curve at the top of the graph represents the sum of the values which occur when the two profiles are matched with each other. By comparing fig 4.11 with to fig 4.8 which shows the matching of two symmetric profiles at this same depth. The total curve is flat for the asymmetric profiles with the excepting of a small 4% over dose at the matching region. The reason why the total curve gives a flat, constant dose for the asymmetric profiles can be seen by studying the profiles themselves. As has already been stated the umbral regions of asymmetric profiles do not give the flat constant dose levels which the symmetric profiles do. Instead there is a definite drop off in dose as one moves from the centre of an asymmetric profile toward the asymmetric edge or penumbral region. This dosage drop which can be seen in the umbral regions of both profiles is matched by a corresponding rise in the dose levels of the penumbral tail regions of both the profiles. When the drop in dose of the umbral region is added to the rise in tail dose the resulting total dose is constant over the whole graph with the only exception being the 4% dose island in the matching region.

Figure 4.12 shows the matching of the same asymmetric profile data with its mirror image, at a depth of 5 cm below the surface of the water tank phantom.
As was seen in the corresponding symmetric profile matching at this depth the maximum dose which is shown by the profiles is 88%, this is due to the attenuation of the X-ray beam with depth. Here again the dose drop which is caused by the asymmetric collimation of the beam adds to the rise in dose of the penumbral tail as one moves from the umbral region toward the matching region. This produces the flat constant dose which is seen in the total dose curve at the top of the graph, once again there is a small over dose in the matching region of the two profiles of 2.6%. In any sort of medical application this sort of over dose would be so small it would not be taken into account. This is in contrast to the results obtained for the symmetric profile.
machine at depth of 5cm.

Figure 4.13 shows the matching of the two asymmetric profiles at a water tank depth of 10cm.

**Field Match 10X10 Asymmetric at 10cm depth**

![Graph showing percentage depth dose against profile (mm) for asymmetric profile match at 10cm depth.]

Because of attenuation of the beam with depth the maximum dose level shown is 68% which is the same as that shown in the symmetric match at this depth (fig 4.10). As with the other asymmetric profiles which occurs as one moves across from the centre of the beam (positioned at x=-60 mm), there is also an underdose of 12% (normalised to the 68% dose value of the central axis) in the middle of the matching region.

4.5.1 Field Mismatching

As explained in chapter 3 when using multiple field treatment techniques in patient
treatment the dose delivered to the line along which the fields are matched is of great importance. Figures 4.14 and 4.15 show the results of mismatching two fields by 3mm.

Figure 4.14 shows the result of overdosing on the matching line by means of a 3mm overlapping of the treatment field and fig 4.15 shows the results of underdosing.

Field Match 10X10 Asymmetric at 1.5cm deep with 3mm overlap

![Graph showing percentage depth dose and profiles.](image)

fig 4.14 Asymmetric profile match with 3mm overlap at 1.5cm deep
The dose profiles for the above figures were taken at $d_{\text{max}}$ for a 6 MV photon beam using the RFA 300 water tank phantom. As seen in fig 4.14 the 3mm mismatch leads to an overdose of 50% being delivered along the matching line and fig 4.15 shows a 48% underdosing along this line.

4.6 Rando Phantom Results
After the results obtained with the asymmetric profile matching in the water tank seemed to confirm that asymmetric collimators did produce more uniform dose distributions within the matching region the second stage of the study commenced.
This consisted of the use of the asymmetric field matching technique on an anthropomorphic phantom that was more clinically realistic where gantry angles could be used unlike the square water tank phantom. The type of phantom used for this second study was a Rando phantom.

The first type of measurement that was performed on the Rando phantom was a 6MV symmetric/symmetric field matching with no angling of the couch or collimators, this would provide the worst possible case as a non clinical benchmark for further studies. A second treatment was again a symmetric/symmetric field match with the couch and collimator position changed to try and obtain better field matching. This second case is of clinical relevance to Illawarra Cancer Care Centre as it is the type of treatment that is used in some hospital situations. The next type of field matching to be studied was that of two symmetric tangential and one asymmetric supraclavicular beams with both the couch and collimators angled to produce the best possible result. This is the type of treatment which is used at the Illawarra Cancer Care Centre as the straight line provided by the asymmetrical supraclavicular beam provides a well defined edge to match to. The final part of the study involved the use of asymmetric collimators for both the supraclavicular and tangential beams. Due to time limitations with the availability of the phantom each of the five different measurements was only done once. The setup of the phantom was performed by a different radiotherapists for each measurement. Evaluation of the TLD's was done by comparing their outputs when read with the outputs of other TLD's exposed to a standard dose. TLD reproducibility was found to typically be about 2%. Similar population standard deviations (ie ± 2% for one standard deviation) were recorded in other reports

The results of the study with the Rando phantom are as follows. The dose distribution which resulted in the matching region when the symmetric tangentials were matched with the symmetric supraclavicular beam and no attempt was made to angle either the couch or the collimators. The target dose that was required was 100 cGy, for an
effective treatment this dose would have to be given over the widest possible area.

The results of matching the two symmetric tangential beams with a symmetric supraclavicular beams while using no collimator or couch rotation is shown in fig 4.16 in section 4.7. This particular set of measurements was done to obtain the worst possible case to use as a benchmark. The target dose that was aimed for was 100 cGy spread over the largest area of the left breast as possible.

As can be seen from fig 4.16 a wide range of doses were obtained, ranging from 170 cGy to less than 70 cGy. The main feature to be noticed in fig 4.16 is that in the region of interest for the treatment area there is a large overdose, this ranges between 155 cGy and 170 cGy, as well as this, the whole treatment region is overdosed.

The cause of this overdose due to the overlapping of the diverging beams within this junction slice, it is seen that this effect can be partially corrected by rotation of the couch and collimators as seen in fig 4.17.

The figure 4.17 shows the same beam arrangement as fig 4.16, the two tangentials and the supraclavicular beams are symmetric, only the treatment couch and the collimator have been rotated. Rotation of the treatment couch and the collimators is a common practice used by radiotherapists to achieve a better beam match in multiple beam treatments, as a result it can also eliminate overdosing. This effect is seen in fig 4.17, again target dose for the region under treatment was 100 cGy which was achieved with an overdose of only 5% (maximum). This improvement in dose is offset however by the fact that the target dose of 100 cGy is distributed over only a small area of the phantom, and the dose falls off substantially as one moves outside this region. Another drawback which can be seen in fig 4.17, is that a dose of between 70 cGy and 88 cGy extends down into the region of the lung which is clinically undesirable.

Field matching using an asymmetric supraclavicular field and symmetric tangential
fields was then studied, the results of this are shown in fig 4.18. This treatment was performed with the aid of moving both the treatment couch and the collimators, this style of treatment is used because the asymmetric edge of the supraclavicular field provides a straight line to match the other field edges to. It is seen in fig 4.18 that the region that is covered by the highest dose has been extended to include more TLD positions than either of the previous two symmetric field treatments, and all the regions of high dose area away from the lung. The target dose was again 100cGy and the region of highest dose ranges from 90cGy to 101cGy, this would indicate a slight underdosing in this area due to the presence of the asymmetric field.

The results of the study performed using asymmetrically collimated tangential fields and an asymmetrically collimated supraclavicular field are shown in fig 4.19. In this case neither the couch or the collimator were moved and the target dose was 100cGy. The highest dose region recorded a dose range of between 80cGy and 90cGy only, this result was disappointing as it meant the whole treatment region was being underdosed. The reason for this underdose may have been due to the umbral dose reduction near the edge of asymmetric fields as mentioned in chapter 1, the typical dropoff at the edge of an asymmetric field is between 5% and 10%, adding this together for both the tangential fields and the supraclavicular field produced the dose deficit of between 10% and 20% seen in the high dose region of fig 4.19. Because of the gantry now being angled the penumbral tail of the opposing beam may not make up the dose short fall due to the lack of scatter.

An attempt was made to correct for this underdosing effect by driving the asymmetric collimators in such a way that they would block off slightly less than half the field. To achieve this the collimators were first driven to the centre of the field in the usual manner and then driven back approximately 1mm. The beam edges produced in this way were then matched and the resulting dose distribution is shown in fig 4.20. The
maximum dose is between 100cGy and 110cGy so the underdosing has been eliminated, also the region covered by this maximum dose has been extended further along the chest than in any of the symmetric/symmetric or symmetric / asymmetric field matching studies. Although the dose distributions extend down into the lung the actual dose to the lung with this method is considerably less than with the symmetric / symmetric field method.

This indicated another possible explanation for the dose deficit. Perhaps a geometric alignment of no better than 1mm. The Rando phantom used for this work was on loan and had to be returned before further investigations could be done. However the ICCC is in the process of acquiring a anthropomorphic phantom and this phenomenon will be investigated further in the future.
Symmetric/Symmetric field with no couch angle

Dose in cGy

Dose > 70
70 > Dose > 80
80 > Dose > 100
120 > Dose > 155
155 > Dose > 170

Chest
Back
Right
Left

fig 4.16
fig 4.17  Symmetric/ Symmetric field with angled couch and collimators
Fig. 4.18 Symmetric/Asymmetric field with angled couch and collimators.
fig 4.19  Asymmetric/Asymmetric field with no couch or collimator angling
fig 4.20  Asymmetric/Asymmetric field with 2mm overlap
4.8 Film

The study that was carried out with film was initiated as a result of the low dose values which were being achieved along the matching line of the asymmetrically collimated beams. The purpose of the film study was:

1) To determine if the collimators were driven precisely to the zero point when the digital readout on the accelerator said they were at zero.

2) To see whether different techniques of driving the collimators could produce improved results.

The type of X-ray film used in the study was Kodak X-Omat V. This film came in individual yellow paper envelopes with dimensions of 27 X 32 cm, each envelope containing an individual sheet of film.

The film in the study was exposed to a 6 MV photon beam, the film was placed on top of a stack of solid water with 1.5 cm of solid water then placed on top of the film so that the film itself was at the $D_{\text{max}}$ position. The film was exposed to 40cGy which is an approximately linear density to dose region25.

Two different techniques were used to drive the collimators to the zero position. The first was to simply drive the collimator directly to the zero position, this was done with the aid of the crosshair marker. The collimator was driven toward the crosshair marker until the field defining light (described in chapter 1) could no longer be seen between the crosshair and the collimator edge and the digital readout read 0.0 cm (ie. drive in).

The second technique employed was to deliberately overdrive the collimators, ie to drive them past the crosshair line and then to drive them back to the crosshair position with the digital readout reading 0.0 cm (ie. drive out) and see if this would produce better
A total of ten films were taken, five using the simple drive to the centre technique and five using the overdrive technique. The films were scanned using the RFA 300 infrared (2mm aperture) film densitometry system.

The results are summarised in Table 4.1

<table>
<thead>
<tr>
<th>Technique</th>
<th>Film No.</th>
<th>Results</th>
</tr>
</thead>
<tbody>
<tr>
<td>Driven to centre</td>
<td>1</td>
<td>3% overdose</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>8% underdose</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>3% overdose</td>
</tr>
<tr>
<td></td>
<td>4</td>
<td>4% underdose</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>indeterminable</td>
</tr>
<tr>
<td>Driven past centre and</td>
<td>6</td>
<td>5% overdose</td>
</tr>
<tr>
<td>Brought Back</td>
<td>7</td>
<td>indeterminable</td>
</tr>
<tr>
<td></td>
<td>8</td>
<td>2% overdose</td>
</tr>
<tr>
<td></td>
<td>9</td>
<td>8% underdose</td>
</tr>
<tr>
<td></td>
<td>10</td>
<td>5% underdose</td>
</tr>
</tbody>
</table>

Table 4.1 Film results

Table 4.1 indicates that the technique which is employed to drive and position the beam collimators is irrelevant. It appears from this small sample that one is just as likely to get an underdose as you are to get an overdose and whether you get an underdose or an overdose is independent of the collimator driving technique employed. The higher underdose percentages result from the asymmetry induced when the collimators are brought to the centre of the beam. This has been introduced in chapter 1.
Chapter 5

Conclusions
5.1 Forming Functions

In order to mathematically model the dose profile a number of different forming functions have been presented. These vary in the accuracy to which they model the penumbral dose. The Cosine forming function for example will only model the edge of the dose profile and is not concerned with the umbral region or with the penumbral tail. The exponential function although it fits the profile better than the Cosine function still requires two distinct regions in order to operate (it is not continuous over the whole range of the profile) and it also has two arbitrary parameters which must be satisfied.

A possible alternative to these functions was suggested in chapter three, the square root function:

\[
Dose = D_0 \left(1 + \frac{(x_0 - |x|)}{\sqrt{(x_0 - x)^2 + n^2}}\right) + D_1 \quad 3.3
\]

Is continuous over the entire profile region and has only one arbitrary parameter \(n^2\) to be satisfied.

Figures 3.8 and 3.9 shows how well the square root function fits the dose profile produced by a single symmetric penumbra compared to the Cosine and the exponential functions.

The failing of this square root function as shown in fig 3.11 is its inability to reproduce the summation curve for the matching of two symmetric profiles as well as the exponential function, although it does model the summation curve better than the Cosine function.

Since the main topic of this thesis was the functions of asymmetric collimators the square root function was modified to model an asymmetric profile. The resulting function is shown as equation 3.4 in chapter 3.
Dose = \left( D_0 - kx \right) \left( 1 + \frac{(x_0 - x)}{\sqrt{(x_0 - x)^2 + n^2}} \right) + D_1

In this function the (-kx) factor accounts for both the decreasing slope of the umbra as one moves towards the collimated part of the beam and it also accounts for the extra dose loss in the penumbral tail.

As shown in fig 3.13 the asymmetric square root function matches the asymmetric profile data better than the exponential function when it, is adjusted to try and model the asymmetric profile data.

The drawback with the asymmetric square root function is that it only applies to the asymmetric half of the profile. This is seen by the lack of absolute value signs seen in fig 3.4. It is these absolute value signs which give equation 3.3 its symmetry around the y (dose) axis.

The asymmetric square root function does however model the summation curve for two matched asymmetric profiles better than both the Cosine function and the exponential function as seen in fig 3.12 and fig 3.13.

The fact that equation 3.4 reduces to equation 3.3 when k is set to zero (and k is a positive number) suggests that k is a function of the collimator position and not some arbitrary parameter.

5.2 Water Tank Results

In doing measurements involving the dose penumbra, the type of detector being used will effect the outcome. For the measurements in chapter four, four different types of detectors were used, the solid state detectors used were the red and grey diodes and the LiF TLD’s, the second type of detectors used were the traditional ion chambers eg. the Farmer chamber and the forth detector type was film.
The results of chapter four show how the penumbral shape is effected by the detector type used. The grey and red diodes have different responses, the grey diode tends to over respond due to its sensitivity to electrons and low energy backscattered X-rays. The RK ionisation chamber shows a less constant dose response in the umbral region than the red diode does, this may be due to the fact that the gas in the chamber has a lower electron density than the solid state detector, so producing a more random response. The lower part of the RK's penumbra seen in fig 4.3 also shows a much wider spread than that of the red diode, this effect is due to the larger active volume of the RK chamber which leads to lower spatial resolution than the smaller sized diode. This widening of the penumbra due to the larger active detection volume of the gas filled chamber also explains why the Farmer chamber's penumbra is also wider than that of the diode.

As seen in figures 4.8 to 4.10 the profiles of two matched symmetric 6MV beams show an overdose with increasing depth of penetration into the phantom. Fig 4.8 shows the two profiles taken at d_{max} positions where the summation curve shows an overdose of 5%.

As the depth is increased the divergence becomes greater, this is seen in fig 4.9 and fig 4.10. The main reason for this overdose is the divergent edges of the beam overlapping at depth.

If this overdosing could be eliminated then the resulting overdosing would also be eliminated, this however is not a trivial problem to overcome with symmetric fields. The data for matching of the asymmetric curves is shown in figures 4.11 to 4.13.

Figure 4.11 shows the two asymmetric profiles at d_{max}. From this figure it is seen that the summation or total curve is flatter than it was in figure 4.8, this is because the downward slope of the asymmetric part of the first profile (described in chapter1) cancels out the upward slope of the penumbral tail on the second profile when the two
are added together. This produces a more uniform dose distribution with only a four percent overdose spike in the matching region.

Figure 4.12 shows the same two beams matched at a depth of 5cm. At this depth the two asymmetric profiles still add together to produce a uniform total dose. The downward slope of profile 1 still cancels out with the upward slope of profile 2's penumbral tail and the 4% overdose seen in fig 4.11 has been reduced to a 2.6% overdose.

Figure 4.13 shows the result of matching the two beams at a depth of 10cm in the water phantom. The total curve is still flat and uniform except for a 12% underdosing in the matching region. The cause of this underdosing can be seen by looking at the point at which the two asymmetric profiles intersect, it occurs at 30% of the maximum dose value in fig 4.13 as opposed to the 50% of the maximum dose value in fig 4.11. This is due to the attenuation at depth.

The asymmetric data taken at depths of 1.5cm and 5cm deep shows that the total dose curve is flat and uniform. This indicates that asymmetrically collimated beams produce a more uniform dose distribution within the matching region.

Figures 4.14 and 4.15 in chapter 4 show the results of mismatched treatment beams with overdoses and underdoses of 50% and 48% respectively. These large dose fluctuations were the result of field mismatching by as little as 3mm. This validates the requirement for extreme care during patient positioning.

5.3 Rando Phantom

The Rando phantom results are also outlined in chapter 4 and figures 4.16 through 4.20. The aim of using the Rando phantom was to produce a target dose value of 100 cGy in the breast region and to have this target dose spread over the widest possible region within the breast area. The first measurement to be done on the Rando was a
bench mark measurement. It was done without the use of collimator or couch rotation and with the use of symmetric field matching, this measurement was taken simply to show the worst possible case for comparison. The second field matching measurement to be carried out again used symmetric fields, this time however the couch and collimators were angled to eliminate the divergence of the beams and so provided a better matching of the profiles. This method also produced a higher dose in the lungs of between 70 cGy and 88 cGy.

A measurement was then made which involved the use of symmetric tangent fields and an asymmetric supraclavicular field. This type of treatment is used in clinical situations because the straight edge of the asymmetric field provides a good base line to match to and symmetric tangent fields are sometimes easier to set up. This treatment method also extended the target dose region to include more TLD spaces than either of the two previous methods. The dose recorded in the target region for this method was between 90 and 101 cGy.

The final set of measurements to be performed on the Rando phantom were made using both asymmetrical tangential and asymmetrical supraclavicular fields. The desired target dose was again 100 cGy however the highest dose recorded in the matching region was between 80 and 90 cGy which meant that the matching region was being underdosed. A possible reason for this underdosing was the dose drop off in the asymmetric edge of the beam that was mentioned in chapter one where the typical dose drop off at the edge of the asymmetric field was found to be between 5% and 10%.

Adding these two dose deficits together for the three fields involved produced the 10% to 20% underdose seen in fig 4.19.

This underdosing problem was overcome by driving the collimators to a position 1mm short of the field centre. The dose distribution from this field matching was found to be between 100 cGy and 110 cGy in the region of maximum dose and the underdosing had been eliminated.
5.4 Film

The results of the film study into different ways of driving the collimators to produce a more uniform dose in the matching region showed that both the method of simply driving the collimator to the centre and the method of overdriving the collimator produced both underdosing and overdosing, which appears to point to a geometric tolerance of about $\pm 1\text{mm}$ at isocentre.

5.5 Future Work

The candidate is now working as a physicist at the Woden Hospital in Canberra where a 2300CD has recently been commissioned. On this machine the lower jaws move asymmetrically as well as the upper jaws. It is hoped that further field matching and collaboration will continue at Canberra and ICCC with the head/neck region being the next site for detailed study for this application.
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