Total Skin Electron Therapy

Using

Beam Modifiers

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BSc. (Al-Yarmouk University)
Post Graduate Diploma in Medical Physics (Jordanian University)

Thesis submitted in requirement for the degree of
Master of Applied Science

At the
School of Applied Sciences,
Royal Melbourne Institute of Technology

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March 2006
Acknowledgements:

I would like to take this opportunity to thank my academic supervisor Dr. Peter Johnston at RMIT for his time, guidance and support throughout this project. Also I would like to thank my field supervisor Mr. Anthony Beal the Chief of Medical Physics Department at Tawam Hospital for his guidance and help.

Many thanks to my friend Sam for his help, and the explanations he gave for some complex materials in the project.

Statement:

The research described in this thesis was performed while I was enrolled as a postgraduate student at the Royal Melbourne Institute of Technology (RMIT). The material presented in this thesis has not been previously published or written by any other person except where due acknowledgment is made in the thesis itself.

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Abstract

The short range of low energy electrons from 2 to 9 MeV has made them useful for the treatment of superficial lesions covering large areas of the body, such as mycosis fungoides and other cutaneous lymphomas. At these electron energies, the beam penetration falls off rapidly beyond a shallow depth. Thus superficial lesions can be treated up to few millimeters without exceeding the tolerance of the bone marrow.

The purpose of this project was to study the effect of the beam modifiers on the characteristics of the Varian 2100C 6 MeV beam using high dose rate total skin electron mode (HDTSe). The technique developed in the study was a modified Stanford Technique. In this technique, the patient is treated with dual six fields using $\pm 17.5^\circ$ angle above and below the horizontal line at 350 cm SSD. The patient is rotated every 60° intervals so that the whole skin surface is covered with the beam. The scattering filter used in the study was two strips of non-exposed developed radiographic films. The filter was mounted on the HDTSe applicator.

The dose uniformity within a rectangle of 160 cm x 60 cm was found to be $\pm 3\%$ along the vertical direction and $\pm 4\%$ along the horizontal direction which meets the beam requirements recommended by the AAPM report (23) [61].

The use of the scattering filter has improved the dose uniformity, but it increased the x-ray contamination beyond the 30 mm depth to about 1.5% which makes the technique unsuitable for TSET.
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Chapter 1

Introduction

1.1 Basic Principles of Radiotherapy:

After the discovery of the x-ray by Roentgen in 1895, the usefulness of ionizing radiation as a means of cancer treatment was soon appreciated. Since that time, radiation therapy has improved and developed into an important specialized medical field. Radiotherapy is used in the treatment of disease, primarily malignant tumours, using electromagnetic and particle radiations. The patient can be treated with radiation from outside of the body, which is called external beam radiotherapy, or by inserting radioactive sources into the body cavities, and this type of treatment is called brachytherapy.

The doctor usually prescribes the treatment to be either radical or palliative treatment. Radical radiotherapy is treatment given with curative intent sometimes using a combination of modalities to cure the patient. In this type of treatment, the side effects sometimes are unavoidable, but they are accepted as an inevitable part of cure.

Palliative radiotherapy is a treatment given for advanced cancer. When the cancer is already spread, this type of treatment will not cure it. The aim of this type is to slow down the growth of the cancer and relieve symptoms such as pain in the pelvis or rectum.

The equipment of choice today for external beam therapy is the linear accelerator (linac). Figure 1.1 shows a schematic diagram of the major components of the linear accelerator. In figure 1.2 the gantry of the linear accelerator rotates about a horizontal on which the isocenter is located. The isocenter is a reference point in space that is common to the axes of rotation for the gantry, collimator, and the treatment table.
Figure 1.1 A schematic diagram of the major components of the linear accelerator [81].

Figure 1.2 The isocentric rotation of the gantry. (Graph adapted from [25]).
The treatment table rotates and moves vertically, longitudinally, and laterally, which helps to position the patient correctly for treatment.

When photons are required for the treatment, photon mode is selected, in which electrons strike a target and produce bremsstrahlung photons. The photons produced are collimated with a primary collimator, and then they are transmitted through a flattening filter to produce a uniform intense radiation field. The beam is then collimated by two pairs of movable jaws to produce a pyramid of useful radiation, (see figure 1.3). The collimator consists of jaws which are used to define the size of the field required in the treatment. The collimator is part of the components of the treatment head of the linac. The radiation field can be selected to be any square or rectangular size from 0x0 up to 40x40 cm\(^2\) at 100 cm from the focus point in the linear accelerator head.

**Figure 1.3** Diagram showing the shape of the x-ray beam after primary and secondary collimation. (Diagram adapted from [25]).
Chapter 1

More details about the major components of the linear accelerator and how the electron beam is produced will be discussed in chapter 2.

1.2 Objective and Content of this Thesis:

The aim of this MSc project is:

- To compare the dosimetry characteristics of matching dual field electron beams for total skin electron therapy (TSET) with and without using beam modifiers within the framework of Stanford technique.
- To check the effect of different scattering materials on the uniformity of the 6 MeV electron beam.
- To try to improve the dose uniformity in the treatment of Cutaneous T-cell lymphoma when the six dual fields technique is used.

The thesis is divided into six chapters. Chapter 2 gives an introduction about the cutaneous T-cell lymphoma such as, causes, symptoms, stages of the disease. It discusses the history and the different techniques used in the treatment of cutaneous T-cell lymphoma.

A description of the history of treatment of the T-cell lymphoma is presented with a discussion of the techniques used by different groups.

A review of the interaction processes of high-energy electron beams with matter is introduced also. These processes are very important in the understanding of the mechanism of energy deposition in media irradiated by electron beams.

Chapter 3 presents a description of all the dosimetry equipment used in the project, such as, ionization chambers, electrometers, phantoms, films, and TLDs. Also a description of the major components of the linear accelerator and the function of each component is presented.
Chapter 1

Chapter 4 presents all the measurements carried out for the 6 MeV electron beam without using beam modifiers (scattering filters). A study of the beam parameters for different setups is carried out for single 6 MeV electron beam at standard source to surface distance of 100 cm SSD, single field at 350 cm SSD, dual stationary field at 350 cm SSD, and six dual fields at 350 cm SSD. Measurements are carried out to study the effect of shielding on the ionization chamber cable. Percentage depth dose measurements are carried out along the central horizontal line and at off axis distances along the vertical direction to investigate the effect of source to surface distance on the electron beam penetration and energy degradation. Beam flatness profiles along the horizontal and vertical direction are checked.

In chapter 5 electron beam modifiers are investigated. Two types of scattering filters are to be used; Perspex and x-ray films without the silver component. The measurements carried out in this chapter are a repeat of the measurements mentioned in chapter 4 but with the use of scattering filters. Surface dose measurements using six dual fields are carried out. Finally, a calibration measurement carried out for the high dose rate 6 MeV electron beam with scattering filter is described in this chapter.

In chapter 6 the results of the beam parameter measurements with and without the scattering filters are discussed. A comparison of both methods is presented followed by the conclusion of the project.
2.1 Introduction:

Total skin electron beam therapy is a well established procedure used in the management of cutaneous T-cell lymphoma [31][76]. One of the most common types of cutaneous T-cell lymphoma is mycosis fungoides. The goal of total skin electron therapy (TSET) is to achieve a cure or control of the disease by delivering a uniform dose to all of the skin both around the circumference of the patient and from head to foot with uniformity of penetration. The depth of radiation penetration is selected on the basis of the stage of the disease, while sparing all other organs from any significant radiation dose. Additionally, it is important to ensure that the bremsstrahlung contamination (hereafter described as x-ray), produced by the inevitable interactions of electrons with materials in the beam path, is acceptably low to prevent serious radiotoxicity arising from the whole body x-ray exposure [6][9][82].

2.2 Irradiation beam requirements in TSET:

The beam requirements in the treatment of cutaneous T-cell lymphoma involve the characteristics of the electron beam such as, field size, penetration, energy, dose, dose rate, beam uniformity in the treatment plane, x-ray contamination (bremsstrahlung), and also boost fields might be required after the treatment.

In the six dual field technique; which will be discussed later in section 2.3.5, the patient is required to be in standing position. The entire body surface to a limited depth close to the surface is to be treated using electron energies less than 10 MeV with low bremsstrahlung. The recommended field size of the composite electron beam at the patient treatment plane is approximately 200 cm in height and 80 cm in width. It is recommended that the beam...
uniformity be $\pm 8\%$ in the vertical direction and $\pm 4\%$ in the horizontal direction over the central 160 cm x 60 cm area of the treatment plane [67]. The measured dose uniformity using phantoms in treatment plane can not be reproduced over the patient. It was found in some measurements that a $\pm 7.5\%$ variation in the treatment plane may increase to $\pm 15\%$ at the patient due to the body curvature of the patient, change in the skin distance, patient setup and motion during the treatment, and self shielding. It was also found that in the perineal region the dose falls to 30-40% of the prescribed dose [54][55].

The prescribed depth will depend on the stage of the disease. The penetration depth range required is from approximately 5mm to more than 15mm at the 50% isodose surface, which includes most lesions.

Self shielding is an important matter, there are body areas shielded partly by some other body parts or sometimes inadequately exposed because of the treatment setup limitations. To solve this problem, small boost fields of electrons or orthovoltage x-rays are sometimes given to the patient.

The x-ray contamination should be as low as reasonably achievable. It can be reduced by angling the beam axes so that the maximum bremsstrahlung is directed outside the body volume above the head or below the feet. The desired x-ray contamination level averaged over the body volume for all fields, ranges from 1 to 4% of the total given electron dose at dose maximum [67]. A 4% x-ray dose (~1.5 Gy) averaged over the body is acceptable by the clinicians at Tawam Hospital.
Chapter 2

2.3 Irradiation Techniques:

2.3.1 Irradiation using X-rays:

The first use of ionizing radiation in the treatment of mycosis fungoides (MF) was in 1902 by Scholtz [70]. Treatments were done mostly with low energy x-rays. It was difficult to give a good treatment to the entire skin area, because of the limitations of the beam geometry, such as, the large field size and field junction limitations [58]. In 1939 Sommerville [73] suggested the use of an x-ray bath for the treatment, but this type of treatment was limited because of the side effects such as bone marrow suppression.

2.3.2 Irradiation using electrons:

Local disease can be treated with low-energy x-rays or electrons. Electrons have an advantage over x-rays because of the characteristic of the depth dose distribution of the electron beam for energies less than 10 MeV [50], where rapid reduction of the dose near the end of the range of the electrons occurs, thus superficial lesions up to depth about 1 cm can be treated with electrons while sparing the other organs such as bone marrow from any significant radiation dose. By using electron beams, the dose maximum occurs just below a normal incident skin surface and a rapid fall-off of dose with depth occurs to a maximum range determined by the incident electron energy (see figure 2.1). Table 2.1 shows the depth dose characteristics for different radiation beams.

It is important to take steps when using an electron beam to minimize the bremsstrahlung contribution in the treatment field and make sure that the radiation beams are monitored properly so that the treated patient is not over-exposed [77]. With measurements done at a depth of 10 cm, it was found that the cumulative x-ray component averaged over the patient volume for all used electron fields ranges from 1-4% of the maximum electron dose received.
at the surface or near the surface [67]. Some clinicians consider a 4% x-ray dose (~1.5Gy) clinically unsatisfactory [57].

The use of the total skin electron beam therapy was first described in 1953 by Trump et al. [79], and it was carried out with a Van de Graaff accelerator. It was followed by a large number of reports to discuss its importance, merits and efficacy. Different techniques of high dose rate external beam therapy using linear accelerators were introduced, developed and published. [6][9][11][84].

<table>
<thead>
<tr>
<th>Treatment</th>
<th>Depth Dose (%)</th>
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<th>2.0 cm</th>
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<td>18</td>
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<td>58.2</td>
<td>38</td>
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<tr>
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<td>100</td>
<td>93.4</td>
<td>86.3</td>
<td>71.9</td>
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<td>6 MeV (with 1 cm bolus)</td>
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<td>76.6</td>
<td>4.0</td>
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</tr>
<tr>
<td>9 MeV (with 1 cm bolus)</td>
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<td>97.4</td>
<td>100</td>
<td>79.8</td>
<td>18.7</td>
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</table>

Table 2.1 Depth dose characteristics for various radiation beams (10x10 cm field). Data adapted from [33].

Many techniques were used with van de Graaff accelerators in fixed positions with treatment beams in a vertical direction. Electron energies used were in the range of 1.5 to 4.5 MeV, with patients translated horizontally on a motor driven couch under the radiation beams. The translation treatment technique was described by Williams et al [83].

At the National Institutes of Health (NIH) in the US, the Van de Graaff TSET technique was modified by using a wide cone. The beam scanned in the x-direction (transverse), while the
table moves in the y-direction (longitudinal) [67]. The dose distribution in the treatment plane depended on the distance below the cone. The uniformity was about $\pm 5\%$.

Figure 2.1 Depth dose curves in water measured at Tawam Hospital for 6 MeV electron beam with SSD at 1m and at 4.5m. Depth of maximum dose at 1 m is 15 mm and at 4.5 m is 8 mm. Both curves are measured at in water using the RFA water phantom and beam analyzer.

Another Van de Graaff accelerator TSET technique was used at the Massachusetts Institute of Technology [47][60]. In this technique, the treatment couch is moving while the accelerator is in fixed position. A beam of electrons is directed through Al foils placed near the vacuum window of the accelerator drift tube. The Al foils are used to scatter the electrons passing though them. The scattered electrons are then directed into a conical collimator with a slit 1 cm x 45 cm at its base close to the surface of the patient. The slit is positioned at right angles to the direction of the couch movement and perpendicular to the beam axis. The dose uniformity to the patient varies as much as $\pm 15\%$ depending on the variation of the distance between the base of the cone and the patient skin during the treatment. The design of the cone
Chapter 2

was modified to improve the uniformity, reduce the energy loss in the scattering foils, increase the effective dose rate at the patient surface, and reduce the x-ray contamination. By this method, the dose uniformity across the field was improved and found to be $\pm 3\%$ and the dose variation because of the change in the distance between the cone and the skin of the patient was found to be $\pm 8\%$. Patients were treated at the beginning in four positions. It was found that some telangiectasias were developed in high dose regions. Telangiectasia are small enlarged blood vessels near the surface of the skin, usually they measure only a few millimeters. They can develop anywhere on the body but commonly on the face around the nose, cheeks and chin. Six and eight fields positions treatments were developed. Internal eye shields were used when the eyelids were involved in the treatment, while external eye shields were used when the eyelids were not involved in the treatment. Some areas required a boost treatment after finishing the multi-fields treatment due to the low doses received in these areas.

Tetenes and Goodwin described the scattered single electron beam technique [77]. 6.5 MeV initial accelerator energy was used with a titanium scattering foil 0.15 mm thick placed at 10 cm from the exit window of the accelerator. A beam flattening filter was inserted in the front of the treatment head. The distance between the exit window of the accelerator and the treatment plane was 7 m. The measured uniformity was from $\pm 1\%$ to $\pm 8\%$ (40 cm to 100 cm radius around the central axis). The electron energy at the treatment plane was about 4 MeV. Unfortunately, this type of treatment technique requires a long treatment distance. This type of treatment is limited to the size of the available treatment room. However, Szur et al [75] developed a technique to reduce the treatment distance by using two horizontal parallel beams whose axes are contained in a vertical plane at a treatment distance at about 2m. 8 MeV electron energy was used; Carbon energy degraders were used to adjust the depth of penetration depending on the treatment depth prescribed by the clinician.
Chapter 2

2.3.3 Irradiation using beta particles:

Beta particles from radioactive sources are an alternate source of electrons for TSET. For example, strontium-yttrium 90 has been used which has a maximum energy of 2.18 MeV, an average energy of 1.12 MeV, and typically 10% depth-dose in the range of 0.4 to 0.8 g/cm² [27] [28][29].

Electrons generated from linear accelerators are preferred in the treatment of TSET rather than the beta rays from radioactive sources such as strontium-yttrium 90 due to shorter treatment times; it takes more than 15 minutes to deliver 2 Gy using the largest beta-particle source. The limited penetration of the beta rays because of their lower energies is another reason and poorer uniformity has been achieved using beta particle treatments compared to linac treatments has been reported [67].

2.3.4 Irradiation using pendulum-arc:

The Pendulum-arc technique was developed by Sewchand et al [69]. It consists of six arcing fields symmetrically dispersed around the body surface for circumferential coverage. An isocentrically mounted linear accelerator of 8 MeV was used. The accelerator is rotated continuously during the treatment in a 50° arc about the isocenter. The angle of the arc was selected to scan the full body height starting from an initial angle with the axis of the treatment beam aimed below the feet of the patient and ending with the axis of the beam aimed above the head of the patient. Beam uniformity within 10% over a height of 180 cm was achieved at a treatment distance of 385 cm. A large Plexiglas sheet of 1 cm thickness placed at a distance of 5 cm from the patient’s skin surface is used to reduce the beam energy and to provide large angle electron scattering near the patient skin. The surface dose uniformity for the degraded beam was measured to be within ± 7% over most of the body
surface. The x-ray contamination was measured at 10 cm depth to be 4.2% of the average electron dose at depth of maximum $D_{\text{max}}$.

2.3.5 Irradiation using dual angled fields:

This is the most common technique used in the treatment of mycosis fungoides (see figure 2.2). This type of treatment was developed in 1960 at Stanford University by Karzmark et al [48]. In the treatment technique a 4.8 MeV linear accelerator was used, with four angled electron beam pairs. The energy of the beam at the treatment plane was about 2.5 MeV. This technique was developed because a single field in the horizontal direction does not give sufficient dose uniformity in the vertical direction when the patient is in the standing position during treatment. However, the dual-field technique with angles from $15^\circ$ to $20^\circ$ above and below the horizontal line can give sufficient dose uniformity. Two to eight pairs of angled electron beams are usually used to obtain large fields for total skin irradiation. The axis of the beam is directed below the patient's feet for half of the treatment and above the patient's head for the remainder.

The current Stanford technique described in detail by Karzmark in 1970 [66]. The dual six fields used in the technique provide a dose rate of 0.25 Gy/min at the treatment plane 3 m from the ion chamber-scatterer located on the front of the treatment head. The region where the effect of the two fields overlaps can be a substantial problem because it will affect the uniformity of the dose distribution during the treatment. Due to this reason the angles between the two fields are selected carefully to give the best uniformity. When $17.5^\circ$ angle above and below the horizontal is used at 3.5 m, the gap between the upper and lower light fields projection is 28 cm. The projected electron beam at this distance doesn't coincide exactly with the light field. It is expected to be less than 28 cm; this is due to several reasons. First, the electrons are generated from a broad virtual source located at 10 cm or more downstream from the position of the target, while the light field is generated from a point source at the
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target. Second, the relatively of the broad source of electrons is closer to the linac jaws than the treatment plane. This would increase the penumbra which is also broadened by scattering through the air. Third, the electrons in the beam will have undergone some scattering, with those in the gap merely having scattered through larger angles than those in the geometric beam.

Many of the treatment techniques incorporated a large clear Lucite energy degrader panel, 1 cm thick and about 200 cm x 100 cm in cross section [10][30][56]. The degrader is placed approximately 20 cm in front of the patient to improve the dose uniformity and to compensate for the oblique body surfaces. However, the usage of the degrader reduces the radiation penetration (energy degradation), which causes the depth dose to fall off to a shallower depth. The degrader can also be used to mount the ionization chamber for dosimetry purposes.

![Figure 2.2](image)

**Figure 2.2** The geometry for TSET using dual angled field. Two beams directed at 17.5° above and below the horizontal axis provide a uniform dose over large area.
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Degraders can be placed either (i) near the accelerator exit window or (ii) at or near the patient surface. Brahme investigated the difference between these two extreme cases [9]. When the degrader is placed near the accelerator exit window, the electrons reaching the patient will have a narrower angular spread than when the degrader is placed near or at the surface of the patient. The narrower angular spread gives less surface dose and a deeper depth dose because of the resulting increase in practical range ($R_p$).

2.4 Cutaneous T-Cell Lymphoma (CTCL):

2.4.1 Introduction:

Cutaneous T-cell lymphoma is a type of non-Hodgkin's lymphoma: a cancer of the lymphatic system. The lymphatic system is part of the immune system and helps in fighting infection. It contains cells known as lymphocytes. Lymphocytes are a type of white blood cells and are an essential part of the body's defense against infection [36].

There are two main types of lymphocytes: B-cells and T-cells. Both B-cells and T-cells help to fight the infection (see figure 2.3).

There are more than 20 types of non-Hodgkin's lymphoma. Cutaneous T-cell lymphoma (CTCL) is a rare disease, which represents up to about 5% of all cases. It is a cancer of the T-lymphocytes and usually occurs in people with ages ranging from 40 to 60 years. CTCL affects mainly the skin of the patient (see figure 2.4). This is caused by the uncontrolled growth of a type of white blood cell in the skin (T-cells).
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Figure 2.3 A diagram showing the main groups of lymph nodes in the body. (Diagram adapted from [36]).

Figure 2.4 Typical plaques of mycosis fungoides
Chapter 2

2.4.2 Causes and symptoms:

The cause of CTCL is unknown. Exposure to chemicals has been suggested; however, the most recent study on the subject failed to show a connection between exposure and development of the disease [37]. In common with other types of cancer, CTCL is not infectious, i.e. the disease can not be transferred from the patient to other people.

The early stage symptoms of CTCL can be difficult to diagnose. The symptoms are seen primarily in the skin, with itchy red patches or plaques and, usually over time, mushroom-shaped skin tumors. The distribution of the rash or tumours on the skin varies from one patient to another. The only really universal symptom of the disease is the itch which is usually what brings the patient to seek medical help. When the disease spreads outside the skin, then the symptoms include swelling of the lymph nodes, usually most severe in those draining the areas with the skin involvement. Spreading the disease to the viscera can cause disorders of the lungs, upper digestive tract, liver, or central nervous system.

In some cases, the cutaneous T-cell lymphoma is developed to a leukemic phase called Sezary syndrome, in which malignant T cells appear in the bloodstream. It is named for the French dermatologist who first identified the abnormal T cells.

2.4.3 Diagnosis of CTCL:

The diagnosis is done by taking a biopsy from the patient. This is done by taking a small sample of the skin from the affected area, and then this sample is analyzed and examined in the laboratory.

2.4.4 Stages of the disease:

Three stages are used to determine how the disease is affecting the skin of the patient. The disease may not progress through all three stages. The stages are:
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- **Pre-tumor stage:** Small, raised red patches on the skin, usually appears on the breast and buttocks, however, they can appear anywhere on the body.

- **Plaque or infiltrative stage:** plaques or irregular shaped red patches are formed. Usually, it affects the buttocks, skin folds and the face, but it can also spread to the other body parts.

- **Tumor stage:** A small percentage of people reach this stage. The lymph nodes might be affected at this stage, but only rarely are the major internal organs, e.g. the liver, spleen and lungs affected (see figure 2.5).

- **Sezary syndrome:** This occurs when areas of the skin are affected and the affected lymphocytes are found in large numbers in the blood.

2.4.5 **Grading of CTCL:**

Non-Hodgkin's lymphomas are divided in to two groups: Low-grade (slow growing lymphomas) and high-grade (fast growing lymphomas). CTCL is a low grade lymphoma, in which the development occurs very slow by and takes many years to move from one stage to another. Most people never progress beyond the first stage.

![Figure 2.5 Tumor stage of mycosis fungoides (MF). Photograph adapted from [39].](image)
2.4.6 Treatment of CTCL:

Treatment of CTCL depends on the stage of the disease. Early stage of the disease may be treated with chemotherapy applied to the skin and ultraviolet A light exposure (PUVA). Treatment with multiple agent chemotherapy may also be used. Treatment using total skin electron beam therapy (TSEBT) may be used when the disease has spread [38].

2.5 Radiobiology:

Radiobiology is the study of the effects of radiation on biological molecules, cells, tissues, and whole organisms. Radiobiology seeks to discover the molecular changes responsible for radiation effects such as cancer induction, genetic changes, and cell death. It is very important to understand the interactions between the charged particles and living matter, because they are directly responsible for the damage we observe. As the electrons pass through a cell the energy is released along their tracks. The released energy causes different kinds of chemical processes which may damage some molecules of biological importance, such as DNA strings, causing cell death [3].

Many studies have been carried out to evaluate the dose response relationships in the treatment of mycosis fungoides. A direct correlation between the dose and the disease free interval was shown by Kim and associates [51]. Low single doses (~1.0 Gy) caused a partial regression of the disease, while complete response required a dose of 7.0 Gy or higher. Split-dose studies showed no difference in tumor regrowth at treatment intervals of one or 7 days, suggesting little recovery between treatment fractions and enabling the use of the protracted fractionation schemes. As a result of that, more limited effects of radiation doses on normal tissues were provided and the therapeutic ratio was better.

Another study made by Cotter et al [16] using dose fractionation treatment ranging from 10 Gy to 40 Gy for individual plaques and tumors verified a similar response rate. On the other
hand, the probability of local recurrence (return of the disease) after the treatment had an inverse relation with the dose, and was 42% for doses less than 10 Gy, 32% for doses of 10-20 Gy, 21% for doses of 20-30 Gy, and no recurrences for doses greater than 30 Gy.

Stanford data for total skin electron therapy (TSET) [30] identified a dose-control relationship which showed an increased rate of complete response with increasing dose. The complete response rate for doses less than 10 Gy was 18%, 55% for doses of 10-20 Gy, 65% for doses of 20-30 Gy, and 3% for doses of 30 Gy or greater. The probability of survival for patients treated with doses of 25 Gy or more was greater than for those treated with lower doses (less than 25 Gy).

2.6 Total skin electron therapy at Tawam Hospital:

Total skin electron therapy at Tawam hospital is based on the dual field technique described by Hoppe et al [31], which permits the treatment of TSET in a treatment room (12.5 m x 11 m). In this method, the high dose rate total skin electron (HDTSe) mode, which is a special procedure mode of the Varian Clinac 2100C, is used in the treatment to provide a 6 MeV electron beam at high dose rates. The reason for using the HDTSe mode is to reduce the treatment time at extended distances because the patient may not be able to remain standing during a more lengthy treatment and thus minimize patient motion and fatigue.

The patient is placed at extended distance of 350 cm SSD on a base which can be rotated manually. The patient is rotated to assume six different positions at 60° intervals that include anterior, posterior and four opposed oblique fields. In each position, the patient is treated with two beams 17.5° above and below the horizontal. The gantry angles used in the treatment are 252.5° and 287.5° as shown in figure 2.2. This type of treatment produces a relatively uniform dose over the body surfaces however there are several body areas that need boosts or partial
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shielding. Eyes are shielded to reduce the dose to the lens. In limited plaque disease that does not involve the face, external lead eye shields are used. In all other patients, internal eye shields are used. An external bolus may also be used over the eyelids to reduce the depth-dose further. Some areas in the body, such as fingers and toes, are so thin that they receive both entrance and exit dosage and therefore need shielding.

To provide this type of treatment at Tawam Hospital a special stand was made in the department for the treatment (see figure 2.6). The first case of Mycosis fungoides was treated at Tawam Hospital in year 2000. This type of disease is rarely found between the local populations of the United Arab Emirates. The total number of patients treated since year 2000 is five.

Figure 2.6 A TSET special treatment stand used at Tawam Hospital.
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2.7 Electron Interactions with Matter

2.7.1 Introduction:

In radiotherapy and radiation dosimetry, it is very important to have accurate information about the electron beam penetration in matter. Section 2.7 presents a discussion of the basic processes of the electron interactions with matter. These processes are very important to understand the mechanism of energy deposition of the electron beam in the medium. An electron beam loses energy in a different way to the photon beam. Photons need an intermediate interaction stage to transfer the energy first to the electrons before transferring any energy to the medium, while electrons start losing energy immediately to the medium [2]. The coulomb electric field surrounding each electron causes the immediate interaction between the electron and atoms in the medium.

When the electron interacts with the medium a small fraction of energy of the electron will be transferred to the medium. So, the electron goes through many collisions before coming to thermal equilibrium with its surroundings. The most important types of interactions which may occur between an electron and an atom are shown in Figure 2.7. In all atoms, electrons usually occupy the lowest possible energy states, starting from the K shell. The inner electrons are the most tightly bound, and they require a large energy to remove them. When the electron passes through the medium, it is highly probable that the interaction will be with the least tightly bound electron.
When an electron beam passes through matter, individual electrons may interact with the atom as a whole, with atomic electrons or with the nucleus. The observed interactions include inelastic collisions with orbital electrons causing excitations and ionizations, elastic scattering (Coulomb scattering), Fermi-Eyges small angle scattering and radiative collisions involving the nucleus or orbital electrons. The type of interaction that an electron undergoes with a particular atom of radius \( a \) is determined by the energy of the passing electron and the perpendicular distance between the electron direction before the interaction and the atomic nucleus, which is defined as the impact parameter \( b \) [22][50][52]. (See figure 2.8).

- For \( b \gg a \), the electron passes the atom at a distance and the interaction is classified as a soft collision with the whole atom and only a small amount of energy will be transferred from the incident electron to orbital electrons.
• For $b \leq a$, the interaction is classified as a hard collision with an orbital electron. An appreciable fraction of the electron's kinetic energy will be transferred to the orbital electron.

• For $b < a$, the incident electron undergoes a radiative interaction (collision) with the atomic nucleus. It will cause the electron to emit a photon (bremsstrahlung) with energy between zero and the incident electron kinetic energy. The energy of the emitted bremsstrahlung photon depends on the magnitude of the impact parameter $b$; the smaller the impact parameter, the higher the energy of the bremsstrahlung photon [22][52].

![Figure 2.8 Electron interaction with an atom, where a is the atomic radius and b the impact parameter.](image-url)
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The different types of electron interactions with matter are [50]:

1. Inelastic collisions with atomic electrons (ionization and excitation).
2. Inelastic collisions with nuclei (bremsstrahlung).
3. Elastic collisions with atomic electrons.
4. Elastic collisions with nuclei.

The energy lost by the inelastic collisions is described as:

- Collisional energy losses.
- Radiative energy losses.

These mechanisms of energy deposition are very important in the medical field, because they are responsible for energy deposition locally in the irradiated medium. The radiobiological effects in the irradiated tissue are directly related to the energy expended in medium.

2.7.2 Collisional Stopping Power:

The stopping powers are widely used in radiation dosimetry, but they are rarely measured, and usually they are calculated using stopping power theory. In radiotherapy range of energies, the most important contribution comes from the collisional energy losses due to excitation and ionization of atoms. Stopping powers are also used to determine the absorbed dose in a medium irradiated with high energy radiation beam based on the cavity theory.

The different types of interactions which the electron experiences during its passage through the medium produce a gradual loss of energy causing the electron eventually to thermalize. The energy lost in individual interactions is usually small which means that the electron is losing energy continuously. The average distance rate of the energy loss is known as the linear stopping power. In many applications, the linear stopping power divided by
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medium density $\rho$ to give the total mass stopping power $(S/\rho)_{tot}$ which represents the kinetic energy (KE) loss by the electron per unit path length $x$ divided by the density, or

$$
(S/\rho)_{tot} = \frac{d(KE)}{\rho \, dx}
$$

(in MeV.cm$^2$/g) \hspace{1cm} (2.1)

where $(S/\rho)_{tot}$ consists of two components: the mass collision stopping power $(S/\rho)_{col}$ resulting from inelastic collisions with atomic electrons of the medium (atomic excitations and ionizations) and the mass radiative stopping power $(S/\rho)_{rad}$ due to electron interactions with the electric field of nuclei resulting in the production of bremsstrahlung [3][52].

$$
(S/\rho)_{tot} = (S/\rho)_{col} + (S/\rho)_{rad}
$$

(2.2)

The energy lost in ionization and excitation of atoms is absorbed close to the electron track whereas the energy carried off in the form of bremsstrahlung travels far before being absorbed. This is an important fact when there is a need to distinguish between the energy imparted close to the electron path, and the total energy lost along the whole of the electron track.

The mass collision stopping power can be obtained from tables calculated by Berger and Seltzer (1983) [7] which are based on the theoretical work of Bethe (1933), but with the inclusion of density correction. The following formula is used [3][8][50][52]:

$$
(S/\rho)_{col} = \frac{2\pi r_e^2 m_e c^2 N_A Z}{\beta^2 M_A} \left\{ \ln \left[ \frac{\gamma^2 (\gamma + 2)}{2(1/(m_e c^2))^2} \right] + F(\tau - \delta) \right\}
$$

(2.3)

where
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- $\delta$ is the density effect correction according to Sternheimer and Peierls.
- $\tau = (E/m_e c^2)$, is the ratio of the kinetic energy, $E$, of the electron to its energy.
- $\beta = \nu/c$, $\nu$ is the velocity of the electrons, $c$ is the velocity of light in vacuum.
- $N_A$ is the Avogadro constant ($6.02252 \times 10^{23} \text{ mol}^{-1}$).
- $r_e$ is the electron classical radius ($2.818 \times 10^{-15} \text{ m}$).
- $Z$ is the effective atomic number of the medium.
- $M_A$ is the molar mass of the medium, and
- $I$ is the mean excitation energy of the medium.

The factor $F(\tau)$ is obtained from:

$$F(\tau) = 1 - \beta^2 + \left[ \frac{\tau^2}{8} - \frac{2\tau + 1}{\ln 2} \right] / (\tau + 1)^2$$  \hfill (2.4)

2.7.3 Radiative Stopping Power:

The radiative stopping power is the distance rate of bremsstrahlung production. Bremsstrahlung occurs when an electron is deflected in the electric field of the nucleus and, to lesser extent, in the field of an atomic electron. At high electron energies, the emission of radiation is mostly in the forward direction (the direction of movement of the electron).

The radiative electron interactions cause large energy losses. Electrons going through these types of energy losses contribute mainly to the tail of the primary electron energy distribution.

Energy loss by an electron in radiative collisions was studied quantum mechanically by Bethe and Heitler [6]. When the incident electron passes near the nucleus, the field in which it is accelerated is the bare coulomb field of the nucleus. If the distance between the electron and the nucleus is great, the partial screening of the nuclear charge by the atomic electrons
becomes important, and the field is no longer coulombic. This means that the effect of the atomic-electron screen will change depending on the distance between the incident electron and the nucleus. It was also mentioned before in the introduction section that the energy loss will also depend on the energy of the incident electron.

The radiative stopping power equation is \[7\][8]:

\[
\frac{S}{\rho}_{\text{rad}} = \frac{4r_e^2 \alpha}{\beta^2} N_A \frac{Z (Z + 1)}{M_A} (\tau + 1) m_e c^2 \ln(183 Z^{-1/3} + 1/18) \tag{2.5}
\]

where \(\alpha\) is the fine structure constant \((\alpha \approx 1/137)\).

Equation 2.5 shows the rate of energy loss of the incident electron is proportional to the energy of the electron and to the square of atomic number. It is shown that the energy loss increases with electron kinetic energy and with \(Z\). This means that the production of bremsstrahlung photons increases by using higher energy electrons and higher atomic number \((Z)\) absorbing materials [3][22][80].

2.7.4 Radiation Yield:

Radiation Yield, \(Y\), is defined as the fraction of the total initial energy of the electron that is emitted as electromagnetic radiation (bremsstrahlung) when the incident electron slows down and comes to rest. The average radiation yield for an electron of an initial kinetic energy \(T\) in MeV when coming to rest in an absorbing material of atomic number \(Z\) is given approximately by the following equation [7]:

\[
Y = \frac{6 \times 10^{-4} ZT}{1 + 6 \times 10^{-4} ZT} \tag{2.6}
\]
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An estimate of radiation yield can give an indication of the potential of bremsstrahlung hazard of the electron source. To keep the bremsstrahlung to the minimum, use a low-Z material as a decelerator to slow the electrons since \( Y \) increases with \( Z \).

2.7.5 Range:

The range of charged particles can be defined as the distance it travels before coming to rest. This range \( R(T) \) is given as a function of kinetic energy \( (T) \), can be calculated using the following equation:

\[
R(T) = \int_{0}^{T} \left( \frac{dE}{dx} \right)^{-1} dE \tag{2.7}
\]

It is assumed in this definition that the kinetic energy of the particle changes as a continuous function as it slows down to rest. This definition is physically appropriate for the heavy charged particles, but it is not always realistic with electrons, because electrons can lose a large amount of energy in single collisions. The data of the range shown in table 2.2 are calculated using this definition. Table 2.2 gives the collisional, radiative, and total mass stopping power as well as the radiation yield and the range for electrons in water.

There are two factors that complicate the estimation of the electron range. First, if the radiation yield \( (Y) \) is high, then the range of the energy associated with the incident electron is large because of the very large mean free path of the produced photon (bremsstrahlung). Second, since the electron does not have a straight path, it is important to recognize the difference between the distance travelled and the penetration distance.

The range of electrons \( (R) \) is related to the kinetic energy \( (T) \) in low-Z materials by the following empirical relations [80]:

For \( 0.01 \leq T \leq 2.5 \text{ MeV}, \)
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\[ R = 0.412 \cdot T^{1.27} - 0.0954 \cdot \ln T \]  
\[ \text{(2.8)} \]

or

\[ \ln T = 6.63 - 3.24 \cdot (3.29 - \ln R)^{1/2} \]  
\[ \text{(2.9)} \]

For \( T > 2.5 \) MeV,

\[ R = 0.530T - 0.106 \]  
\[ \text{(2.10)} \]

or

\[ T = 1.89 \cdot R + 0.200 \]  
\[ \text{(2.11)} \]

2.7.6 Electron Scattering:

When an electron beam passes through an absorbing medium, the surrounding coulomb force of the electrons interacts with the electrons and nuclei of the atoms of the medium the beam passes through. These interactions cause energy losses and electron scattering, in which the direction of the electron path is changed.

Scattering of electrons is divided into two types:

1. Interactions of incident electron with an orbital electron.

2. Interactions of the incident electron with the Coulomb field of the nucleus.
<table>
<thead>
<tr>
<th>Kinetic Energy</th>
<th>Collisional (MeV cm(^2) g(^{-1}))</th>
<th>Radiative (MeV cm(^2) g(^{-1}))</th>
<th>Total (MeV cm(^2) g(^{-1}))</th>
<th>Radiation Yield</th>
<th>Range (g cm(^{-2}))</th>
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</table>

**Table 2.2** Electron Collisional, Radiative, and Total Mass Stopping Power; Radiation Yield; and Range in water [80].

The electrons are deflected continuously when they interact with the nuclei. Small-angle deflections are most frequent, while large-angle deflections are occurring from time to time. In small-angle deflections, electrons undergo a large number of collisions with a random distribution in space. The most popular theory that deals with multiple small-angle scattering is Fermi-Eyges theory. This theory accounts for the lateral movement and change of direction as the electron moves a small fraction of its range. It also includes the effects of electron energy degradation. In radiotherapy, the Fermi-Eyges theory provides a good approximation to the angular and spatial spread of electron beams.
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This theory was developed for cosmic rays, therefore the electron is assumed to have a very high energy that remains constant as the electron progresses through the medium, despite numerous collisions. Then the Fermi distribution function was formulated based on the assumption that electrons achieve only small net deflections, due to multiple small angle scatterings, and that no electron are lost from the beam.

If the electron is traveling parallel to the z axis (depth) through a medium in the z-x plane, \( \theta \) is the projected deflection angle which the electron makes with the z axis and x to be its distance from the z axis. Then the Fermi distribution probability function \( (P) \) [52] is

\[
\frac{\partial P}{\partial Z} = -\theta \frac{\partial P}{\partial X} + \frac{1}{4} K(Z) \frac{\partial^2 P}{\partial \theta^2}
\]  

(2.12)

Where \( K(Z) \) is the parameter related to electron energy.

In 1948, Eyges gave a solution to the problem of the collision energy loss in the Fermi distribution function and derived a modified solution which takes this effect into account. In his approach the electron energy is assumed to change with depth, where the energy degradation rate in unit density material is significant, being about 2 MeV cm\(^{-1}\).

2.8 Summary:

In this chapter the history and the different techniques used in the treatment of the T cell lymphoma investigated by different groups and institutes was discussed. An introduction about the T cell lymphoma, symptoms, causes, diagnosis, stages, and treatment was discussed. Finally the basics of electron interaction with matter were presented.

In the next chapter the description of the dosimetry equipment and machines used in the project will be provided. The design and major components of the linear accelerator will be
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presented. The dosimetry equipment description will involve the electrometers, ionization chambers, films, TLDs, and phantoms.
Chapter 3

Equipment and Dosimetry

3.1 Introduction:

Many different types of detectors have been developed to measure the dose delivered to tissue and tissue equivalent media. In radiotherapy, detectors are normally characterized as absolute or relative. Absolute dosimeters include the calorimeter, the chemical dosimeter, and the standard free ionization chamber. Relative dosimeters include films and solid state dosimeters, such as luminescent dosimeters and diode dosimeters.

In radiation therapy the patients are usually treated by radiation from a linear accelerator. The quantity of radiation prescribed to the patient is called the dose, which is determined by the doctor. Before treating the patient, the doctor needs to know the dose to the target area and to the nearby organs.

It is very important to measure the dose delivered to the patient. The need for accurate dosimetry is of utmost importance in radiation therapy for cancer, because in radiotherapy a large dose of radiation is delivered to the tumor and at the same time the treatment is planned so that the dose to normal tissue is minimized. Thus, one of the major roles of the physicist in radiotherapy is to ensure the accurate delivery of radiation dose to the patient.

In radiation dosimetry the absorbed dose and the dose rate are measured. Absorbed dose is a measure of energy deposited in a medium by ionizing radiation. It is equal to the energy deposited per unit mass of the medium, and so has the unit J/kg, which is
given the special name gray (Gy). It is important to know that absorbed dose may not
by itself be a good indicator of the likely biological effect. For example 1 Gy
delivered by alpha radiation would create more biological damage than 1 Gy
delivered by photon radiation.

3.2 Medical Linear Accelerator:

3.2.1 Principles of Operation:

The linear accelerator Varian 2100C generates therapeutically useful megavoltage x-
ray and electron beams with optional capabilities including total body irradiation with
x-rays or electrons. The linac is a machine in which high power microwaves are used
to accelerate electrons traveling in a straight line to the required output energy. The
major components of the linear accelerator were shown in Figure 1.1. DC power from
the power supply is provided to the modulator in which the pulse forming network is
located. The high voltage DC pulses from the modulator are delivered to the
microwave power source and simultaneously to the electron gun.

3.2.2 Electron gun:

The electron gun is a device used for the production of electrons in linear accelerators.
It is effectively a small single stage electrostatic accelerator where the anode has a
hole in it through which the electrons may pass, and then the electrons are injected
into the bunching section of the waveguide of the linear accelerator.

3.2.3 Production of microwave (Klystron):

There are two devices used in the production of microwave power in linear
accelerators, a magnetron or a klystron. Our focus will be on the klystron because it is
the one used in the Varian 2100C linear accelerator. The klystron is a microwave amplifier which consists of three parts: the electron gun, RF section, and the collector. Electrons are emitted from the heated cathode surface in the electron gun and then accelerated by a potential difference between the cathode and the anode. The cathode is maintained at high negative DC voltage while the accelerator guide is kept at ground potential and serves as anode. A focusing electrode is used to compensate for the divergence of the electron beam because electrons experience electrostatic repulsion. The RF section contains cavities, known as bunchers, separated by drift tubes. Electrons introduced to the cavity are accelerated by the RF waves which create an alternating electric field across the cavity. As a result, some electrons are sped up while others are slowed down forming tight groups of electrons referred to as bunches. The name of this process is velocity modulation. When electron bunches arrive at the collector, they produce charges on the ends of the cavity which cause a retarding electric field. Consequently, the electrons are decelerated producing high power microwaves with the same frequency as the inputted RF. The produced RF is directed to the accelerator waveguide through a circulator in one direction only [65].

3.2.4 Linear accelerator waveguide:

The electrons produced from the electron gun are injected to the heart of the linear accelerator - the waveguide, where the acceleration takes place. There are two basic types of waveguide used in linear accelerators, the traveling waveguide and the standing waveguide.

The traveling wave accelerator uses a unidirectional traveling electromagnetic wave to accelerate the electrons. This wave is injected in to the high vacuum waveguide on the electron gun side and is absorbed by a dummy load at the other end of the
structure, preventing the wave from reflecting back. On the other hand, the standing waveguide, such as the one used in the Clinac 2100C, consists of a series of copper interconnected evacuated cavities. The cavities are designed and calibrated so that the time varying wave has the effect of accelerating the electrons close to speed of light at the end of the waveguide. In the standing waveguide type there are linear cavities and also side cavities which are used for coupling RF power from one linear cavity to the next. The advantage of the standing waveguide is that a much stronger electric field can be generated and acceleration to required energy can take place over a shorter distance [49].

The length of the waveguide used in the 2100C Varian linear accelerator is 1.0 m which is made parallel to the patient in the superior to inferior direction (see figure 1.1). When the electrons pass through the waveguide, they enter the bending magnet which is a bending chamber surrounded by a strong magnet where the path of the electrons is bent through 270° so the radiation beam can be then directed into the patient.

3.3 Plane-Parallel Ionization Chambers:

Ionization chambers are used in radiation dosimetry to measure the ionization produced in the gas within the chamber. For radiotherapy the gas is usually air. Ion chambers provide the most developed and popular method in dosimetry. When measuring an electron beam, the signal that an ionization chamber gives is proportional to the product of the primary electron fluence and the mass collision stopping power of air. Ionization chambers are popular due to their stability of
performance and high precision of measurement attainable compared to other radiation measuring devices.

When the radiation beam enters the sensitive volume of the chamber it causes ionization of the gas in the chamber producing negative and positive ions. The two parallel plates constitute the electrodes for ion collection (see figure 4.1). The applied polarizing voltage to the electrodes creates an electric field which drives the ion collection process. This causes an ionization current (I) in the external circuit of the chamber which is proportional to the dose rate.

![Figure 3.1 A schematic diagram of an air ionization chamber of a plane-parallel geometry.](image)

The design of the plane parallel ionization chamber reduces the perturbation of the electron fluence when the electrons enter the sensitive volume of the chamber. This is why this chamber is preferred to other chambers types. The low level of perturbation is achieved by using low atomic number (Z) materials in the construction of the chamber, and by making the sensitive air volume in the chamber thin in the electrons forward direction. When the electron beam energy is less than approximately 5 MeV (R_{50} < 2.6) [41], a plane parallel ionization chamber must be used. This is because the
plane parallel chamber has a very thin front window. The depth dose curve of the electron beam exhibits a high surface dose with lower energies, and the dose then rises from the surface to $D_{\text{max}}$ and then drops again rapidly.

There are many types of the plane-parallel chambers. Our interest in this project will be in the Markus chamber (see figure 3.2). This chamber is designed in the form of a coin-shaped air cavity to be used in a position facing the electron source. Figure 3.3 shows a cross section of the plane-parallel chamber designed by Markus in 1976. This chamber has a small sensitive volume (nominally 0.05 cm$^3$) with electrode separation of 2 mm and collector diameter of 5.4 mm.

![Figure 3.2 Parallel-plane Markus electron chamber.](image)

The entrance window is positioned approximately 0.2 mm below the chamber body surface to protect it from damage. It is made of a thin polyethylene foil coated with graphite layer serving as the polarizing electrode. The chamber is built into a
cylindrical acrylic body with a 30 mm diameter and a thickness of 13 mm, so there are materials around the sensitive volume which contribute some side scatter during in-air calibration [52].

**Figure 3.3** Details of the Markus chamber design.(diagram extracted from Klevenhagen) [52].

### 3.4 Scanditronix P-Si Electron Detectors:

P-type silicon diodes have been implemented for clinical use to measure the entrance doses for external beam electron radiotherapy (see figure 3.4). The use of the diodes has significant advantages because of their unmatched spatial resolution as well as their excellent response time. These detectors are manufactured mainly of silicon, traditionally on high resistivity single crystal float-zone material.
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In the future, GaAs might be an alternative to silicon, but now it seems to be an expensive and not fully mastered technology of potentially better radiation hardness [5][14][15][71]. Another possible candidate for the diodes in the future is diamond [4].

Figure 3.4 Electron Field Detector (Blue), Photon Field Detector (Yellow), and Reference Field Detector (Gray).

Semiconductor detectors are employed as a p-n junction which creates a region which is depleted of charge when in operation. When an ionizing particle penetrates the detector it produces electron-hole pairs along its track, the number being proportional to the energy deposited in the depletion region. An externally applied electric field separates the pairs before they can recombine; this causes the holes to drift towards the cathode and the electrons toward the anode; the charge is then collected by the electrodes. A current pulse on the electrode is produced by the collected charge, whose integral equals the total charge generated by the incident particle. The readout
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goes through a charge-sensitive preamplifier; resulting measured charge is directly proportional to the absorbed dose. The silicon detectors are preferred for use in the dosimetry of the total skin electron therapy (TSET) because [26][68]:

- Direct dose measurements can be carried out in electron beams without using stopping power ratio corrections. Since the mass collision stopping power ratio between the water and the silicon is independent of electron energy, an ionization curve, measured with a silicon detector can be directly used as a depth dose curve.
- Silicon detectors have high and uniform spatial resolution in the beam plane because the active area of the detector can be as small as 1 mm which makes them good for relative measurements requiring high spatial resolution.
- The sensitive region of the detector is between 0.5-0.7 mm thick which gives excellent resolution when measuring depth dose curves.
- No need to correct for the temperature or pressure which make them easy to use. Their sensitivity changes with temperature at about 0.35% per °C. However in this project they were used only for relative dosimetry so this variation is of no consequence.

3.5 The DPD-12pc Diode System (Scanditronix):

The DPD-12pc Direct Patient Dosemeter is designed to monitor the radiation dosage during treatment. It can also be used for quality control measurements of therapeutic radiation devices. The DPD-12pc provides multi-channel dose or dose-rate measurements utilizing up to 12 semiconductor detectors.
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The semiconductor detectors detect the radiation in positions defined by the user. The diodes are connected to a multi-channel electrometer (emX) which measures the signal from the diodes, digitizes and normalizes it and transfers it to the display unit (PC). The electron diodes used with the system are EDD-2. These diodes have a 2 mm water equivalent build up, primarily designed to be used in the electron beam measurements. EDD-2 electron diodes have very low directional dependency and cause little perturbation of the field.

3.6 Thermoluminescence Dosimetry (TLD):

Thermoluminescence dosimetry (TLD) has been in use in the measurement of ionizing radiation for nearly 100 years. Different types of TLD materials with varying physical properties and sensitivities allow the determination of absorbed doses for a wide range of radiation doses, energies and radiation particles.

Thermoluminescent dosimeters (TLDs) can be used in radiotherapy at dose levels up to several Gy depending on the TLD material. The major advantages of the TLD detectors are their easy usage when measuring the dose as they require no cables or other external connections during measurements, and their small physical size helps when measuring the absorbed doses for small tissues or organs such as the lens of the eye. Therefore TLD can be used for point dose measurements in phantoms and also for in vivo dosimetry on patients during the treatment [53].

Dose measurement using TLD can be described in two fundamental stages: Stage I, the perturbation of the system from equilibrium into a metastable state; and stage II, the thermally stimulated relaxation of the system back to equilibrium.
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When TLD materials are exposed to ionizing radiation and energy is absorbed from the radiation beam some of the electrons of the crystals in the TLD material are raised to higher energy levels. Most of the electrons immediately return to the ground state, but some remain in trap levels in the bandgap due to impurities in the crystals. When the TLD material is heated the electrons are released from these traps and give up their surplus energy by emitting visible light. The total amount of light emitted is proportional to the number of electrons that were trapped and this in turn is proportional to the amount of dose deposited in the TLD material.

Compared to all other TLD materials, Lithium Fluoride (LiF) has found most widespread application in electron dosimetry because of its high TL output and also its atomic number is very close to the human tissue making it energy independent [74].

3.7 Film Dosimetry:

Radiographic film consists of a transparent base usually made of cellulose acetate or polyester resin coated with an emulsion containing very small crystals of silver bromide. When the ionizing radiation interacts with the crystals, it causes chemical changes that form a latent image. By developing the film, the crystals that had undergone chemical changes are reduced to small grains of metallic silver, while crystals unaffected by the radiation are washed away by the fixing solution, leaving the clear transparent base in their place.

When the film is calibrated against a beam of the required energy it can be then used in radiation dosimetry but only over suitable range of doses. The radiation dose received by the film is proportional to the optical density (OD) of the exposed film,
which is a function of the amount of light transmitted through a specific portion of the film. It can be described mathematically by:

\[
OD = \log \left( \frac{I_o}{I_t} \right)
\]  

(3.1)

Where \(I_o\) is the incident light intensity on the film and \(I_t\) is the transmitted light intensity through the film. The reasons behind using the logarithmic scale when representing the optical density are the large differences in numerical values.

To relate the optical density measured from the film to the dose absorbed by the film, the sensitometric curve has to be measured. This is done by calibrating the film by exposing it to a range of known doses, then plotting the relation between the optical density measured from the film and the dose given to it. To have good results, calibration films from the same batch should be exposed and developed at the same time.

A special device which is used to analyze irradiated films is called a film densitometer. It consists of a small aperture through which the light is directed and a light detector, like a photocell, to measure the light intensity transmitted through the film.

The system used in the project is the Scanditronix RFA-300 film densitometer which is mounted on the top of the Scanditronix RFA water phantom (see figure 3.5). The film is placed on a plate of glass placed on the top of the water phantom. By setting up the required parameters in the computer system controlling the film densitometer, the system can automatically scan the film and profiles can be plotted and stored in the computer for later analysis.
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It is not easy to keep the same developing conditions for the film every time the film is used, such as temperature and concentration of the developing chemicals; this is why the film is not usually used to measure the absolute dose.

When a low energy photon beam is used, the photoelectric effect is the major process, and because of the dependence of this process on the $Z^4$; where $Z$ is quite high because the film contains silver particles, the film will have an over response. This is why film dosimetry is usually used in relative measurements; on the other hand, film has the ability to record a two dimensional distribution of dose with a single irradiation. Film has the highest spatial resolution of all dosimeters [50].

3.8 Three-dimensional water scanner:

Electron diodes and ionization chambers in conjunction with a three dimensional scanner were used to measure the beam parameters of the Clinac 2100C. The chamber or diode can be placed by the scanner to any position within the water tank of dimensions 50x50x50 cm$^3$, see figure 3.5. Since the output of the linear accelerator may vary with time, a stationary reference ionization chamber is used to enable the system's computer to store the relative signal between the field and the reference detector. Having this system is very useful to do the linear scans to obtain depth doses and beam profiles.

The radiation field analyzer instrument available in the institute is the RFA-300 (Scanditronix) which can perform linear scans to measure depth doses, beam profiles and two dimensional isodose distributions. Another good option provided with the system is the film densitometer which is a two dimensional scanner.
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A servo mechanism is fitted to the water tank to allow positioning the field detector in the three dimensional water tank. This helps control the movement of the field detector in three dimensions.

![Figure 3.5 Scanditronix RFA-300 water phantom.](image)

### 3.9 Phantoms:

It is not possible to measure the dose distribution directly in the patient, this means that the data needed to calculate the dose distribution in the patient is collected from measurements in water phantoms. Water is commonly used as a substitute for tissue in phantoms, because it is very similar to the absorption and scattering properties of the soft tissue, and the human body being 75% of water by weight [85]. Water has become a standard phantom material for radiotherapy measurements [78] because it is naturally abundant and cost effective, and has universally reproducible radiation properties.

The use of a water phantom for the dosimetry can cause practical difficulties. For example, when the ionization chamber is submerged in the water it can experience leakage. Also, when measuring the dose at the surface of the water the surface tension
can cause an uncertainty in the position of the chamber which affects the results of the measurement [3]. Additionally, water is not suitable for use in film dosimetry. For these reasons a solid substitute for water “solid water” was developed for phantoms. The attenuation and scattering properties of the solid water phantoms resemble those of water. It means that, the same thickness of both the water and the solid phantom will have the same absorption and scattering properties. This requires that both phantom materials will have matched mass attenuation coefficients and matched electronic densities. In this project PLASTIC WATER™ from Nuclear Associates was used as the solid phantom. This phantom comes in slab size 30 cm x 30 cm and with different thicknesses (2 mm, 3 mm, 5 mm, and 10 mm). This phantom can be used in the project for absolute and depth dose measurements. There are no correction factors required for this phantom, and it can be setup and used easily in the measurement. The other advantage of using this phantom is that it is compatible with a variety of ion chambers. Table 3.1 compares the depth doses between water and plastic water for different electron energies.

In the measurements of dose uniformity, Perspex plates of thickness 10 mm and size 2 m x 1 m were used to measure the dose uniformity at the surface of the plates. The use of these large plates in this project reduced the complexity in the measurements especially when the dose uniformity is scanned along a large field (80cm x 170cm). The maximum scan dimension available is 50 cm (RFA water tank), which cannot cover the whole field.

A paraffin wax cylinder phantom was used also in this project to measure the B factor when the calibration of the high dose rate 6 MeV electron beam was carried out. The size of this phantom is 67 cm in diameter and 30 cm in height.
Table 3.1 Depth-dose comparison between water and Plastic Water for electron beams [35].

Another type of phantom that was used in this project is the RANDO Phantom, see figure 3.6. This phantom allows detailed mapping of the dose distribution in a realistic anatomical scenario which is of considerable assistance in evaluating radiotherapy plans [40]. Two types of RANDO phantom are available, RANDO man and RANDO woman. Accessories are available for both types (breast accessories).

RANDO phantoms are designed with a natural human skeleton cast inside material that is radiologically equivalent to soft tissue. The phantom is designed with different densities to simulate different areas in the body such as, lungs, soft tissue or skeletons.

Films or individual dosimeters can be used with the RANDO phantom for radiation dosimetry. Special holes are drilled in the phantom slices to be used for dosimetry purposes.

| Percent Dose | Depth (cm) | | | | |
|--------------|------------|------------|------------|------------|------------|------------|
|              | 6 MeV      | 12 MeV     | 20 MeV     |            |            |            |
|              | Plastic Water | Water    | Plastic Water | Water    | Plastic Water | Water    |
| 100%         | 1.2        | 1.1        | 2.4        | 2.4        | 1.2        | 1.2        |
| 80%          | 1.8        | 1.7        | 3.9        | 3.8        | 6.6        | 6.6        |
| 50%          | 2.2        | 2.1        | 4.6        | 4.5        | 8.2        | 8.1        |
3.10 Electrometers:

Electrometers are used to measure the current produced in the ionization chamber. The current cannot be measured with an ammeter, because the produced current is very small, in the range $10^{-6}$ to $10^{-14}$ ampere. Special electrometer circuits have been designed for this purpose [3][50]. The most widely used is the negative feedback operational amplifier that can be thought of simply as an ultrahigh impedance voltmeter. In figure 3.7, the triangle represents the operational amplifier and the positive and the negative inputs refer to the inverting and non-inverting inputs. When a negative charge $Q$ flows from the ionization chamber, the input circuit is driven to the negative potential $V_i$, while, the output potential rises to $10^5$ times greater positive potential $V_o$ which is applied to the capacitor $C$. The total potential across $C$ is $V_i + V_o$, where the charge is:

\[ C (V_i + V_o) = Q - C_i V_i \]  \hspace{1cm} (3.2)
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Where $C_i$ is the capacitance distributed to the ground from the input. The input impedance of the OP amp can be assumed to be high to inhibit the passage of the charge. The charge collected $Q$ is determined from the measured voltage ($V$) across the capacitor and the known value of the capacitance $C$.

**Figure 3.7** Operational-amplifier (Op amp) electrometer circuit for charge measurement. To measure the current, the capacitor is replaced by a resistor.

3.11 Summary:

In this chapter the dosimetry equipment used in the project was described. Also a basic introduction about the major components used in this study and their functions was presented.

The next chapter is a description of the measurements carried out to study the beam characteristics of the high dose rate 6 MeV electron beam with different setups. The measurements described in Chapter 4 used no beam modifiers.
Chapter 4

Total Skin Electron Measurements without Beam Modifiers

4.1 Introduction:

This chapter describes the measurements carried out to study the characteristics of the 6 MeV electron beam based on the dual six field technique. All the measurements in this chapter were carried out for the purpose of this research. The results from this project were compared to the old data collected in year 2000 which they did meet. The reason for repeating the measurements is to use the same equipment for both cases; with and without electron beam modifiers.

Most of measured data in this project were carried out using flat geometries (flat phantoms), which is not representative to the clinical realities. Since the human body is not a flat surface shape or a simple cylindrical, not only are there areas of overexposure, but there are marked unexposed areas which often requires supplementary treatments. In vivo TLD dose measurements can be used to identify areas requiring local boost fields. The dose uniformity will change depending on the shape and size of the patient.

The measurements in this chapter were carried out without using beam modifiers. The main topic of the measurements described in this chapter includes:

- Determination of the dual angle which will be used in the project.
- Study of the cable effects when irradiating with large electron fields.
- Central percentage depth dose measurements for a single field, a dual stationary field, and dual six fields.
- Off axis percentage depth doses along the vertical and horizontal planes for a dual stationary field.
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- Dose uniformity along the vertical and horizontal planes.

### 4.2 The matching of dual field TSET system:

The purpose of the measurements was to determine the dual field angle, such that in the composite dose distribution, acceptable dose uniformity could be achievable. The angle and associated gap between the light field edges is dependent on the beam energy and the scattering conditions.

Measurements were carried out using the computerized Scanditronix RFA water phantom. A p-Si electron diode was used in this measurement. The water phantom was positioned at a distance of 3.5 m SSD. The diode was positioned facing the electron beam as shown in figure 4.1. The direction of the beam scan was in the vertical direction from superior to inferior, behind the vertical thin Mylar window which forms part of the surface of the water phantom. The diode was moved to a distance of 10 mm (depth) in the phantom along the horizontal line, then a vertical scan was done from -180 mm to +180 mm in the vertical direction with electron fields directed at $\pm 15^\circ$, $\pm 17.5^\circ$, and $\pm 20^\circ$ angles above and below the horizontal line. Profiles in the vertical direction from each dual field were summed to obtain a composite profile. Given the constraints of the set-up, scanning range was limited to cover $\sim 36$cm at best, in the vertical direction without moving or adjusting the water phantom position. The steps of the scans were done at 2 mm. This why the produced curves appear to be smooth.

Results of the measurements are shown in figures [4.2 to 4.7]. Figure 4.2 shows the dose profile for the two single fields at $250^\circ$ and $290^\circ$ ($\pm 20^\circ$). The summation of both curves is shown in figure 4.3. The curves were normalized to the reference point of (0,0,10 mm) along the horizontal line. The dose uniformity was found to be within 4 %, from 100% to 104% along the vertical direction.
Figure 4.1 Schematic drawing of the set-up used for dual angle determination using the RFA water phantom and electron diode.

Figure 4.4 shows the dose uniformity measured for the two single fields using $\pm 17.5^\circ$ angle above and below the horizontal line (252.5$^\circ$ and 287.5$^\circ$), while figure 4.5 shows the summation of both curves. The variation in the dose uniformity fell down to 2% at these angles, from 100% to 98% along the vertical direction.

Figure 4.6 shows the dose uniformity measured for the two single fields using $\pm 15^\circ$ angle above and below the horizontal line (255$^\circ$ and 285$^\circ$), while figure 4.7 shows the summation of both curves. The variation in the dose uniformity was found to be 6%, from 100% to 94% along the vertical direction.

Thus from the summation of the measured profiles, the best dose uniformity achieved over the 36 cm region in the vertical plane was found to be when using the $\pm 17.5^\circ$ angles (see table 4.1).
Dose uniformity results of three different angles used to determine the dual angle in TSET.

<table>
<thead>
<tr>
<th>Angle</th>
<th>Dose uniformity</th>
</tr>
</thead>
<tbody>
<tr>
<td>± 15°</td>
<td>6%</td>
</tr>
<tr>
<td>± 17.5°</td>
<td>2%</td>
</tr>
<tr>
<td>± 20°</td>
<td>4%</td>
</tr>
</tbody>
</table>

Table 4.1

Figure 4.2 Beam profiles measured along the vertical axis using two fields ±20° angle above and below the horizontal axis.

Figure 4.3 Summation of profiles measured using ±20° angle above and below the horizontal line at 10 mm depth.
Figure 4.4 Beam profiles measured along the vertical axis using two fields ±17.5° angle above and below the horizontal axis.

Figure 4.5 Summation of profiles measured using ±17.5° angle above and below the horizontal axis.
Figure 4.6 Beam profiles measured along the vertical axis using two fields $\pm 15^\circ$ angle above and below the horizontal axis.

Figure 4.7 Summation of profiles measured using $\pm 15^\circ$ angle above and below the horizontal axis.
4.3 Cable Effect Measurements:

It is very important to minimize the radiation induced signals from the cable of the ionization chamber in radiation dosimetry of radiotherapy beams. In standard radiotherapy measurements the extra signal, mainly caused by the irradiation of the chamber physical structure (surrounding the active volume) is prevented by the guard ring. This effect can be reduced by optimizing the chamber design and by taking the average of the electronic signals measured at two equal but opposite polarizing potentials (Gerbi and Khan 1987) [24].

When a total skin electron therapy treatment is applied, a large electron field is used, (as large as 120 cm x 200 cm), this could cause an extra electronic signal to be produced as a result of the direct irradiation of the chamber's cable or its connector by the electrons from the main field. Reasons for this effect are:

i. The TSET electrons may knock out the electrons in the cable's conductor, creating positive charges in the conductor, and then the electrons from the TSET beam may stop in the conducting media, causing additional negative charges. If the number of knocked out electrons is not the equal to the stopped TSET electrons, then the net cable induced electronic charge becomes a part of the measured signal. The magnitude of this effect depends on the energy of the primary electrons, the length of the irradiated cable, and the radiation exposure time. This effect can be reduced by using the two polarity averaging technique by using two equal voltages but of opposite polarity.

ii. The extra-cameral volumes present in the irradiated cable and connector may cause an electronic signal similar to that produced in the chamber's active volume. This type
of effect can not be reduced by using the two polarity averaging technique, but can be
by shielding the irradiated cable [12][13][23][45][62].

The measurements carried out in this project were to investigate the induced effect of the
ionization chamber cable when irradiated with large electron fields. First, measurement was
carried out to check the response of the Markus parallel plate chamber when different parts of
the cable, connector, and extension cable are irradiated. The setup for this measurement was
as follows: The Markus chamber embedded in solid phantom plates was then exposed to a 6
MeV high dose rate electron beam at 100 cm SSD. The HDTSe applicator was used to open
the collimator to 36x36 cm. 250 MU was used to measure the charge at the surface of the
cable at different distances up to 200 cm distance along the cable. The polarity was -300V.
The extracameral signal measured for the irradiation of the chamber's cable, the adaptor
connecting between the chamber cable and the extension cable, and the extension cable to the
electrometer. The results are shown in figure 4.8.

**Figure 4.8** Extracameral signal measured at different distances along the chamber's cable, the
adaptor, and the extension cable.
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Figure 4.8 shows that the main contribution of the change in the extracameral signal is at the adaptor connection [67] which is about 2.4% of the total signal. The errors during the measurement can be reduced by keeping the adaptor outside the irradiation field. Other way is to shield the cable and the adaptor. A 10 cm of solid phantom material was used to shield the adaptor during the irradiation the percentage of extracameral signal was reduced to about 0.5% of the total signal. When a steel pipe of 1 cm thickness was used to shield the cable and the adaptor the signal was reduced to about 1%.

Another measurement was carried out to check the effect of the length of the irradiated cable on the extracameral signal. A similar setup was used as the previous measurement except that the length of the irradiated cable was increased every time by 30 cm increments. The results are shown in figure 4.9.

![Figure 4.9 Extracameral signal measured at different cable lengths.](image-url)
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The extracameral signal measured for the irradiation of the chamber's cable, connectors between the chamber and the extension cable, and the extension cable to the electrometer. In figure 4.9, the percentage of extracameral signal of the total signal was about 5.5% for a cable length 210 cm. The contribution from the adaptor is about 2.4% (similar to the previous measurement). This effect can be reduced by shielding the adaptor as mentioned above.

The last part of the cable effect measurements was carried out to measure the difference in the depth dose curves for a large electron field with and without the cable shielding. A parallel plate Markus ionization chamber was used with an NE-Ionex Dosemaster electrometer. The chamber was positioned in a solid phantom at a distance of 350 cm SSD. A single horizontal 6 MeV electron beam with gantry at 270°, high dose rate total skin electron mode (HDTSe), an effective field size of 126 cm x 126 cm was used. 1000 MU readings were taken. A depth ionization curve was measured without shielding the cable, and then the same measurement was repeated with the cable shielded by a steel pipe. The results are shown in figure 4.10.

![Figure 4.10 Depth ionization curves measured with and without cable shielding.](image-url)
Chapter 4

Figure 4.10 shows the depth ionization curves with and without cable shielding. The features seen from the diagram are:

- The percentage surface dose is the same for both (~78%);
- The maximum depth of absorbed dose is the same for both (~12 mm);

4.4 Percentage depth dose curves along a horizontal line:

The purpose of these measurements is to study the characteristics of the 6 MeV electron beam. The most important characteristic of an electron beam is its energy spectrum. The energy spectrum of an electron beam depends on the intrinsic energy spread of the accelerator beam. The beam is also modified when passing through various materials by scattering and energy loss. Figure 4.11 shows the differential distribution in energy of an electron beam when the beam passes through a slab of material [45]. It is not easy to measure this distribution because the high fluence rate (flux density) can cause saturation for the radiation detector.

![Figure 4.11 Distribution of electrons in energy after passing though a slab of material (from ICRU 35 [45])]
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Figure 4.11 shows two important energy parameters at the waveguide exit window and at depth in a phantom: the “most probable energy”, $E_p$, which is the energy value corresponding to the peak of the curve, and the “mean energy”, $\bar{E}$, which lies usually below $E_p$. These two parameters can be estimated by practical measurements on the linear accelerator.

In this project, depth dose measurements along the central line were carried out for the following setups:

- Single field at 100 cm SSD with gantry angle at 0° and;
- Single field at 3.5 m SSD with gantry angle at 270° and;
- Dual stationary field at 3.5 m SSD with gantry angles 252.5° and 287.5°.

The depth ionization curves were obtained using the Markus ionization chamber with the Ionex Dosemaster using the square solid water slabs of 30 cm x 30 cm size. The depth ionization was converted to depth dose by applying stopping power ratios ($S_{w/air}$) water to air for electron beams. The mean energies at the surface, $\bar{E}$, and at depth $z$, $E_z$, are required to make a selection of physical data such as $S_{w/air}$ in order to do the conversion from depth ionization to depth dose. The mean energies $\bar{E}$ and $E_z$ can be determined from the measured range parameters:

(i) Range (depth) parameters $R_{50}$ and $R_p$. The mean energy of an electron beam at the surface of the phantom, $\bar{E}$, can be related to the depth in water at which the ionization is 50% of the maximum ionization, $R_{50,I}$, or the depth at which the absorbed dose is 50% of the maximum absorbed dose, $R_{50,D}$. The practical range $R_p$ is defined as the depth at which the tangent to the steepest part of the descending region of the depth ionization or depth dose curve intersects with the extrapolation of the bremsstrahlung tail of the distribution (see figure 4.13). The practical range $R_p$ is required to estimate the energy at depth, $E_z$. 

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The depth ionization curve measured in water along the central axis can be used to determine $R_{50,I}$, while $R_{50,D}$ can be determined either by converting the measured depth ionization curve using stopping power ratios or by measuring the central axis depth dose curve directly by using p-type diodes in water.

(ii) The most probable energy, $E_p$. It can be calculated from the empirical relation derived by Markus [61] between energies and range parameters from depth dose or depth ionization curves. The empirical relation is:

$$E_p = 0.22 + 1.98 R_p + 0.0025 (R_p)^2$$

(4.1)

(iii) The mean electron energy at the surface of the phantom, $\bar{E}$. In general the relationship between $\bar{E}$ and $R_{50}$ is taken to have the form

$$\bar{E} = C R_{50}$$

(4.2)

Where $C$ is a constant. Equation 4.2 can be used only when $R_{50}$ has been obtained from depth ionization or depth dose curves for a non-divergent beam at fixed source-to-chamber distance (SCD). When $R_{50}$ is taken from a dose distribution obtained with a constant SSD then the tabulated data provided by NACP (1980) [64] determine $\bar{E}$ either from depth dose curves or depth ionization curves. These tabulated data given in the form of a second order polynomial (IAEA 1996) [44].

$$\bar{E} \text{ (MeV)} = 0.818 + 1.935 R_{50,I} + 0.040 (R_{50,I})^2$$

(4.3)

for $R_{50,I}$ determined from a depth ionization curve and

$$\bar{E} \text{ (MeV)} = 0.656 + 2.059 R_{50,D} + 0.022 (R_{50,D})^2$$

(4.4)
for \( R_{50,D} \) determined from a depth-dose curve.

\[ (iv) \text{ The mean electron energy at depth } z, E_z. \text{ The mean electron energy at depth is given as a function of } z/R_p. \text{ The data is shown in table } 4.2. \]

### 4.4.1 Central axis depth dose curve at standard 100 cm SSD:

An RFA water phantom and parallel plate Markus chamber were used for this measurement. The gantry was at 0\(^\circ\) and the distance to the surface of the phantom (SSD) was 100 cm. The High Dose Rate Total Skin Electron applicator (HDTSe) was used. The field size at 100 cm SSD was 36 cm x 36 cm. The measurement was carried out up to depth 5 cm in the phantom. A depth ionization curve was measured (see figure 4.12) then the curve was converted to a depth dose curve by applying the stopping power ratios (see figure 4.13).

The mean electron energy at the surface of the phantom, \( \bar{E} \), can be found from the tabulated data provided by NACP (1980) [64] either from ionization curves measured by ionization chamber or from depth dose distributions; this also given by IAEA (1987) [43], or by using the second order polynomial equations 4.3 and equation 4.4.

Using the value of \( R_{50,D} \) found from figure 4.14 in equation 4.4 gives:

\[
\bar{E} \text{ (MeV)} = 0.656 + 2.059 (2.37) + 0.022 (2.37)^2
\]

\[
\bar{E} \text{ (MeV)} = 5.7 \text{ MeV}
\]

The most probable energy \( E_p \) can be calculated from equation 4.1:

\[
E_p = 0.22 + 1.98 (2.9) + 0.0025 (2.9)^2
\]

\[
E_p = 6.0. \text{ MeV}
\]
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Using the IPEMB code of practice for electron dosimetry for radiotherapy beams (1996) [41], the following beam parameters were found:

\[ D_s = 73.2\% \]
\[ d_{\text{max}} = 15 \text{ mm} \]
\[ R_{50, D} = 24 \text{ mm} \]
\[ R_p = 29 \text{ mm} \]
\[ D_x = 0.5\% \]

where \( D_s \) is the percentage surface dose, \( d_{\text{max}} \) is the depth at which the absorbed dose is maximum, \( R_{50, D} \) is the depth at which dose is 50% of that at the maximum for the depth-dose curve, \( R_p \) is the practical range, and \( D_x \) is the percentage x-ray contamination.

**Figure 4.12** Depth ionization curve for single field at standard 100 cm SSD measured in RFA water phantom using a parallel plate Markus ionization chamber.
Table 4.2 The ratio of the mean energy at depth $z$, $E_z$, and the mean energy at the phantom surface, \( \bar{E} \), for electron beams in water; depths are given as fractions of the practical range, $R_p$ (Andreo and Brahme 1981) [1]; values at 2 MeV are extrapolated from table V of IAEA 1987 using a third degree polynomial fit at each depth.
Figure 4.13 Depth dose curve for 6 MeV electron beam measured in water at 100 cm SSD. The data was converted from the ionization curve to the depth dose curve by multiplying by the stopping power ratios at different depths.

4.4.2 Central axis depth dose curves for a single horizontal electron field at 350 cm SSD:

This measurement was carried out using the parallel plate Markus chamber with the Ionex Dosemaster electrometer and the solid phantom. A steel pipe to shield the cable was used in the measurement. The gantry was rotated to 270° and the HDTSe applicator was used. The collimator was automatically opened to 36 cm x 36 cm field size. The phantom was mounted on the top of a wooden table with a distance to the surface of the phantom of 350 cm SSD. Measurements along the central axis were carried out and the depth ionization curve was measured by adding slabs of the solid phantom to the surface of the phantom and adjusting the SSD to 350 cm. The depth ionization curve produced was converted to the depth dose curve shown in figure 4.14.

From the depth dose curve it was found that $R_{50,D} = 21$ mm; using equation 4.4 gives:
\[ \bar{E} \text{ (MeV)} = 0.656 + 2.059 (2.12) + 0.022 (2.12)^2 \]

\[ \bar{E} \text{ (MeV)} = 5.1 \text{ MeV} \]

while, the most probable energy, \( E_p \), is:

\[ E_p = 0.22 + 1.98 (2.7) + 0.0025 (2.7)^2 \]

\[ E_p = 5.6 \text{ MeV} \]

From figure 4.14 it was found that the beam parameters are:

- \( D_x = 75\% \)
- \( d_{\text{max}} = 12 \text{ mm} \)
- \( R_{50,D} = 21 \text{ mm} \)
- \( R_p = 27 \text{ mm} \)
- \( D_x = 1.5\% \)

**Figure 4.14** Depth dose curve for 6 MeV electron beam measured in solid water at 350 cm SSD. The measurements were made with a single horizontal beam directed at point (0,0) in the treatment plane using a parallel plate ionization chamber. The curve was converted to depth dose from depth ionization.
4.4.3 Depth dose curves for a dual stationary field at 350 cm SSD:

Two angled fields can provide improved dose uniformity over areas the size of patient dimensions. In these measurements, two equal exposures were given, one from each of the two angled components. The measurement was carried out using the parallel plate Markus chamber with the Ionex Dosemaster electrometer. The cable of the chamber was shielded with steel pipe. The measurement was carried out in a solid water phantom mounted on the top of a wooden table. The distance to the surface of the phantom was 350 cm SSD. The measurement was done along the horizontal line. The dual angles used were 252.5° and 287.5°. The readings at the two angles were summed, and then the depth ionization curve was obtained. The depth ionization curve was converted to the depth dose curve using stopping power ratios. The results are shown in figure 4.15.

From the depth dose curve it was found that \( R_{50,D} = 20.2 \text{ mm} \); using equation 4.4 gives:

\[
\bar{E} \text{ (MeV)} = 0.656 + 2.059 \times (2.02) + 0.022 \times (2.02)^2
\]

\[
\bar{E} \text{ (MeV)} = 5.0 \text{ MeV}
\]

while, the most probable energy, \( E_p \), from equation 4.1 is:

\[
E_p = 0.22 + 1.98 \times (2.58) + 0.0025 \times (2.58)^2
\]

\[
E_p = 5.3 \text{ MeV}
\]

From figure 4.16 it was found that the beam parameters are:

\( D_x = 78\% \)

\( d_{\text{max}} = 11 \text{ mm} \)

\( R_{50,D} = 20 \text{ mm} \)

\( R_p = 26 \text{ mm} \)

\( D_x = 1.0\% \)
Figure 4.15 Depth dose curve for dual stationary field using angles 252.5° and 287.5° gantry angles at 350 cm SSD.

Figure 4.16 shows the three depth dose curves produced at standard 100 cm SSD, single horizontal field at 350 cm SSD, and dual stationary field at 350 cm SSD. An important feature that can be seen in the figure was the shift of the depth of the maximum towards the surface with the increase in the SSD. This is because the electrons lose more energy when they travel a longer distance through air. Using figure 4.16 and table 4.3 which records the clinical parameters for the 6 MeV TSET beam derived from the measured percentage depth dose curves, it was found the following:

- The mean electron energy at the surface for single field at 100 cm SSD, single horizontal field at 350 cm SSD, and dual stationary field at 350 cm SSD were 5.7 MeV, 5.1 MeV, and 4.9 MeV respectively. The change in the mean energy with the dual stationary field is more than that for the single horizontal field because the
measurements were carried out along the central axis of the horizontal single field, while for dual stationary field; it was along the horizontal line through the gap between the two angled fields. No direct beam was involved in the measurement of the dual stationary field. Also by using dual field technique the source to surface distance along the central axis of each field is increased (>350 cm SSD).

- As shown in figure 4.16, the depth of maximum absorbed dose shifts slightly towards the surface when the SSD is increased. This is due to the energy degradation of the electron beam when the source to surface distance is increased. The electrons lose energy as they travel in air.

- The electrons percentage depth doses are not strongly dependent on the distance between the source and the surface of the phantom. The energy degradation is small because of the low density of air. It causes a small change in beam penetration when comparing the energy at 100 cm SSD and at 350 cm SSD.

- A slight increase in the x-ray contamination (bremsstrahlung) occurs when the distance increases from 100 cm to 350 cm. It increases from 0.5% for the single field at 100 cm SSD to 1.5% for the single horizontal field at 350 cm SSD. When the source to surface distance increases, the electron scattering increases within air which causes the electron dose to reduce, whereas the photon dose stays constant, thus giving rise to an increase in bremsstrahlung component. It was shown by El-khatib et al [20] that the bremsstrahlung component can be reduced to nearly the level of the scanned beam by modifying the ion chamber and scattering foils of the linear accelerator. In the Stanford technique, described by Karzmark et al [48][49], the electron beam after emerging from the accelerator window, is scattered by a mirror (0.028 inch Al), an
aluminum scatterer located externally at the front of the collimator (0.037 inch Al), and about 3 m of air before incidence on the patient. Also, the x-ray contamination is reduced by angling the beam 10 degrees to 15 degrees above and below the horizontal line. Because most of the x-rays produced in the scatterers are directed along the central axis, they largely miss the patient. The x-ray contamination may vary from one accelerator to another for the same beam energy and depends significantly on the scatterer – degrader materials in the beam.

- The x-ray contamination of the single field at 350 cm is higher than the dual stationary field because most of the produced x-rays are directed along the central axis.

**Figure 4.16** Comparison between the depth dose curves for the TSET 6 MeV electron beam at standard clinical distance 100 cm SSD and at TSET treatment distance 350 cm SSD for a single field and a dual stationary field.
### Parameters

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Single field at 100 cm SSD</th>
<th>Single field at 350 cm SSD</th>
<th>Dual stationary field at 350 cm SSD</th>
</tr>
</thead>
<tbody>
<tr>
<td>$d_{\text{max}}$</td>
<td>15 mm</td>
<td>12 mm</td>
<td>11 mm</td>
</tr>
<tr>
<td>$R_{50,D}$</td>
<td>24 mm</td>
<td>21 mm</td>
<td>20 mm</td>
</tr>
<tr>
<td>$R_p$</td>
<td>29 mm</td>
<td>27 mm</td>
<td>26 mm</td>
</tr>
<tr>
<td>$D_x$</td>
<td>0.5%</td>
<td>1.5%</td>
<td>1.0%</td>
</tr>
<tr>
<td>$E_p$</td>
<td>6.0 MeV</td>
<td>5.6 MeV</td>
<td>5.3 MeV</td>
</tr>
<tr>
<td>$\bar{E}$</td>
<td>5.7 MeV</td>
<td>5.1 MeV</td>
<td>4.9 MeV</td>
</tr>
</tbody>
</table>

**Table 4.3** Beam parameters comparison between single fields at 100 cm, 350 cm SSD, and a dual stationary field for the clinical 6 MeV HDTSe.

### 4.5 Off-axis depth dose curves for dual stationary field in the vertical plane:

This measurement is important because it gives information about the uniformity of beam depth penetration over the entire height of the patient when the dual field technique is used. The solid phantom was positioned on the top of a wooden table at 350 cm SSD. Kodak X–Omat V films were used in the measurement. The films were sandwiched between the solid phantom slices at different off axis distances in the vertical direction as shown in figure 4.17. Films were positioned at the central line, 20 cm, 40 cm, 60 cm and 80 cm along the superior direction. The HDTSe applicator was used and the field size was 36 cm x 36 cm at 100 cm SSD. Films were exposed using the 252.5° and 287.5° dual field using 200 MU per field. They were developed at the same time and under the same conditions (see figure 4.18).
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Figure 4.17 schematic drawing of the film positions in the solid phantom.

Figure 4.18 A radiographic film placed in the solid phantom at the level of beam axis irradiated with dual stationary field. The dark region indicates the electron dose delivered to the film.
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The computerized Scanditronix RFA 300 water phantom with the film densitometer was used to scan the films to obtain the percentage depth doses. The results were normalized to the maximum dose along the horizontal line (at the 0 cm film). The results are shown in figure 4.19.

Distinct features can be seen when analyzing the curves, from the films, as the off axis distance increases:

- The percentage surface dose at 0, 20 cm, 40 cm, 60 cm, and 80 cm distances are 85%, 86%, 83%, 84%, and 79% respectively. These results show that the surface dose around the center is almost flat and it decreases at 80 cm off axis distance.

- When the depth doses were normalized to the $D_{\text{max}}$ of each individual film, it is clearly seen that the depth of maximum absorbed dose moves slightly towards the surface as the off-axis distance increases. This is shown in figure 4.20. This effect is caused by the energy degradation by increasing the source to surface distance (SSD). The beam parameters of the 80 cm off axis film is showing that $R_p= 23$ mm, $R_{50}= 17$ mm. then the “mean energy”, $\bar{E}$, and the “most probable energy”, $E_p$, were found to be 4.1 MeV and 4.8 MeV respectively, while for the film positioned along the central line (0 cm off axis), $\bar{E}$ and $E_p$ were found to be 4.4 MeV and 5.1 MeV respectively. A summary of beam parameters at different off axis distances is shown in table 4.4. The mean energy measured at 80 cm off axis was found to be higher than the 60 cm off axis distance. This is due to the increase of mean energy when the measurement is carried out along the central axis of a single electron field at extended distance. 80 cm is closer to the central axis of the upper field.
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- The percentage of x-ray contamination (bremsstrahlung) increases slightly from about 0.6% to about 1% (see figure 4.21). It is probably mainly due to the measurement plane moving close to the central axis of upper beam.

- The curves appear to be very smooth, this due to the small steps used to scan the films (0.2 mm).

<table>
<thead>
<tr>
<th>Off axis distance (cm)</th>
<th>Rp (cm)</th>
<th>R_50 (cm)</th>
<th>E (MeV)</th>
<th>E_p (MeV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0 cm</td>
<td>2.4</td>
<td>1.8</td>
<td>4.3</td>
<td>5.1</td>
</tr>
<tr>
<td>20 cm</td>
<td>2.4</td>
<td>1.8</td>
<td>4.3</td>
<td>5.0</td>
</tr>
<tr>
<td>40 cm</td>
<td>2.4</td>
<td>1.7</td>
<td>4.2</td>
<td>5.0</td>
</tr>
<tr>
<td>60 cm</td>
<td>2.3</td>
<td>1.6</td>
<td>4.0</td>
<td>4.8</td>
</tr>
<tr>
<td>80 cm</td>
<td>2.3</td>
<td>1.7</td>
<td>4.1</td>
<td>4.8</td>
</tr>
</tbody>
</table>

Table 4.4 Off axis beam parameters of a dual stationary 6 MeV field.

Figure 4.19 Off-axis depth dose curves for a pair of angled beams (252.5° and 287.5°) without using filters. Curves were normalized to D_{max} of the curve produced from the film positioned along the central line (0 cm).
Figure 4.20 Percentage depth dose curves in the region of $D_{\text{max}}$. Curves were normalized to the maximum dose of each curve.

Figure 4.21 Tail area
4.6 Depth dose curve using dual six fields along central line:

So far in this chapter the beam parameters measured and analyzed were for stationary beams and the direction of the phantom surface was not changed. But the technique I want to study is the dual six fields in which the phantom is rotated 60° after every dual stationary field (252.5° and 287.5°).

The prescribed depth for the treatment of mycosis fungoides varies depending on the stage of the disease and its type. It also depends on the shape of the body. All of the measurements carried out until now are with a flat surface phantom.

In this measurement a RANDO phantom was used. The film was cut to fit in between the RANDO phantom slices number 20 and 21. The phantom was positioned on the top of a rotating stand at a distance of 350 cm SSD along the horizontal central line. Dual six fields were used to expose the film with 200 MU per field. The phantom was rotated every 60° to make a full cycle using angles 0°, 60°, 120°, 180°, 240°, and 300°. The film was then developed and scanned using the computerized Scanditronix RFA 300 phantom with the densitometer. The resulting film is shown in figure 4.22. The dark area around the film represents the change in the density which is related to the dose given from the six fields.

Figure 4.23 shows the percentage depth dose for dual six fields using gantry angles 252.5° and 287.5°. The dose was normalized to the maximum dose at a depth of about 0.4 mm or almost at the surface along the horizontal line at calibration point (0,0).

The depth dose curve for the dual six fields and the dual stationary field are represented together in figure 4.24. It shows the rapid fall-off dose with depth for the dual six field
irradiations in comparison with the dual stationary field. This is a favorable feature when the beams are used for the treatment.

Figure 4.22 Radiographic film placed in-between the slices of the RANDO phantom at the level of the beam axis, irradiated with dual six electron fields.

It was found that the surface dose from the dual stationary field is about 85 % and the 90 % dose is at a depth of 16 mm, suggesting that the dual stationary field alone is not suitable for TSET treatment. This means that the beam is too penetrating to be used for TSET.

On the other hand, when the dual six fields are used the surface dose was found to be ~92%, the 90% is at 1.5 mm depth, and the maximum dose is almost at the surface (~0.4 mm).

The beam penetration properties using the dual six fields were significantly changed. The two possible reasons for this change are the change in the source to surface distance (SSD) during the rotation and the effect of the oblique beam entrance.
The effect of the change in the source to surface distance (SSD) may not be the predominant mechanism; oblique incidence may be more important. The SSD was discussed earlier in this chapter in section 4.4 and was shown in figure 4.20. It was found that the electrons percentage depth doses are not strongly dependent on the distance between the source and the phantom. The energy degradation is small because of the low density of air. Therefore, the changes between the dual stationary field and the dual six fields (rotational) can only be caused by oblique electron beam incidence effects.

The oblique incidence of an electron beam can change the central axis depth dose and shift the depth of maximum dose to the surface. These effects were discussed by Ekstrand and Dixon [18]. The incident beam can be considered as an integration of many pencil beams or slit beams. Depending on the depth in the phantom, neighboring pencil beams may have more or less contribution to the dose at any point at the central axis compared to the contribution at normal electron beam incidence. Figure 4.25 explains the effect of the electron beam obliquity on electron beam depth dose and shows the difference between the normal electron beam incidence and oblique beam incidence.

Figure 4.25 (a) shows the contribution of three pencil beams at normal incidence to two points at the central axis at different depths. At a point at shallow depth the dose to the point is due to the electrons from the central pencil beam. But at a point at greater depth, the neighboring pencil beams will contribute significantly to the dose.

Figure 4.25 (b) shows the effect oblique beam incidence. The adjacent pencil beams contribute to the dose to a point at shallow depth, while, a point at greater depth receives a smaller contribution. As a result of that, the dose of the oblique electron beam incidence is
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found to produce an increased dose at shallow depths and decreased dose at normal treatment depths.

Table 4.5 shows the 6 MeV electron beam parameters for single stationary and dual six fields measured at 350 cm SSD. It shows a difference between the data measured by Markus ionization chamber compared to the films; this is possibly due to some reasons. First, a slight fluctuation in the processing conditions of the film, such as temperature and purity of the developer fluid as well as the speed of the film processor. Second, the positioning accuracy of the ready pack film used due to the thickness of its envelope which makes it not easy to align the edge of the film with the edges. It is difficult to position the film at the same level of the phantom slices. Third, the effective depth of the chamber is at about 1 mm which requires moving the chamber about 1 mm towards the linac exit window. This can cause damage to the chamber window when the solid phantom slices are added at the surface of the chamber.

![Composite depth dose curves](image)

**Figure 4.23** composite depth dose curves for 6-MeV total skin electron therapy using the six-dual field technique without using a filter. Scanned at 0.2 mm steps.
Figure 4.24 Comparison between percentage depth dose curves for single dual fields and six dual fields.

Figure 4.25 A diagram shows the difference between depth dose of (a) normal electron beam incidence and; (b) oblique electron beam incidence [18].
Table 4.5 Beam parameters for both dual field stationary and dual six fields (rotational) measured at 350 cm source to surface distance (SSD).

4.7 Beam profile measurements:

Beam profile measurements were carried out along vertical and horizontal lines passing through the calibration point (0,0,0). Also, profiles were collected at off axis distances in the vertical and horizontal directions (see figure 4.30). The purpose of these measurements is to study the dose uniformity over a total height of 160 cm and width of 80 cm for a dual stationary angle field.

4.7.1 Beam profiles along the vertical axis:

Dose uniformity along the vertical direction was measured using the Scanditronix DPD-12pc diode system. EDD-2 electron diodes connected to the system have 2 mm water equivalent
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build up. The dependency of these diodes on the direction of the field is very low. Diodes numbered (1, 2, 3, and 4) were used in the measurement as shown in figure 4.26. They were mounted on the surface of the flat Perspex phantom. Four phantom slices were used each with size 2m x 1.5m x 1 cm. The phantom was positioned at a source to surface distance of 350 cm (SSD). Diodes were mounted along the vertical central axis starting from the superior end going down to the inferior end of the phantom. The HDTSe mode was used with a dose rate of 888 MU/min. Collimator setting at 100 cm SSD was 36 cm x 36 cm. dual stationary angle fields with angles 252.5° and 287.5° were used with 250 MU per field. Measurements were carried out to cover a distance of 170 cm in the vertical direction. The diodes were shifted down to the next set of positions after each of the dual stationary field exposures.

Results are shown in figure 4.27. It shows the beam profiles produced from each single field which is represented with dashed lines, while the summation of the two single beam profiles is represented by the solid line. The distance between the inferior edge of the upper field and superior edge of the lower field (the gap between the light fields) found to be 28 cm. In this region, no direct beam is available. The dose received in this gap is from the scatter radiation produced by the two angled fields. No normalization was used in figure 4.27, while the summation of both single beams in figure 4.28 is normalized to the calibration point (0,0,0) along the horizontal line at the surface of the phantom.
Figure 4.26 A schematic diagram of dose uniformity measurements at the surface of a Perspex phantom using DPD-12pc diodes.

Figure 4.27 Dose uniformity of dual stationary field along the vertical central axis (superior to inferior) using electron diodes. The profile is unnormalized.
From figure 4.28, it was found that the dose in the vertical direction from 85 cm to -85 cm (170 cm) is varying from 89% to 100%. The variation in the dose along the 160 cm line was found to be from 91% to 100%. If the results are renormalized to 95.5% then the dose uniformity will be within $\pm 5\%$ along 160 cm in the vertical direction.

**4.7.2 Beam profiles along the horizontal axis:**

Dose uniformity along the horizontal direction was measured using the Scanditronix DPD-12pc diode system. The electron diodes were mounted on the surface of the Perspex phantom at 350 cm source to surface distance (SSD) along the horizontal line to cover a distance from 40 cm to -40 cm (80 cm total distance). The HDTSe mode was used. Collimator setting at 100 cm SSD was 36 cm x 36 cm. dual stationary angle fields with angles 252.5° and 287.5° were
used with 250 MU per field. Diodes were moved from the target end (40 cm) to the gantry end (-40 cm) along the horizontal plane. The results are shown in figure 4.29.

![Graph showing beam profiles](image)

**Figure 4.29** Beam profiles measured along the horizontal central line at 350 cm SSD. The direction of measurements is from target to gantry. DPD-12pc electron diodes were used in the measurements.

By analyzing the results from figure 4.29 it was found that the dose variation was from 87% to 100% within 80 cm along the horizontal line while the variation within the 60 cm region was from 92% to 100%. If the results are renormalized to 96% then the variation in the dose uniformity will be within ± 4.2%.

### 4.7.3 Vertical off axis beam profiles:

Off axis beam profiles were measured using Scanditronix DPD-12pc electron diodes. Diodes were mounted on the surface of the Perspex phantom in the positions shown in figure 4.30. The source to surface distance (SSD) used was 350 cm. HDTSe mode was used with field size 36 cm x 36 cm at 100 cm SSD. Dual angles 252.2° and 287.5° were used with 250 MU per field. Vertical beam profiles were measured every 10 cm on the right and left side of the
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central axis at distances -40 cm, -30 cm, -20 cm, -10 cm, 10 cm, 20 cm, 30 cm, and 40 cm. The results are shown in figure 4.31.

4.7.4 Horizontal off axis beam profiles:

Off axis beam profiles were measured using electron diodes on the surface of Perspex plates at 350 cm SSD. The profiles were measured every 20 cm above and below the horizontal at distances 80 cm, 60 cm, 40 cm, 20 cm, 0 cm, -20 cm, -40 cm, -60 cm, -80 cm. The results are shown in figure 4.32.

![Diagram showing off axis beam profiles](image)

**Figure 4.30** A diagram shows the positions of the DPD-12pc diodes used to measure the dose uniformity along the vertical and horizontal direction.
Figure 4.31 Off-axis dose uniformity along the vertical direction (superior to inferior) using DPD-12pc electron diodes.

Figure 4.32 Off-axis dose uniformity along the horizontal direction (T to G) using electron diodes.
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The results of the dose distribution obtained from the vertical and horizontal profiles are shown in figures 4.33 and 4.34.

Figure 4.33 Dose distribution diagram using stationary dual stationary field at 350 cm SSD.
Figure 4.34 Percentage of doses measured by electron diodes using dual stationary angle field 252.5° and 287.5° without using beam modifiers.

Figure 4.34 shows the dose distribution measured at 350 cm with a dual stationary field. This dose distribution is done with a stationary field on a flat surface, but when the patient is placed in all six positions, with a dual field irradiation at each position, the depth dose is less uniform than indicated in figures 4.33 and 4.34 due to body curvature, the varied angles of electron incidence and the finite number of beam orientations [67].

4.8 Summary

Beam parameters for the TSET open beam were measured in this chapter. The results of the measurements are summarized in this section as follows:

- The dual angle for the measurement was found to be 17.5° (252.5° and 287.5°) above and below the horizontal. This angle was considered to be acceptable because the specifications with respect to uniformity are within the desirable range of limits for
this type of treatment. The dose received in the gap between the two fields in the vertical direction is due to change in the direction of the electrons as a result of elastic nuclear collisions. A series of collisions, known as multiple scattering, causes electron trajectories to deviate from initial directions. Another reason is the electron backscattering.

- The average x-ray contamination can be reduced by angling the beam axes so that the regions of maximum x-ray intensity lie outside the body. The x-ray contamination was reduced from 1.5% with the single horizontal beam at 270° to 1% of the maximum dose with the dual stationary field.

- It is important to shield the ionization chamber cable when the measurements are carried out. The main contribution of the extracameral signal comes from the adaptor connecting the chamber's cable with the extension cable. The percentage of extracameral signal for a 210 cm cable length is about 5.5% of the total signal. This can be reduced by shielding the cable and the connection using a steel pipe of 1 cm thickness. The measured extracameral signal at the adapter was reduced from 2.5% to about 1% using the steel pipe.

- Central axis depth dose curves were measured for a single electron beam at 100 cm SSD, a single horizontal beam at 350 cm SSD, and for dual stationary beams at 350 cm SSD. The results are recorded in table 4.3. The x-ray contamination was found to be slightly smaller for the dual stationary field (~1%) than the single horizontal field (~1.5%). The percentage x-ray contamination is increasing with the increase of the source to surface distance. At 100 cm SSD the x-ray contamination measured 0.5% while at 350 cm it was 1.5%.
• Off axis depth dose curves were measured using films. The dose at the surface was found to vary slightly from 83% to 86% of $D_{\text{max}}$ from the reference point (0,0,0) to 60 cm vertical distance above the reference point, while, it drops to 79% at 80 cm distance. Also, a slight increase in the x-ray contamination from 0.6% to 1% was caused by the increase in the off axis distance in the vertical direction. Another feature found from the off axis depth dose curves, was the shift of the maximum absorbed dose towards the surface of the phantom. This effect is due to the energy degradation caused by the increase in the source to surface distances. It might also be caused by the increase obliquity.

• The beam penetration using the dual six fields was changed compared to the dual stationary field. This is slightly due to the change in the source to surface distance (SSD). However, it depends significantly on the oblique incidence of the electron beam.

• The beam parameters obtained from the dual six fields are shown in table 4.5. The maximum dose was found on the skin. It falls to 90% of the dose at 1.5 mm, and to 10% at 22 mm. The recommendations regarding the TSET beam mentioned in section 2.2 require that the maximum dose be at the surface, the 90% from about 3 mm to 5 mm, and the 10% at about 20 mm. Therefore, the measured data almost meet the recommendations.

• Beam profiles were measured along the vertical and horizontal direction. It was found that the dose uniformity along the vertical direction varies within $\pm 5\%$ along 160 cm distance (from -80 cm to 80 cm). The variation along the horizontal direction was
found to be just about $\pm 4.2\%$ along 60 cm distance (from -30 cm to 30 cm). The results of the measured uniformity were found to meet the beam requirements mentioned before in section 2.2. It was recommended in the AAPM report 23 [67] that the dose uniformity to be $\pm 8\%$ in the vertical direction and $\pm 4\%$ in the horizontal direction over the central 160 cm x 60 cm area of the treatment plane.

In the next chapter measurements carried out to study the effect of the beam modifiers on the dual stationary field, and the dual six fields, are repeated.
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Measurements of Total Skin Electron Fields Using Beam Modifiers

5.1 Introduction:

The measured beam parameters of non modified electron fields were reported in the previous chapter. Measurements were carried out for a single electron field at 100 cm SSD, a single horizontal field at 350 cm SSD, a dual stationary field at 350 cm SSD, and six dual fields at 350 cm SSD. The results of the measurements showed that the single horizontal beam and the dual stationary beam at 350 cm SSD are more penetrating than the dual six beams at the same distance. The dose distribution measured was very acceptable and meets the beam requirements recommended by the AAPM report number 23 [67].

In this chapter, measurements were carried out to study the effects of beam modifiers on the Varian 2100C 6 MeV HDTSe beam characteristics are reported. A comparison between the beam characteristics with and without the beam modifiers is presented.

5.2 Beam modifiers used by other centers:

Beam modifiers with thick materials can be used to reduce the electron energy “energy degraders” or “decelerators”, see figure 5.1. Thin materials used primarily to spread out the beam are called “beam scatterers”. Interposed material scatters the incident electron beam, reduces its energy, and generates a contaminant megavoltage x-ray background [67]. All three effects occur in a given material, but their relative magnitudes depend on the atomic number, Z, of the material, and the energy of the electron beam. It is important to understand that scatterers also act as degraders, and
vice versa, even when a material of appropriate Z is used. Beam modifiers are placed inside the linear accelerator head or externally, either at the surface of linear head or at a distance between the head of the linear accelerator and the patient.

Studies have been carried out by several centers to improve the beam uniformity using scattering filters. Some centers used large Lucite scatterer panels about 1 cm thick and 2 m x 1 m in cross section [17] placed at about 20 cm in front of the patient. This method improved the dose uniformity but reduced the penetration of the beam (beam degrader).

Brahme [10] investigated the effect of placing the scattering filter near the exit window of the linear accelerator or at the surface of the phantom. He found that the angular distribution was completely different between both methods. It was found that when the scatterer is placed near the exit window the electrons have narrower angular spread compared to those produced when the scatterer is placed at the surface of the phantom. The wider angular spread of the electrons causes a higher surface dose and less penetration.

At the Massachusetts Institute of Technology [42][55], Al foils were used as scatterer placed near the vacuum window of the linear accelerator drift tube. The electrons were directed through a conical collimator having a slit of 1 cm x 45 cm at its base close to the patient skin. A Gaussian distribution of intensity across width of the cone was measured with $\pm$ 10% variation at transverse plane 118 cm from the electron window. It was also found that the patient dose varies as much as $\pm$ 15% due to the change in the distance between the cone and the skin during treatment.
El-khatib et al [19][21] opened the collimator to the maximum size 40 cm x 40 cm without using the electron trimmer. The field size at 400 cm source to surface distance was 160 cm x 160 cm and the distance along the diagonal was 226 cm. By rotating the collimator to 45° the patient could be placed along the diagonal, which is large enough to encompass most of the patient. The dose uniformity was improved by using a Plexiglass flattening filter consisting of a 14.4 cm square plate of 0.18 cm thickness. More plates were added concentrically with different sizes and thicknesses, 9.6 cm square and 0.32 cm thick, 4.8 cm square and 0.42 cm thick, and 2.4 cm square and 0.45 cm thick. The filter was mounted on the linear accelerator head. The dose uniformity produced was ± 8% over a distance of 160 cm along the vertical direction, and ± 4% over a distance of 60 cm along the horizontal direction.

Tetenes and Goodwin [77] used a shaped polystyrene scatterer beam flattening filter. The filter was mounted on the front of the treatment head with a distance of 7 m between the accelerator exit window and the treatment plane. The variation in the dose uniformity of the produced beam was found to be ± 8% for a 200 cm diameter circle.

5.3 Inherent filtration used in Varian 2100C linear accelerator:

The aim in this section is to study the effect of the scattering filters on the characteristics of the 6 MeV electron beam, and to design a scattering flattening filter to improve the dose uniformity and satisfy the beam requirements recommended by the AAPM report 23 [67].
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The linear accelerator used in the project is a Varian 2100C. A description of the major components and their function was given in section 3.2. It has two scattering foil systems used for all energies. The scattering foil system used with the 6 MeV electrons consists of a primary scattering foil which is designed to produce a beam which has a lateral distribution similar to a Gaussian shape. The material used to make the scattering foil has a high atomic number and has a uniform thickness. The scattering foil is placed at 9.5 cm from the x-ray target and it consists of a 0.075 mm thick disc of tantalum.

The second scattering foil system is used to improve the flatness of the central part of the electron beam. The thickness of the second foil is not uniform like the first one; this is to modify the distribution at the peak of the profile curve. The material used in this scattering foil has a low atomic number to reduce the energy losses. It is located at 13 cm from the x-ray target. It consists of two materials, a 0.15 mm thick disc of aluminium and a 0.025 mm thick tantalum disc of 4 mm diameter placed concentrically.

5.4 Scattering filter materials used in the project:

The purpose of using the scattering filters is to modify the beam profile shape to improve the dose uniformity over the area of interest. In order to achieve this, different types of materials were used to make scattering filters. Two different types of readily available materials were used in this project. The first type was Perspex which is available with 2.5 mm thicknesses. The second type of scattering filter was developed unexposed x-ray films.
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It was explained before in section 2.7 that the stopping power results from collisional and radiative energy loss. Collisional stopping power is higher with low atomic number (Z) materials, because in low Z material electrons are more weakly bound and therefore they are more available for these interactions. However, in high Z materials, the electrons are more tightly bound which allows fewer interactions with incident electrons.

On the other hand, radiative stopping power increases with atomic number (Z) and energy. The electron interactions with high Z materials generate more bremsstrahlung, i.e. contribute more x-ray contamination. The x-ray contamination is caused by the interaction between the electrons and the exit window of the accelerator, scattering foil, ion chambers, collimator jaws, air, external scattering filters, and patient.

In a standard electron beam treatment, the x-ray contamination is small, about 1% [52]. However, in the total skin electron therapy the dose from the x-ray contamination can be about 10% to 15% [50] of the prescribed electron dose. This is due to the multiple fields used in this type of treatment.

The data provided by Berger and Seltzer [7] of the stopping powers show that Perspex has a high collisional stopping power and a low mass scattering power. Table 5.1 shows the collisional mass stopping power and total mass stopping power as a function of energy for Perspex.
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### Stopping Power

<table>
<thead>
<tr>
<th>Energy MeV</th>
<th>Collision MeV(\text{cm}^2/\text{gm})</th>
<th>Radiative MeV(\text{cm}^2/\text{gm})</th>
<th>Total MeV(\text{cm}^2/\text{gm})</th>
<th>CSDA Range g/cm(^2)</th>
<th>Mass Scattering Power Radian(^2) cm(^2) g(^{-1})</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.0</td>
<td>1.788</td>
<td>0.011</td>
<td>1.799</td>
<td>0.450</td>
<td>2.600</td>
</tr>
<tr>
<td>1.5</td>
<td>1.760</td>
<td>0.017</td>
<td>1.776</td>
<td>0.731</td>
<td>1.400</td>
</tr>
<tr>
<td>2.0</td>
<td>1.762</td>
<td>0.023</td>
<td>1.785</td>
<td>1.012</td>
<td>0.898</td>
</tr>
<tr>
<td>3.0</td>
<td>1.784</td>
<td>0.038</td>
<td>1.822</td>
<td>1.567</td>
<td>0.470</td>
</tr>
<tr>
<td>4.0</td>
<td>1.809</td>
<td>0.053</td>
<td>1.862</td>
<td>2.109</td>
<td>0.292</td>
</tr>
<tr>
<td>5.0</td>
<td>1.832</td>
<td>0.069</td>
<td>1.901</td>
<td>2.641</td>
<td>0.201</td>
</tr>
<tr>
<td>6.0</td>
<td>1.851</td>
<td>0.087</td>
<td>1.938</td>
<td>3.162</td>
<td>0.147</td>
</tr>
<tr>
<td>7.0</td>
<td>1.868</td>
<td>0.104</td>
<td>1.972</td>
<td>3.673</td>
<td>0.118</td>
</tr>
</tbody>
</table>

**Table 5.1** Collisional mass stopping power and total mass stopping power as a function of energy for Perspex [8].

The Perspex is a good material to be used for energy degradation. Table 5.1 shows that the Perspex which is mainly composed of carbon, oxygen and hydrogen, has a high collisional stopping power and a low radiative stopping power. Figure 5.1 shows the effect of 6 mm and 10 mm thicknesses of Perspex [trays] on the 6 MeV electron beam penetration for a single horizontal field at 350 cm SSD, using the HDTSe mode. The maximum absorbed dose for the non filtered beam is at 12 mm, shifted to 10 mm with 6 mm Perspex, and to 7 mm with 10 mm Perspex.

In figure 5.2 dose uniformity was measured along the horizontal plane using three different setups, without beam modifiers, with 2.5 mm Perspex, and with 5 mm Perspex. The size of the Perspex sheets was 23.5 cm x 7 cm. The filter was mounted on the HDTSe applicator along the vertical direction and the scan was carried out along the horizontal plane. The of dose was normalized to the dose at the calibration point (0,0,0) at the surface of a Perspex phantom with a non-modified field (open).
The produced profiles do curve up rather than down at the edges. The uniformity of the beam was not expected to improve along the vertical direction with this type of beam filtering, because the thickness of the Perspex is not changing along the vertical direction to compensate to the variation in the measured dose. Figure 5.1 shows the effect of filter thickness on the beam degradation. This is why Perspex can not be used in the project. The minimum available thickness available is 2.5 mm. To be used as a scatterer very thin Perspex sheets are required. This is hard to obtain in the local market.

**Figure 5.1** Beam modifiers effect on the penetration of the electron beam. The surface electron energy was reduced from 5.1 MeV to 4.1 MeV with 6 mm Perspex tray, and to 3.5 MeV with 10 mm Perspex tray.
The other material used as a beam modifier was developed non-exposed films. Kodak X-Omat V film was used. The film was developed without getting any exposure. The aim was to remove the silver component from the film. The silver has a high atomic number \((Z = 47)\). The remaining components left in the film are mainly Hydrogen, Carbon, and Oxygen. The electrons in the high Z materials are more tightly bound which reduces interactions with the incident electrons. The effective atomic number of the film becomes lower after removing the silver. With low Z materials, the collisional cross section is high and the radiative cross section is low. The low radiative cross section reduces the x-ray contamination produced when the electrons pass through the x-ray films. One other important advantage of using the films is their
5.5 Scattering filters using films:

The TSET technique used in the project is a modification of the well-known Stanford technique \[66\], which has two beams for each of six patient positions. The modification will be the addition of the electron scattering filter. The angles used were $287.5^\circ$ and $252.5^\circ$ at a distance of 350 cm SSD. The gap between the inferior edge of the upper projected light field and the superior edge of the lower projected light field was measured to be 28 cm at the surface of the Perspex phantom. It is not necessary that the projection of the electron beam will coincide exactly with the projection of the light field at the treatment plane. This was explained in section 2.3.4.

The aim when designing a scattering filter is to reduce scattered electrons in the area from -40 cm to 40 cm as shown in figure 5.3. This will improve the dose uniformity along the vertical direction. Figure 5.3 schematically shows which area of the beam profile of 6 MeV total skin electron beam will be filtered.

The use of the scattering filter in figure 5.3 would shift the beam profile around the center down and push it up around the sides. The x-ray film was developed then it was cut into strips of the same size (24 cm x 4.5 cm). Two strips were used as a scattering filter. One strip was added on the top of the other. The filter was then mounted on the HDTSe applicator as shown in figure 5.4.
Figure 5.3 A schematic diagram to show the function of the scattering filter on a vertical beam profile at 350 cm using dual stationary field.

Figure 5.4 Left image shows strips of developed non-exposed x-ray film used as a scattering filter mounted on the HDTSe applicator. The right image shows the position of the filter on the head of the linear accelerator. The collimator is rotated to 90°.
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At the beginning, no collimator rotation was used. The gantry is rotated to 287.5° then the film was mounted close to the upper edge of the HDTSe applicator. Then when the gantry is rotated to 252.5°, the film is moved close to the lower edge of the HDTSe applicator. This method was not good in practice because it requires the physicist or the technician to enter the room to change the position of the film after every dual angle.

This problem was solved by using collimator rotation. The collimator was rotated to 90° and the gantry to 287.5° then the film was mounted close to the upper edge of the HDTSe applicator. After giving the exposure the gantry is rotated to 252.5° and the collimator is rotated to 270°, so the film is then moved to the lower edge of the applicator (see figure 5.5).

The projection of the filtered beam is shown in figure 5.5. The scattering filter was positioned using the projection of the light field on the Perspex phantom. It was required that the projection of the light field filtered starting from the two edges of the 28 cm gap for both dual angle fields. When the gantry is moved to 252.5°, the collimator is rotated to 270° and the scattering filter will be close to the lower edge of the collimator, and with the gantry at 287.5° the collimator is rotated to 90°, the scattering filter then is close to upper edge of the collimator.
5.6 Beam profile measurements along the vertical plane using scattering filter:

The first set of measurements was carried out without rotating the collimator to 90° and 270°. Scanditronix DPD-12pc electron diode system was used in the measurements. The diodes were mounted on the surface of the Perspex phantom at 350 cm SSD. It was mentioned before that the electron diodes have an inherent filtration of 2 mm water equivalent. Measurements were carried out along the vertical
plane. The results of dose uniformity without the collimator rotation are shown in figure 5.6. The profiles in the figure are not normalized.

Figure 5.7 shows the beam profile along the vertical plane normalized to the calibration point at the surface of the Perspex phantom at (0,0,0). The dose was found to vary from 98% to 103% along 170 cm distance and from 99% to 103% along 160 cm distance in the vertical plane, so the dose uniformity over 160 cm distance is ±2%. The dose uniformity along the vertical direction was improved with the use of the scattering filter compared to the non-modified beam (±5% over 160 cm).

This method is not practical. It will require the radiotherapy technician to enter the room after every given field to change the position of the scattering filter. This will require more treatment time and it is not comfortable to the patient and to the radiotherapy technician. This is why the measurements in this project will be carried out using collimator rotation.

The beam flatness measurement along the vertical plane was repeated with two collimator angles; 90° and 270°. The scattering filter was mounted on the HDTSe applicator. The first field was at 287.5° gantry angle with collimator rotation 90°. The second one was at 252.5° with collimator angle 270°. The results of the measurement are shown in figure 5.8. The dose with collimator rotation was found to vary from 93% to 100% along 170 cm, and from 94% to 100% or ±3% over 160 cm distance.
Figure 5.6 Beam profile in the vertical treatment plane using scattering filter. Collimator was at 0°. No normalization was used.

Figure 5.7 Field flatness in the treatment vertical plane using scattering filters measured with collimator at 0°.
Figure 5.8 Field flatness in the treatment vertical plane with scattering filters measured with collimator at 90° and at 270°.

A comparison between the two beam profiles produced with and without collimator rotation is shown in figure 5.9. The beam uniformity was changed slightly about 1% from ± 2% without a collimator rotation to ± 3% with collimator rotation. The possible reason for this could be the slight shift in the jaws (Y2 and Y1) when the collimator is rotated due to its heavy weight and the gravity when the gantry is at lateral position. The beam uniformity without rotating the collimator is slightly better than the uniformity with collimator rotation.

Another comparison is shown in figure 5.10 between the beam profile along the vertical plane with and without the use of the scattering filter. The beam uniformity along the vertical plane was improved by applying the scattering filter. A summary of the beam uniformity data is shown in table 5.2.
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<table>
<thead>
<tr>
<th></th>
<th>Without filter</th>
<th>With filter Collimator at 0°</th>
<th>With filter Collimator at 90° and 270°</th>
</tr>
</thead>
<tbody>
<tr>
<td>160 cm</td>
<td>± 5%</td>
<td>± 2%</td>
<td>± 3%</td>
</tr>
</tbody>
</table>

**Table 5.2** Results of beam uniformity measurements with and without scattering filter.

**Figure 5.9** Comparison between the beam flatness using scattering filter with and without collimator rotation.

**Figure 5.10** Comparison between the beam flatness along vertical plane with and without scattering filter.
5.7 Beam profile measurements along the horizontal plane using scattering filter:

Dose uniformity was measured along the horizontal direction for a dual stationary field with collimator angles at 90° and 270° as described before. The results are shown in figure 5.11. The beam flatness along the horizontal plane is normalized to the calibration point at the surface of the Perspex phantom at (0,0,0). The dose along the horizontal plane was found to vary from 92% to 100% or ±4% over 60 cm and from 96% to 100% over a 40 cm region.

A comparison between the field flatness with and without the scattering filter along the horizontal plane is shown in figure 5.12. It was found that the beam flatness did not improve along the horizontal plane when the scattering filter was applied. The field flatness is the same with and without the scattering filter.

![Figure 5.11](image)

**Figure 5.11** Field flatness in the horizontal treatment plane normalized to a calibration point at the surface of a Perspex phantom at (0,0).
Figure 5.12 Comparison between the beam flatness along the horizontal plane with and without scattering filter.

5.8 Vertical off axis beam profiles with scattering filter:

Off axis beam profiles along the vertical direction were measured with the electron diodes system at the surface of the Perspex phantom. The profiles were measured at off axis distances -40, -30, -20,-10, 10, 20, 30, and 40 cm on the left side and right side of the vertical central line. The results are shown in figure 5.13.
5.9 Horizontal off axis beam profiles with scattering filter:

Off axis beam profiles along the horizontal direction were measured using the electron diodes system. The profiles were measured every 20 cm above and below the horizontal at distances 80 cm, 60 cm, 40 cm, 20 cm, 0 cm, -20 cm, -40 cm, -60 cm, -80 cm. The results are shown in figure 5.14.
The dose distribution for a dual stationary field within an area of 160 cm x 80 cm is shown in figure 5.15. It also shows the difference between the dose distribution with and without the scattering filter. The beam uniformity was improved using the scattering filter. The difference in the percentage of doses at the surface of Perspex phantom is shown in figure 5.16.

These results are for a stationary dual stationary field. Some other measurements will be described later which measured the dose distribution on the surface of RANDO phantom using dual six fields.
Figure 5.15 Dose distribution diagram using dual stationary field at 350 cm SSD without scattering filter (top), and with scattering filter (bottom).
Figure 5.16 Percentage of doses without scattering filter (top), and with scattering filter (bottom), measured at the surface of Perspex phantom.
5.10 Percentage Depth Dose with Scattering Filter:

The percentage depth dose curve along the central line was measured using a Markus ionization chamber in the solid phantom. Figure 5.17 shows the percentage depth dose curve for a dual stationary field with scattering filter at a TSET treatment distance of 350 cm SSD. The most probable energy, $E_p$, was calculated using equation 3.1. $R_p$ and $R_{50}$ were estimated from the depth dose curve to be 25.7 mm and 19.8 mm respectively. The $E_p$ was found to be:

$$E_p = 0.22 + 1.98 \times (2.57) + 0.0025 \times (2.57)^2$$

$$E_p = 5.3 \text{ MeV}$$

The mean electron energy at the surface, $\bar{E}$, is calculated using equation 4.4. The $\bar{E}$ was found to be:

$$\bar{E} = 0.656 + 2.059 \times (1.98) + 0.022 \times (1.98)^2$$

$$\bar{E} = 4.8 \text{ MeV}$$

Using the IPEMB code of practice for electron dosimetry for radiotherapy beams (1996) [41] the following beam parameters were found:

$$D_s = 81 \%$$

$$d_{max} = 10 \text{ mm}$$

$$R_{50,D} = 20 \text{ mm}$$

$$R_p = 26 \text{ mm}$$

$$D_x = 1.7 \%$$

Figure 5.18 shows a comparison of the depth dose curve with and without the scattering filter. It shows also the effect of the scattering filter on the x-ray contamination component.
Figure 5.17 Percentage depth dose curve for a dual stationary field with scattering filter at TSET treatment distance of 350 cm SSD.

Figure 5.18 The variation of percentage depth dose with and without scattering filter.
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The important features shown from the comparison between percentage depth dose curves with and without scattering filter in figure 5.17 and figure 5.18 are:

- The surface dose was increased from 78% to 81% with the scattering filter.
- The maximum absorbed dose is shifting slightly towards the surface with scattering filter. The depth of maximum dose was reduced from 11 mm to 10 mm with the scattering filter.
- A slight degradation in the mean energy at the surface from 4.9 MeV to 4.8 MeV.
- X-ray contamination was increased from 1.0% to 1.7% with the scattering filter.

5.11 Off Axis percentage depth doses with scattering filter:

The setup for this measurement was described before in section 4.5. Figure 4.16 shows how the Kodak X-Omat V films were sandwiched between the solid phantom slices along the vertical plane. The percentage depth dose curves produced after scanning the films are shown in figure 5.19. All the curves were normalized to the maximum absorbed dose along the central line on the 0 cm off axis film.

There are some features which can be seen from the percentage depth dose curves when the off axis distances increases:

- The percentage surface dose is increasing slightly from about 81% at the center to about 85% at 80 cm off axis distance. However, it was found in
chapter 4 that for a non-modified electron beam the percentage of surface dose is decreasing at 80 cm off axis distance.

- The depth of maximum absorbed dose ($D_{\text{max}}$) is shifted slightly towards the surface. Maximum absorbed dose was found at depth 9 mm for the central film, while it shifts to 7 mm at 80 cm off axis distance. This is possibly due to the energy degradation when the source to surface distance was increased. It was mentioned also that the scattering filter reduced the depth of maximum dose from 11 mm to 10 mm along the central line according to the Markus ionization chamber measurements.

- An expected difference in the dosimetry between the Markus chamber and the films was found of about 1 mm. The reasons for this difference were explained in section 4.6 in the previous chapter.

- The x-ray contamination is increasing with the increase of the off axis distance to about 2%, while it was increased with the non-modified electron beam to about 1%.

### 5.12 Depth dose curve using dual six fields with scattering filter:

This measurement is similar to the measurement mentioned in section 4.6. It was carried out in the humanoid phantom (RANDO). Kodak X-Omat V film was cut to fit between slices number 20 and 21 in the phantom. The film was positioned along the central horizontal line at a distance of 350 cm SSD. Dual six fields were used to expose the film with 200 MU each. The film was then developed and scanned by the RFA Densitometer. The percentage depth dose curve produced is shown in figure 5.20. This curve was produced from one location which is through the horizontal line through the anterior side of the phantom.
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Figure 5.19 Percentage depth dose curves for dual stationary field with scattering filter. X-ray contamination component is also shown.

Figure 5.21 shows the difference between the percentage depth dose for a stationary field with scattering filter and dual six fields with scattering filter. The rapid fall off of dose with dual six fields compared to the stationary field is shown. The beam from the stationary field is too penetrating. The surface dose from the dual six fields is about 97% and the depth of maximum dose is at 0.2 mm. The reasons for the change in beam characteristics with the dual six fields were explained in section 4.6. The change depends mainly on the oblique electron beam incidence and slightly on the change of SSD.

Figure 5.22 compares the percentage depth dose for the dual six fields with and without scattering filter. The features shown the in the figure are:

- Percentage surface dose was increased with the scattering filter from 92% to 97%.
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- Depth of maximum absorbed dose was reduced from 0.4 mm to 0.2 mm.
- A slight increase in the x-ray contamination with the scattering filter from 1.3% to 1.5%.

The beam characteristics parameters for dual six fields with and without the scattering filter are summarized in table 5.3.

Figure 5.20 Composite depth dose curve for 6-MeV total skin electron therapy using the six-dual field technique with scattering filter.

Figure 5.21 Percentage depth dose comparisons between dual stationary field and six dual fields.
Figure 5.22 Percentage depth dose comparisons for dual six fields with and without scattering filter.

<table>
<thead>
<tr>
<th>Depth parameters</th>
<th>Without scattering filter</th>
<th>With scattering filter</th>
</tr>
</thead>
<tbody>
<tr>
<td>$E_p$</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>$\bar{E}$</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>$D_\text{s}$</td>
<td>92 %</td>
<td>97%</td>
</tr>
<tr>
<td>90% (proximal edge)</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>95% (proximal edge)</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>$d_{\text{max}}$ (100%)</td>
<td>0 mm</td>
<td>0 mm</td>
</tr>
<tr>
<td>95% (distal edge)</td>
<td>1 mm</td>
<td>1.5</td>
</tr>
<tr>
<td>90% (distal edge)</td>
<td>1.5 mm</td>
<td>3 mm</td>
</tr>
<tr>
<td>80% (distal edge)</td>
<td>3 mm</td>
<td>4 mm</td>
</tr>
<tr>
<td>50%</td>
<td>10 mm</td>
<td>11 mm</td>
</tr>
<tr>
<td>$D_e$</td>
<td>1.3 %</td>
<td>2%</td>
</tr>
</tbody>
</table>

Table 5.3 Beam characteristics of dual six fields with and without scattering filter.
5.13 Surface dose measurements for dual six fields

In this section the surface dose distribution of dual six fields with scattering filter for a 6 MeV electron beam was measured with TLDs. These measurements were used to check how much dose is received by the phantom circumference. The phantom used in the measurements was the humanoid phantom (RANDO).

A set of 49 TLDs was calibrated at source to surface distance of 100 cm SSD, for a 6 MeV electron beam, using the 10 cm x 10 cm electron applicator, at depth of maximum absorbed dose ($d_{\text{max}} = 15$ mm). The dose given to the TLDs was 100 cGy. After exposing the TLDs they were pre-annealed, then the TLD reader was used to read the response of each TLD. An individual calibration factor (ICF) was obtained for each TLD chip.

The calibration for the TLDs was repeated three times. The individual calibration factor for each TLD was taken as the mean from the three calibration measurements. The results from the TLDs readings were within $\pm 1\%$. These calibration factors were saved in an EXCEL sheet to be used later in measurements of the surface dose distribution.

The surface dose distribution measurements were carried out using the RANDO anthropomorphic phantom. The surface curvature of this phantom is not constant and this causes variation in the dose. It is not necessary that regions on the phantom closer to the source will receive higher doses than the regions at longer distances; regions further away like the head will receive higher doses for the dual six fields due to the small diameter of the head. It is also important to know that RANDO is a rigid body
phantom only, without arms and legs. On the real patient, greater dose variation would be expected. Kumar et al. found that the variation of dose in the treatment plane might increase to ± 15% at the patient because of the variable skin distance, and the self shielding [55][56]. The measurements were carried out with the phantom positioned at a distance of 350 cm SSD. TLDs were positioned on the surface of the phantom on the anterior and posterior side. The electron scattering filter was used. The result of the skin dose distribution on the anterior side is shown in figure 5.23. The distribution of the skin dose on the posterior side is shown in figure 5.24.

The anterior view is showing a variation of skin dose from 95% to 107% which is about ± 6.3%. The posterior view is showing a variation from 96% to 107% which is about ± 5.7%. This variation is due to the surface curvature of the RANDO phantom.

**Figure 5.23** Anterior view of the skin dose distribution for the dual six fields with scattering filter using RANDO phantom.
Figure 5.24 Posterior view of the skin dose distribution for the dual six fields with scattering filter using RANDO phantom.

5.14 Absolute Dose Measurements:

It was recommended by the AAPM that TSET absorbed dose be measured at a calibration point located at the surface of the phantom and on the horizontal axis [67]. This dose is called the calibration point dose.

Another term defined given by the AAPM is the treatment skin dose. This is defined as the mean dose along a circle at or near the surface of a cylindrical polystyrene phantom 30 cm in diameter and 30 cm high which has been irradiated as a hypothetical patient with all dual six fields. This dose lies at or very close to the skin surface. The calibration point dose is then related to the treatment skin dose by
multiplication with factor $B$, which is the ratio of the calibration point dose from a
dual stationary field to the treatment skin dose from dual six fields. $B$ is in the range
2.5 to 3.1. Thus $B$ is the factor relating the skin dose with the calibration point dose,
both measured at the surface of a cylindrical polystyrene phantom. Since dose
measurements can be made at depth, say $d_{\text{max}}$ (at 10 mm) in the reference plane for a
dual field with less uncertainty than at the surface, in this project the $B$ factor was
redefined as the ratio of mean circumferential surface dose from the dual six fields to
the dose at 10 mm deep from a dual stationary field. A cylinder of diameter 27 cm and
height 30 cm was made from paraffin wax for this measurement. Kodak X-Omat V
films were wrapped around the perimeter of the cylinder. 10 mm of buildup material
was used on the top of the film. The second film was wrapped around on the top of
the bolus material, so that the optical density can be measured at the surface as well as
at 10 mm depth. Two sets of exposures were used. First set used only a dual stationary
field and the second set used dual six fields, rotating the phantom at 60 degrees
intervals between dual fields. 235 MU per field was used to give about 50 cGy at 10
mm deep from the dual stationary field. Same exposure was used for the dual six
fields. It was found that the average surface dose from the dual six fields is 2.6 times
the dose at the calibration reference depth (10 mm) from dual stationary field. This
build up $B$ factor will vary with the diameter of the phantom used. It will increase
when the diameter of the phantom is smaller due to the overlapping of the fields.

Calibration for the TSET 6 MeV electron beam with scattering filter using the IPEMB
code of practice for electron dosimetry for radiotherapy beams (1996) [41] was
carried out. The Ionex Dosemaster electrometer was used with the Markus parallel

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plate ionization chamber in a solid phantom. Measurements were carried out using dual stationary field with the chamber at 10 mm depth.

The absorbed dose to water is given by:

\[ D_{w} (Z_{ref,w}) = M_{ch,w}^{e} * N_{D,air,ch} * S_{w/air}^{e} (\bar{E}, Z_{ref,w})P_{ch}^{e} (\bar{E}_{z}) \] (5.1)

Where \(M_{ch,w}^{e}\) is the corrected chamber reading, \(N_{D,air,ch}\) is the absorbed dose to air chamber factor, \(S_{w/air}^{e} (\bar{E}, Z_{ref,w})\) is the appropriate water to air stopping power ratio for an electron beam of mean surface energy \(\bar{E}\) at depth in water \(Z_{ref,w}\); these data are given in the protocol in table 5 and \(P_{ch}^{e} (\bar{E}_{z})\) is the perturbation factor applicable to the particular ionization chamber in an electron beam at the mean energy \(\bar{E}_{z}\) at depth \(Z_{ref,w}\). The \(P_{ch}^{e} (\bar{E}_{z})\) values for Markus chamber are given in table 2 in the protocol.

Correction for the instrument reading includes:

- The pressure, temperature and humidity correction factor can be calculated using:

\[ f_{T,P} = \frac{(273.15 + T)(1013.25)}{(293.15 * P)} \] (5.2)

- Polarity correction factor is given by

\[ f_{Pol} = \frac{(|M^{+}| + |M^{-}|)}{2M} \] (5.3)

where superscripts + or – indicate the reading (M) with collecting voltage positive and negative respectively and M in the denominator is the reading taken with normal polarity used during measurements.
• Ion recombination correction factor can be measured using the two voltage technique. For small corrections \((P_{\text{ion}} - 1) < 0.05\), the following equation can be used to calculate ion recombination correction factor \(P_{\text{ion}}\):

\[
P_{\text{ion}} - 1 = \frac{(M_1/M_2 - 1)}{(V_1/V_2 - 1)}
\]

Where \(V_1\) is the normal collecting voltage, \(M_1\) is the reading measured using \(V_1\), \(V_2\) is a lower voltage, and \(M_2\) is the reading at \(V_2\).

The Markus parallel plate ionization chamber and NE-Ionex Dosemaster electrometer were used with solid phantom. The chamber was positioned at 10 mm depth in the phantom at 350 cm SSD. The HDTSe applicator was used with collimator setting 36 cm x 36 cm at the isocenter. A dual stationary field was used in the measurement with scattering filter mounted on the HDTSe applicator. Steel pipe was used to shield the cable of Markus chamber. 500 MU was used. Different voltages were used in the measurement. The normal collecting voltage is -250 V. The results of measured readings in nC with different voltages are shown in table 5.4. Using voltages -250 V, +250 V, and -100 V, polarity and ion recombination correction factors were calculated using equations 5.3 and 5.4.

\[
f_{\text{Pol}} = \frac{(1.92975 + 1.8285)}{(2 \times 1.92975)}
\]

\(f_{\text{Pol}} = 0.974\)

\[
P_{\text{ion}} = \frac{(1.92975 / 1.9217 - 1)}{(250 / 100 - 1) + 1}
\]

\(P_{\text{ion}} = 1.0028\)
Table 5.4 Readings collected using different collecting voltages. Gantry at 270°.

<table>
<thead>
<tr>
<th>Voltage (V)</th>
<th>Reading (nC)</th>
</tr>
</thead>
<tbody>
<tr>
<td>300</td>
<td>1.8316</td>
</tr>
<tr>
<td>250</td>
<td>1.8285</td>
</tr>
<tr>
<td>200</td>
<td>1.8278</td>
</tr>
<tr>
<td>100</td>
<td>1.8225</td>
</tr>
<tr>
<td>-100</td>
<td>1.9217</td>
</tr>
<tr>
<td>-200</td>
<td>1.9293</td>
</tr>
<tr>
<td>-250</td>
<td>1.92975</td>
</tr>
<tr>
<td>-300</td>
<td>1.9325</td>
</tr>
</tbody>
</table>

Temperature (T) = 22.2°C
Pressure (P) = 982.9 mbar

Using equation 5.2, temperature and pressure correction was found to be:

$$f_{T,P} = 1.0386$$

From section 5.10 the mean electron energy at the surface ($\overline{E}$) was found to be 4.8 MeV.

$$Z/R_{p} = 10 \text{ mm} / 26 \text{ mm} = 0.39$$

The ratio of mean energy at depth $Z$, $E_Z$, and the mean energy at the surface of the phantom, $\overline{E}$, is given in table 3 in the IPEMB electron dosimetry code of practice (1996).

$$E_Z / \overline{E} = 0.5392, E_Z = 2.59 \text{ MeV}$$
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The perturbation factor, $P_{\text{Markus}}$, for the Markus parallel plate chamber at the reference depth $Z = 10\text{mm}$ can be found from IPEMB code (1996) in table 2.  
$P_{\text{Markus}} = 0.981$

The stopping power ratios, $S_{\text{w/air}}$, for electron beams as a function of $\bar{E}$ and depth in the phantom can be found using table 5 in the same protocol.  
$S_{\text{w/air}} = 1.081$

The absorbed dose to air chamber factor ($N_{D,\text{air,Markus}}$) provided by the NPL was used then to calculate the absorbed dose to water using equation 5.1. $N_{D,\text{air,Markus}} = 48.186$ cGy/nC. The MUs used in the calibration was 500 with fields at angles $252.5^\circ$ and $287.5^\circ$. The total reading in nC measured for both fields was 1.92975 nC. Then using equation 5.1 the absorbed dose ($D$) at 10 mm was calculated.

\[D_w (Z_{\text{ref,w}}) = 1.92975 \times 1.0386 \times 0.974 \times 1.0028 \times 48.186 \times 0.981 \times 1.081\]

$D_w (Z_{\text{ref,w}}) = 100.03 \text{ cGy/500MU per dual stationary}$

$D_w (Z_{\text{ref,w}}) = 50.0 \text{ cGy/500MU per field}$

The absorbed dose to water per dual stationary field is $50.0 \text{ cGy} \times 2 / 500 \text{ MU}$

$D_w (Z_{\text{ref,w}}) = 0.200 \text{ cGy/MU}$

Therefore the MU per field can be calculated using the B factor. If the doctor prescribe a skin dose of 150 cGy per treatment, then $0.200 \text{ cGy/MU} \times 2.6 = 0.520 \text{ cGy/MU}$

The monitor unit per field can then be calculated,
5.15 Summary:

- Two different types of beam modifier materials were tried in the project, Perspex and Kodak X-Omat V films. The Perspex could not be used due to the difficulty of finding a suitable thickness, while a suitable thickness can be readily obtained using multiple films. Two developed non exposed films were used with total thickness of about 0.4 mm and size of 24 cm x 4.5 cm.

- Beam uniformity along the vertical direction was improved with the scattering filter. Uniformity without filters is $\pm 5\%$ over 160 cm distance, while with the filter the uniformity was found to be $\pm 2\%$ without using collimator rotation and $\pm 3\%$ with collimator rotation. However, the uniformity was measured on a flat phantom which does not represent the distribution of the surface dose on a real patient.

- No change was found in the beam uniformity along the horizontal direction with and without using the scattering filter.

- The measured uniformity meets the recommendation of the AAPM for the electron beam requirements for TSET treatment.

- Dual stationary field with scattering filter depth dose analysis shows that the depth of maximum dose is at 10 mm. The dose falls off to 90% of the dose at $d_{max}$ at depth of 14 mm and to 10% of the dose at $d_{max}$ at 24 mm.
The x-ray contamination from a dual stationary field with scattering filter was found to be 1.7% of the dose at $d_{max}$ beyond the depth of 30 mm.

Dual six fields with filter analysis show that the depth of maximum dose is at about 0.20 mm (at skin). The dose falls off to 90% at 3 mm and falls to 10% at 21 mm depth. Since the mycosis fungoides disease is confined to the skin, it is required to have 90% of the maximum dose at a depth of 3 mm to 5 mm which is the thickness of the dermis layer. The measured data for the dual six fields almost meet the requirement of the beam penetration.

The x-ray contamination measured from the dual six fields was about 2% of the dose. This does meet the recommended beam requirements mentioned in section 2.2. It requires that the x-ray contamination to be less than 4% of the maximum dose beyond the 30 mm depth.

The surface dose measurements from the dual six field on the RANDO phantom show a dose variation of ± 6.3% on the anterior surface, and ± 5.7% on the posterior surface.
6.1 Discussion:

To summarize, the beam characteristics of Varian 2100C 6 MeV electron beam using high dose rate total skin electron mode (HDTSe) with and without beam modifiers have been investigated. The aim was to study the effect of the beam modifiers on the 6 MeV electron beam and to compare the results to the non-modified electron beam. The AAPM report 23 [67] recommended that the beam uniformity be $\pm 8\%$ in the vertical direction and $\pm 4\%$ in the horizontal direction over the central 160 cm x 60 cm area of the treatment plane.

The ideal beam penetration (depth dose) should have:

- The maximum absorbed dose at the surface of the skin.
- Dose fall off to 90\% of the maximum dose be in the range of 3 mm to 5 mm.
- Dose fall off to 10\% of the maximum dose to be at approximately 20 mm depth.
- The bremsstrahlung to be less than 4\% of the maximum dose beyond a depth of 30 mm.

The dual angle used in the project was found to be $\pm 17.5^\circ$ above and below the horizontal axis. Dose uniformity within a 36 cm range in the vertical direction (from -18 cm to 18 cm) was found to be within $\pm 1\%$ using 17.5\%. Scanning range was limited to 36 cm due to the measuring system constraints. This small scanned distance gave a good indication of the beam uniformity using different angles.

It is important to shield the ionization chamber cable when the measurements are carried out. The main contribution of the extracameral signal comes from the adaptor connecting the chamber's cable with the extension cable. The percentage of extracameral signal for a 210 cm
cable length is about 5.5% of the total signal. This can be reduced by shielding the cable and the connection using a steel pipe of 1 cm thickness. The measured extracameral signal at the adapter was reduced from 2.5% to about 1% using the steel pipe.

The beam characteristics of the 6 MeV HDTSe beam from the Varian 2100C linear accelerator were:

- The percentage depth dose measurements for a dual stationary field showed that the beam is too penetrating to be used in the treatment of mycosis fungoides. The disease is confined approximately within the depth range 1 mm to 5 mm in the skin tissue.

- The depth of maximum dose for a dual stationary field was found at 11 mm. The maximum dose falls of to 90% at 15 mm and to 10% at about 25 mm.

- The x-ray contamination of the dual stationary field beyond the 30 mm depth is 1%. The x-ray contamination increases slightly at 80 cm off axis distance in a vertical direction from about 0.6% to 1%.

- Off axis percentage depth dose curves for dual stationary field showed that the surface dose decreases from 85% to 79% of the maximum central dose at off axis distance of 80 cm in the vertical direction. This is due to the energy degradation as the distance increases between the source and the surface of the phantom (SSD). The mean energy at the surface at the center was found to be 4.4 MeV; it decreases slightly to 4.1 MeV at 80 cm off axis distance.
The beam penetration using the dual six fields is significantly different from that obtained with the dual stationary field. This difference depends slightly on the change in the SSD which causes small energy degradation. The energy degradation is small due to the low density of air. The main reason for the change can be only the oblique electron beam incidence effects.

The percentage depth dose measurement for the dual six fields showed that the maximum dose is at the surface. The dose falls off to 90% of the maximum dose at about 1.5 mm depth and to 10% of the maximum dose at 22 mm. The x-ray contamination from the dual six fields beyond 30 mm depth is 1.3%.

The measured beam penetration parameters do not produce an ideal beam penetration curve as mentioned before, where the 90% depth dose should be within 3.0 to 5.0 mm. The x-ray contamination are found to be within the 1% to 4% of the maximum dose.

Dose uniformity measurements along the vertical direction showed that the variation is within ± 5% over 160 cm distance, and within ± 4.2% over 60 cm in the horizontal direction. The dose uniformity along the vertical direction meets the recommendations of the AAPM report 23 [67] and almost meets the recommendations along the horizontal direction. It was recommended in the AAPM report 23 [67] that the dose uniformity be within ± 8% in the vertical direction and ± 4% in the horizontal direction over the central 160 cm x 60 cm area of the treatment plane.
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The scattering filter measurements were introduced in chapter 5. The aim was to study the effect of the filter on the 6 MeV HDTSe beam characteristics, and to improve the beam uniformity even though the uniformity meets the recommendations of the AAPM. The first type of filter material used in the project was Perspex, but it was difficult to obtain the required thickness of this material in the local market. The available thickness of the Perspex in the department could be used for energy degradation but not as a scattering filter. The second scattering filter material was obtained using developed non-exposed radiographic film (no silver component). The composition of the film material is almost the same as the Perspex, but because films are available with a thickness of approximately 0.2 mm it is easier to obtain the required thickness than using Perspex.

The beam characteristics of the 6 MeV HDTSe beam from the Varian 2100C linear accelerator with scattering filter were:

- The depth of maximum dose for a dual stationary field was found at 10 mm. The dose drops to 90% of the maximum dose at 14.5 mm and to 10% at approximately 24.5 mm. The x-ray contamination was increased with the scattering filter from 1.0% to 1.7% compared to the non-modified dual stationary field.

- Off axis percentage depth doses along the vertical direction of dual stationary field showed that the percentage of dose at the surface was increased from 81% of the maximum dose along the central line to about 85% at 80 cm distance. A slight shift was found in the depth of maximum dose towards the surface of the phantom.

- The percentage depth dose measurement for dual six fields with scattering filter showed that the maximum dose is at the surface. The dose falls off to 90% at ~3 mm
depth and to 10% of the maximum dose at 21 mm. The x-ray contamination from the
dual six fields beyond 30 mm depth is 1.5%.

• The dose uniformity using the scattering filter was improved. The variation over 160
cm in the vertical direction is ± 3%, and along 60 cm in the horizontal direction is ±
4%. The results meet the AAPM beam requirement recommendations.

• The dual six fields surface dose measurements using TLDs with RANDO phantom
showed the effect of the surface curvature on the uniformity of the dose. There are
protruding and indented regions, which cause a variation in the dose. The anterior
surface of the phantom showed a dose variation of ± 6.3%, with ± 5.7% on the
posterior surface.

• B factor was redefined in this project as the ratio of skin dose from dual six fields to
the dose at 10 mm depth from a dual stationary field. The IPEM code of practice
protocol (1996) [41] was used to measure the absorbed dose using a Markus ionization
chamber. This chamber is not recommended in the new IPEM protocol (2003) [42].
This is due to the design of the chamber.

The dual six fields technique with scattering filter investigated in this MSc project has
improved the desired beam characteristics required in the treatment of mycosis fungoides. The
x-ray contamination was increased by the scattering filter to 1.5% of the dose which is with in
the recommendations of the AAPM [67]; 1% to 4% of the maximum dose. The beam
uniformity was also improved even though the uniformity meets the AAPM recommendations.
Most of measured data in this project were carried out using flat geometries (flat phantoms), which is not representative to the clinical realities. Since the human body is not a flat surface shape or a simple cylindrical, not only are there areas of overexposure, but there are marked unexposed areas which often requires supplementary treatments. In vivo TLD dose measurements can be used to identify areas requiring local boost fields. The dose uniformity will change depending on the shape and size of the patient. Therefore, by applying some extra in vivo measurements, this technique can be used in the clinically to treat mycosis fungoides disease.
Bibliography


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