A NEW PENUMBRA GENERATOR
FOR
MATCHING OF ELECTRON FIELDS.

by

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ABSTRACT

We describe the geometric and dosimetric characteristics of a device developed to modify the penumbra of an electron beam and thereby improve the dose uniformity in the overlap region when fields are abutted. The device is a Lipowitz metal block placed on top of the electron applicator's insertion plate and positioned to stop part of the electron beam. The air-scattered electrons beyond the block increase the penumbra width from about 1.4 to 2.7-3.4 cm. The modified penumbra is broad and almost linear at all depths for the 9 MeV and 12 MeV electron beams used in this study. Film dosimetry was used to obtain beam profiles and isodose distributions. Without the penumbra generator, lateral setup errors of 2 to 3 mm may introduce dose variations of up to 20% in the junction region. Similar setup errors cause less than 5% dose variations when the penumbra generator is used to match the fields.
RÉSUMÉ

Nous décrivons les caractéristiques géométriques et dosimétriques d'un dispositif développé pour modifier la pénombre d'un faisceau d'électrons et ainsi améliorer l'uniformité de la dose dans la région de chevauchement lorsque des champs sont joints. Le dispositif est un block métallique placé sur la plaque d'insertion de l'applicateur d'électrons, et positionné pour arrêter une partie du faisceau d'électrons. Les électrons diffusés dans l'air au-delà du block accroissent la largeur de la pénombre d'environ 1.4 à 2.7-3.4 cm. La pénombre modifiée est large et presque linéaire à toutes les profondeurs pour les faisceaux d'électrons de 9 MeV et 12 MeV utilisés dans cette étude. La dosimétrie par film a été utilisée pour obtenir des profiles de faisceaux et des distributions d'isodoses. Sans le générateur de pénombre, des erreurs de positionnement latérales de 2 à 3 mm. peuvent introduire des variations de dose de plus de 20% dans la région de jonction. Des erreurs de positionnement similaires provoquent des variations de dose de moins de 5% lorsque le générateur de pénombre est utilisé pour joindre les champs d'électrons.
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Megavoltage electron beam radiotherapy is ideal for irradiating superficial tumors because of the limited range of electrons in tissues. However, the treatment of extended areas with electrons often requires the use of two or more adjacent electron fields. In such cases, unacceptably large dose variations may arise at the junction of the fields. These dose variations come from the presence of large bulges in the isodose curves created by electron beam divergence and lateral electron scattering in tissues. Overlapping of these bulges creates a high-dose region at depth, while the constriction of the isodose curves near the surface may produce a low-dose region. The shape and extent of these regions depend critically on the separation of the electron fields. This thesis presents a new and simple method to modify an electron beam to produce a wide penumbra and yield an excellent dose uniformity at the junction between adjacent electron fields.

Chapter One provides a general introduction to radiotherapy and discusses the rationale for the thesis.

Chapter Two is a brief review of the currently known approaches that may be used to solve the problem of adjacent electron beams.

Chapter Three discusses electron interactions with matter and presents the parameters used to characterize the energy of an electron beam in clinical practice. Different aspects of clinical electron beams are reviewed.

Chapter Four is a general review of electron beam dosimetry. Particular attention is given to film dosimetry because of the extensive use of this modality in the work presented in this thesis. Many important aspects of film dosimetry are discussed by either theoretical considerations or through experiments performed by the author.

Chapter Five describes the geometric and dosimetric characteristics of the new device proposed to modify the penumbra of an electron beam and thereby improve the dose uniformity in the overlap region when fields are abutted. The
experimental procedure used to find an optimal field separation and to quantify the uniformity of the dose distribution in the junction region is presented.

Chapter Six regroups the results of a systematic study of electron field matching (on a flat surface) with the penumbra generator.

Chapter Seven describes the potential of the field matching technique for the irradiation of curved surfaces. A clinical application of the penumbra generator is also presented.

Chapter Eight describes the future investigations necessary to thoroughly evaluate electron field matching on curved surfaces and discusses the possibility of using the penumbra generator for matching electron and photon fields. A summary of the thesis is also presented.
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1.1 Introduction  

This chapter introduces radiotherapy by describing the basic characteristics of radiotherapy treatment units and by providing an overview of the radiotherapy process. The goals and applications of radiotherapy together with the basic quantities used to describe the behavior of a radiotherapy beam in the patient are discussed. The rationale for the thesis is also presented.

1.2 Basic Characteristics of Radiotherapy Treatment Units.

Radiotherapy is the application of ionizing radiation in treatment of disease. It is one of the available modalities for treatment of cancer. In modern oncology, about 40% of cancer patients are treated by surgery, some 40% by radiotherapy, and 20% by chemotherapy. These three types of therapy are often used in combination with each other. Thus, radiotherapy is used for either cure or palliation as the goal. Two main approaches are to be distinguished in radiotherapy: brachytherapy (or curietherapy) and external beam radiotherapy. Brachytherapy is the method of treatment in which sealed radioactive sources are used to deliver radiation at a short distance by interstitial, intra cavitory, or surface application. When radiation is delivered from an outside source (usually...
about 1 meter from the patient), the technique is referred to as external beam radiotherapy (or teletherapy).

Photons and electrons, with energies in the kilo-electron-volt (keV) and mega-electron-volt (MeV) range, are the principal types of radiations used in radiotherapy. The high energy photons used in teletherapy come from two sources: gamma rays arising from the disintegration of a radioisotope or x rays produced by the deceleration of electrons in a target. Cobalt-60 is the only radioisotope used routinely in modern teletherapy. X rays are produced by two main sources: orthovoltage units and medical linear accelerators (linacs). Orthovoltage units are similar to diagnostic x-ray tubes, in which the electrons are accelerated across a large static potential difference. Orthovoltage units produce photons in the keV range which are useful for treating superficial lesions because of their relatively low penetration in tissues. To treat deeper-seated lesions photons in the MeV range are used. Such photons are produced by linacs, in which electrons are accelerated either by traveling or stationary electromagnetic waves in a waveguide. The frequency of the waves used is in the microwave region (~3000 MHz). Linacs are also used to obtain electron beams. As they exit the window of the accelerator tube, the electrons form a narrow pencil beam. In the electron mode of linac operation, the electron pencil beam is made to strike a scattering foil instead of the x-ray target. This scattering foil spreads the beam and provides a uniform electron fluence across the treatment field. The thickness of the foil is such that most of the electrons striking it are scattered instead of suffering bremsstrahlung. However, a small fraction of the total electron kinetic energy is still converted into bremsstrahlung and appears as x-ray contamination of the electron beam. Because of their limited range in tissues, megavoltage electron beams are ideal for irradiating superficial lesions. This thesis deals with external electron beam radiotherapy.

The radiation oncology department of the Centre Hospitalier Universitaire de Québec located at the Pavillon L'Hôtel-Dieu de Québec houses one Theratron cobalt unit, one orthovoltage unit and six Siemens Mevatron linear accelerators (four KD-2, one MD and one MD-2). The experiments presented in this thesis were performed with the Siemens linear accelerator (KD-2) which has two levels of photon energies (6 MV and 23 MV) and six levels of electron energies (6, 9, 12, 15, 18 and 21 MeV). In the photon mode of operation (at 6 MV for
example) the electrons are accelerated to a kinetic energy of 6 MeV and they produce a spectrum of photon energies ranging from 0 to 6 MeV which is referred to as a 6 MV x-ray beam. In the electron mode of operation, the electrons coming out of the accelerator have a fairly small dispersion in kinetic energy and the spectrum is distributed around the stated nominal energy level.

The Medical Linear Accelerator (linac).

The structure of the linac consists of five main parts: the head which contains the beam transport and beam forming components, the gantry to which the head is attached and which enables rotation of the radiation source around the patient, the couch upon which the patient is positioned for the treatment delivery, the stand which contains the electronic components and supports the gantry, and the console which controls the linac operation. A schematic representation of a typical linac is presented in Fig. 1.1.

The majority of modern linacs are mounted isocentrically. Isocentric linacs are constructed such that the source of radiation rotates about a horizontal axis referred to as the isocenter axis. The point of intersection of the beam central axis and the axis of rotation of the gantry is known as the isocenter. This arrangement provides the needed flexibility to treat the diseased site from different directions or entry points without having to reposition the patient for each treatment beam. The Siemens Mevatron KD-2 linac has its isocenter at 100 cm from the radiation source, and we refer to this distance as the source-axis-distance (SAD). The treatment couch moves in the up and down direction, it can be moved left and right, and towards and away from the gantry. The couch rotates about a vertical axis which passes through the linac isocenter, and an additional rotation is provided between the upper part of the couch and the base holding it to the floor (the column rotation). Alignment lasers on the walls and ceiling of the room as well as the light indicators (Optical Distance Indicator (ODI)) giving the distance between the source and the surface of the patient aid in positioning the patient accurately for treatment.

The components of the treatment head of a typical linac are depicted in Fig. 1.2. The treatment head consists of a thick shell of high-density shielding material (usually lead-tungsten alloy) which provides sufficient shielding against leakage
Figure 1.1  Schematic representation of four main components of an isocentric linear accelerator: treatment head, gantry, stand and couch.

Figure 1.2  Components of the treatment head of a typical linear accelerator. (A) X-ray therapy mode. (B) Electron therapy mode.
radiation. As discussed above, it contains the x-ray target and the electron scattering foils. In the photon mode (Fig. 1.2 (A)), the x-ray target is placed in the path of the electron pencil beam. Because the linac produces electrons in the megavoltage range, the intensity of the generated x-ray beam (referred to as bremsstrahlung) is peaked in the forward direction. To make the beam useful for radiotherapy, a flattening filter is inserted in its path to provide a uniform intensity across the field.

In the electron mode (Fig. 1.2 (B)), the x-ray target and the flattening filter are moved out of the beam path, and the scattering foil is placed in the path of the electron beam. The flattened x-ray beam or the electron beam is incident on a monitor chamber. This monitoring system usually consists of two independent transmission type ionization chambers covering the entire beam. Its function is to monitor the dose rate, integrated dose, and field symmetry. It also shuts the beam off when the preset number of monitor units (MU) has been delivered. After passing through the ion chambers, the beam is collimated by a continuously movable x-ray collimator. This collimator consists of two pairs of lead/tungsten blocks (jaws) which allow the radiation field to be shaped to any rectangular size, from 0x0 to the maximum field size (generally 40x40 cm² at the isocenter). The collimators can rotate about the beam central axis. The treatment head also contains a light localizing system which consists of a combination of mirror and a light source located in the space between the chambers and the jaws. The localizing system projects a light beam as if it was emitted from the x-ray focal spot. Thus, the light field coincides with the radiation field and aids in positioning the patient for the treatment.

In the photon mode, the field is defined by the x-ray jaws at about 30 cm from the source. In the electron mode, because the electrons undergo appreciable scattering in air, beam collimation is achieved close to the skin surface of the patient. Thus, when treating with electrons, the x-ray jaws are opened sufficiently wide and an auxiliary collimator for electrons is attached to the treatment head. This collimation is often referred to as the secondary collimation, in contrast to the primary collimation provided by the x-ray jaws. The electron collimation systems vary widely from one linac manufacturer to the other. In the case of the Siemens Mevatron KD-2 used in this study, the auxiliary electron collimator consists of a set of attachable square cones of various sizes.
The Siemens electron cones will be described in detail in a Chapter V. For the moment, it is important to point out that due to the electron scattering the dose distribution in an electron field is influenced significantly by the secondary collimation design. The work presented in this thesis deals with the modification of the secondary electron collimation device in order to change the dose distribution in the electron field.

1.3 Basic Quantities used to describe a Radiotherapy Beam.

There are three basic quantities which are used for an effective description of the behavior of a single beam of radiation in a patient (or in a block of tissue-equivalent material called a phantom): the percent depth dose (PDD), the off-axis ratio (OAR) (dose profile), and the dose output (relative dose factor).

![Percent depth dose curve](image)

**Figure 1.3** Percent depth doses in water for electron beams of various energies for the Siemens Mevatron KD-2 linac and a 25x25 cm² electron cone.

The percent depth dose gives the variation of the dose with depth in a medium. It is measured along the beam central axis and is normalized to 100% at the depth of maximum dose $d_{max}$. Figure 1.3 shows PDD curves for electron beams of various energies measured in water (water is the best tissue-equivalent material) for a 25x25 cm² electron field. The region between the surface and
depth dose maximum is referred to as the dose build-up region. The extent of the build-up region and the dose at the surface depend on the electron beam energy and field size as well as on the absorbing medium. Furthermore, the overall shape of the PDD curve is dependent on the source-surface distance (SSD).

The *dose profile* (or beam profile) gives the variation of the dose at a given depth in a medium in a direction perpendicular to the beam central axis. Beam profiles are measured at the center of the field, either along the width or the length, and are normalized on the beam central axis. The shape of beam profiles depends on the depth in the phantom (patient), especially for electron beams where lateral scattering in tissues is important. Figure 1.4 shows a typical beam profile taken in water at a depth of $d_{\text{max}}$ for a 9 MeV electron beam with a 15x15 cm$^2$ electron cone. The region of the dose fall-off near the edge of the field is known as the penumbra region. For high energy photon beams the penumbra width is relatively small; in the case of high energy electron beams, on the other hand, the penumbra width increases significantly with depth in tissues because of lateral scattering of the electron beam.

![Figure 1.4](image)

*Figure 1.4* Beam profile taken in water at a depth of $d_{\text{max}}$ (2 cm) for a 9 MeV electron beam. Data obtained with the Siemens Mevatron KD-2 linac and a 15x15 cm$^2$ electron cone.
The output of a radiotherapy treatment unit is the absolute dose imparted at a specific point in phantom. Usually, the output is measured at $d_{\text{max}}$ with the phantom surface at the isocenter. The output is field size dependent, and calibration of a treatment unit is usually performed with a reference field size. In the case of linacs, adjustments are made to the ion chamber sensitivity so that, for every level of photon and electron energy, the output for the reference field size (at nominal SSD) is 1 cGy/MU. Knowledge of outputs, PDDs, and beam profiles for the range of available field sizes enables the determination of absolute dose at any point in phantom.

An important means for the representation of volumetric or planar variation in absorbed dose is the isodose curve. Isodose curves are lines which connect points of equal dose in a medium. The curves are usually drawn at regular intervals of absorbed dose and expressed as a percentage of the dose at a reference point. Isodose distributions are the most useful dose distribution representation in treatment planning. An isodose distribution for a typical electron beam is shown in Fig. 1.5.

![Figure 1.5 Isodose distribution in polystyrene for a 9 MeV electron beam. Data obtained with the Siemens Mevatron KD-2 linac and a 15x15 cm$^2$ electron cone.](image_url)
1.4 Overview of the Radiotherapy Process.

The main steps which are followed when radiotherapy (teletherapy) is prescribed will now be reviewed. These steps may vary from one radiotherapy center to another and also depend somewhat on the disease treated, but the general process is as follows:

1) Immobilization devices, such as plaster casts, alpha-cradles or orfits are fitted to the patient. These devices facilitate accurate patient set-up for treatment delivery. Although not always used, they are especially important for the treatment of lesions in the head and neck region.

2) The patient undergoes treatment simulation. This is achieved by setting-up the patient (with the immobilization device) on a treatment simulator, which is a machine similar to the actual treatment unit but producing x-rays in the diagnostic energy range instead of megavoltage photons. The simulator can be used to take radiographic films or it can be used in the fluoroscopy mode. On the simulator, the physician establishes or verifies under fluoroscopy the region to be treated. The target volume is the volume which needs to be irradiated to a specified absorbed dose. The target volume should include sufficient margins to allow for the uncertainty in anatomic localization of this volume. Additional margins must be provided around the target volume to allow for limitations of the treatment technique. The target volume together with this additional margin constitutes the treatment volume.

Reference marks are sketched with special ink on the patient's skin to locate the position of the isocentre within the patient, and to identify the position of the different fields used. An axial contour of the patient at the level of the isocentre is then taken. If the patient shows large variations in thickness in the treatment field, contours at other levels are also taken. Finally, films of the proposed treatment fields are produced.

In certain cases, when more detailed information on the internal structures is required, studies using a CT scanner are implemented.
those cases, the target volume is drawn on the CT scan and the patient contour can be extracted from the CT scan data.

3) The information obtained during simulation (manual body contour or CT scan and simulator x-ray films) is sent to the dosimetrist who produces a treatment plan consisting of an isodose distribution showing the dose distribution at all relevant points in the patient. If not previously determined during simulation, the dosimetrist has six variables to work with to produce the isodose distribution: the number of beams used, the field size of each beam, their point of incidence on the patient, their energy, their weighting, and accessories (such as wedges) to be used. For fields which are not rectangular, either the machinist is asked to make the special shielding required or a multileaf collimator is used for treatment.

4) The treatment plan (the isodose distribution produced by the dosimetrist) is examined by the radiation oncologist who decides whether or not it is acceptable. The two main criteria of acceptability are: 1) the dose uniformity over the target volume must be acceptable and 2) the dose to the surrounding healthy tissues must be as low as possible. If the treatment plan is found to be appropriate, then the prescription (prescribed dose of radiation to be given to the treatment volume) is decided upon. This prescription is then translated into the appropriate machine setting (i.e., the number of MU for each daily treatment field on a teletherapy machine).

5) The patient is treated according to the prescription of radiation in small daily doses (fractions) over a period of three to five weeks. The treatment set-up and the positioning of the patient are verified by check films or by images acquired using an electronic portal imaging device. The patient's response to the treatment is monitored by weekly consultations with the physician. Finally, upon completion of treatment, the patient is followed by the physician on a regular basis to make sure that rehabilitation is complete.
1.5 Rationale for the Thesis.

No matter how sophisticated radiotherapy becomes, there will always be an uncertainty associated with the delivery of dose. This uncertainty can arise from any link in the chain of activities in the radiotherapy process: accurate delineation of the tumor volume, calibration of the treatment unit, treatment planning process, etc. It has been shown that small changes in dose can lead to appreciable changes in the probability for tumor control. The relationship between radiation dose and the probability for tumor control of a homogeneous group of tumors is sigmoidal. Thus, with increasing radiation doses, more and more diseased cells are killed until ultimately, all clonogenic cells are destroyed and a cure is achieved. Dose response relationships for local control of homogeneous tumor groups have been empirically determined. Unfortunately, the dose of radiation which can be delivered to a tumor is limited by the probability of complications associated with the inevitable irradiation of normal tissues. Therefore, the prescription of a tumor dose is based on the relative probability of tumor control and normal tissue complications. Figure 1.6 shows a typical theoretical dose response relationship for tumor control and normal tissue complications.

The two curves of Fig. 1.6 have been drawn parallel to one another for simplicity, although it is likely that in practice the tumor-control curve will be shallower than that for normal tissue response. The greater the displacement between the two curves, the higher the probability of tumor control will be. Little quantitative information about dose response curves for particular tumor types is available. However, the available information is sufficient to allow some conclusions to be made regarding accuracy of dose delivered to the patient. The slopes of the curves are sufficiently steep that a change in dose of 5% is expected to produce a change of over 10% either in tumor control or complication rate. For this reason, the ICRU report has recommended that an overall accuracy of ±5% in dose delivery be considered as reasonable, in which case higher accuracy is required for each individual step in the radiotherapy process. Although the degree of precision required varies with the type of cancer, the ±5% accuracy in dose delivery recommended by the ICRU is the currently accepted standard.
The previous discussion emphasizes the need to strive for high precision in radiotherapy. In treatment planning, the dose uniformity criterion over the target volume appears as an extremely important goal to insure local tumor control over the entire volume of the disease. This brings us to a problem related to treatment with megavoltage electron beams. The treatment of extended areas with electrons often requires the use of two or more adjacent fields. This situation might come up when the extent of the area to be irradiated is greater than the largest available electron applicator. Sometimes, the shape of the target volume is such that beams of different energies have to be abutted. Another example of such a situation is when the curvature of the area to be irradiated is too pronounced to allow irradiation with a single field. In such cases the dose reduction at the edges of the field due to the increased SSD and the
oblique incidence\textsuperscript{7,8,9,10} make it impossible to obtain a uniform dose distribution over the entire target volume. When abutting electron fields, unacceptably large dose variations may arise at the junction of the fields. These dose variations come from the presence of large bulges in the isodose curves which can be seen at the edges of the isodose distribution presented in Fig. 1.5. These bulges are created in part by electron beam divergence, but mostly by the important lateral scattering of electrons in tissues. Overlapping of these bulges creates a high-dose spot at depth, while the constriction of the isodose curves near the surface may produce a low-dose spot, depending critically on the field separation. This may lead to either overdose complications of the normal surrounding tissues or underdosage of the potentially malignant regions. Systematic studies\textsuperscript{11,12,13} of the problem of field abutment have reported that dose variations at the field junction of adjacent electron fields may be $\pm 20\%$ or more. To overcome this problem, several authors have proposed techniques for matching electron beam edges in such a way as to make the overlap region as uniform as possible. These different, currently used, approaches to the problem will be reviewed in Chapter II. In this thesis, we propose a new solution to the problem. We describe a new and simple method to modify an electron beam to produce a wide penumbra whose shape is independent of depth. This modified penumbra yields an excellent dose uniformity at the junction between adjacent electron fields.

1.6 References.


2.1 Introduction

In this chapter, the different approaches that have been used or are currently used to solve the electron field matching problem are reviewed. As we have seen, the presence of large bulges in the penumbra of electron fields makes the dose distribution in the overlap region between adjacent fields difficult to optimize. We can classify the different solutions to this problem into three main categories: 1) simple optimization of the separation between the fields to minimize the dose variations at the junction, 2) modification of the electron beam at the edge to be abutted to improve dose uniformity, and 3) matching electron beams without secondary collimation.

2.2 Beam-Edge Modification for Matching of Electron Fields.

The problem of electron field matching was presented in Section 1.5. As we have seen, overlapping of scatter-induced bulges in the isodose curves is responsible for the presence of a high dose region at depth while the constriction of the isodose curves produces a low dose region near the surface. The situation can
be pictured clearly if we describe the penumbral dose distribution in terms of dose decrement lines, as shown in Fig. 2.1.

![Figure 2.1](image)

**Figure 2.1** Electron isodose curves and dose decrement lines for a 20 MeV electron beam: (A) unmodified penumbra; (B) hypothetical ideally modified penumbra. Curve (A) was recorded in water with a 14x10 cm² electron cone. Redrawn from Kalend et al.

Decrement lines join points at which the dose is a fixed percentage of the central axis dose at the same depth in the medium. Here, electron decrement lines are approximated by straight lines to simplify the discussion. As shown in Fig. 2.1 (A), the decrement lines of an electron penumbra diverge asymmetrically as a consequence of beam divergence and lateral electron scattering. Furthermore, we can observe the close spacing of the decrement lines near the surface which is responsible for the severe dose inhomogeneity coming from small beam positioning errors. From these observations, we can deduce an electron penumbra ideal for abutment. Such an ideal penumbra would be characterized by decrement lines which are widely and evenly spaced parallel lines, as illustrated in Fig. 2.1 (B). Such parallel decrement lines produce a beam edge whose shape is independent of depth, while the equal spacing provides a linear dose gradient in the penumbra region. With two beam edges modified in this manner, one being the mirror image of the other, the dose decrement lines from one beam will add to the ones of the second beam in a complementary fashion.
An ideal matching is thereby achieved over the entire depth, width, and length of the overlap region. Thus, the aim of beam-edge modification is to produce planes of matched dose throughout the volume of the junction region. Furthermore, by providing a smooth and linear dose gradient, a modified beam-edge should make the junction region between adjacent fields less sensitive to lateral positioning errors.

2.3 Skin Gap Optimization.

The simplest approach to the problem of electron field matching is to optimize the separation (measured at the surface of the patient) between two adjacent electron field edges. Optimization is achieved by a complete set of trial and error measurements in order to find a gap between the field edges for which an acceptable dose uniformity in the junction region is obtained. Dose uniformity of the order of ±5% can be achieved with this method, however, this approach has several major drawbacks.

The lateral scattering of electrons in tissues is more important at lower energies. This means that the bulges in the isodose distribution are more and more pronounced as the electron beam energy is decreased. Furthermore, as we have seen in Chapter 1, the dose distribution in an electron field (in particular in the penumbra region) is influenced significantly by the secondary collimation design. Thus, because of increased lateral scatter of low-energy electrons and machine specific characteristics of an electron beam penumbra, the determination of an optimal skin gap for electrons is somewhat more complicated than is the case for photon beams. For photon beams, optimal gaps can be calculated from simple considerations of geometric beam divergence. In the case of electrons, on the other hand, a large amount of data must be accumulated for each beam energy to achieve optimization, and optimal gaps obtained for one treatment unit cannot be used with other treatment units.

The main limitation to the usefulness of the optimized skin gap technique is the strong sensitivity of the dose distribution in the field junction region to small deviations in field separation or in the angulation of the incident electron beams, making it strongly dependent on positioning errors. Several studies have documented the frequency and magnitude of patient set-up errors and positional
uncertainties in external beam radiotherapy. Some pioneering studies used verification films to demonstrate a high incidence of localization errors for different treatment sites, but the recent advent of commercially available Electronic Portal Imaging Devices (EPIDs) has provided an invaluable tool for such clinical investigations. Svensson estimated that positional uncertainty is of the order of 5 mm, corresponding to one standard deviation. Rabinowitz et al. reported on the analysis of simulator and portal films of 71 patients. Some discrepancies were noted between the simulator and the localization portal films. The standard deviation of the variations had an average value of 3 mm. Dunscombe et al. reported on discrepancies between prescribed and treated field edges for 29 patients treated for prostate cancer. The distribution of field edge discrepancies is described by a standard deviation of 4.5 mm and has an average absolute value of 3.5 mm. Others have documented similar localization uncertainties.

Considering all of these studies, it is reasonable to expect an uncertainty of at least 1-2 mm in the patient set-up for a single electron field. When two fields are abutted, the uncertainty in the field separation can be of the order of ±2-4 mm. As will be demonstrated later by measurements performed by the author, lateral set-up errors of ±2 mm in matching electron fields can introduce dose non-uniformity of ±20% or higher. Thus, because of the strong dependence of the junction region on lateral set-up errors and the inherent positional uncertainties in the radiotherapy process, it is difficult to predict the dose distribution in the patient from one fraction to the other when using the optimized skin gap technique.

2.4 Beam-Edge Modifying Devices.

Tissue-equivalent wedges (strips).

The first reported electron field abutment technique is based on a beam edge modifier in the form of tissue-equivalent wedges inserted in the penumbra of the adjoining field edges. These wedges are in the form of plastic strips which are aligned with the edges to be abutted. The electron scattering generated in the wedges acts to broaden the electron penumbra near the surface of the patient. With an appropriate choice of wedge dimensions, a tissue-equivalent
A wedge may produce isodose curves of nearly equal spacing in the overlap region, as illustrated in Fig. 2.2. However, as will be discussed in Chapter III, interposing any material into an electron beam will degrade the electron beam energy. Thus, these wedges may seriously degrade the beam energy and electron penetration depth in the outer portions of the penumbra. Because of their extreme sensitivity to beam energy and set-up parameters, these wedges are highly customized in patient treatment planning, and, because of difficulties in achieving an optimal dose distribution, they are rarely used clinically.

Figure 2.2 Electron isodose curves and 50% decrement lines for beams with penumbra modified by tissue-equivalent wedges: (A) energy = 7.5 MeV, wedge dimensions = 1x2 cm, abutting angle = 45°; (B) energy = 20 MeV, wedge dimensions = 3x2 cm, abutting angle = 10°. Curves recorded in water with a 14x10 cm² electron cone. Redrawn from Kalend et al.¹.

High-density material comb.

Kalend et al.¹ proposed a comb shaped beam-edge modifier made of a low melting-point alloy. The device is a comb with gratings triangular in shape to reduce the penumbra electron flux linearly toward the beam edge. The comb gratings were made 2.3 cm long to reduce the beam-edge dose gradient by a
factor of about 10 (at $d_{max}$), and their thickness was sufficient to stop electrons in the clinical energy range of 5-25 MeV. The maximum spacing between the gratings of the comb was 3 mm. The device was attached to the electron cone (at 5 cm from the phantom surface) and it had to be tilted toward the source to correct for the geometric divergence in the incident beam. Examples of isodose distributions with the penumbra modified by the comb are given in Fig. 2.3.

![Figure 2.3](image)

**Figure 2.3** Electron isodose curves for beams with penumbra modified by a high-density comb: (A) energy = 7.5 MeV; (B) energy = 20 MeV. Curves recorded in water with a 14x10 cm² electron cone. Redrawn from Kalend et al.

The modified penumbra shows, in all essential respects, the characteristics of an ideal penumbra for abutment. Kalend et al. reported dose variations of less than ±2% in the match region between adjacent fields modified with their device. However, the device generates ripples in dose profiles immediately beneath the comb and transverse to the gratings. These ripples reflect the geometry of the comb, and their magnitude is a function of energy. To eliminate this problem, two phase-shifted complementary combs had to be used for field matching. Because of the precision needed in the fabrication and positioning of the device, this technique, similarly to tissue-equivalent wedges, is difficult to implement clinically.
Plastic Wedge Penumbra Generators.

Kurup et al.\textsuperscript{25} proposed a method using plastic wedge penumbra generators. Unlike the small wedges discussed above, these polystyrene wedges are designed to fit inside the electron cone, covering the entire field. They were shaped to match the insert of the electron cone. The distance from the bottom of the wedge to the phantom surface was approximately 6 cm. The thick part of the wedge, which increased the penumbra width by generating electron scattering, was along the edge of the field to be abutted. The physical angle of the wedges used was 3° to 6°. Figure 2.4 gives examples of isodose distributions with the penumbra modified by a wedge. These electron wedges can be designed from a few measurements, and they were quite successful in improving the dosimetry of the junction region; however, the presence of the wedges causes a slight degradation of the beam energy and penetration. Furthermore, the penumbra broadening occurs at all field edges (in the thick part region) and is not limited only to the region of abutment.

Figure 2.4  Electron isodose curves for beams with penumbra modified by polystyrene wedges: (A) energy = 9 MeV, wedge angle = 3°; (B) energy = 15 MeV, wedge angle = 6°. Film measurements in white polystyrene phantom for a 10x10 cm² electron cone. Redrawn from Kurup et al.\textsuperscript{25}.
2.5 Matching Electron Beams without Secondary Collimation.

Another approach is to match electron beams without the secondary collimation. In this case, the electron collimation system of the unit (electron applicator) is not used, and field shaping is achieved through the combined effect produced by the photon jaw settings and lead shielding on the patient's skin. The scattering of the beam by the scattering foil, the monitor ion chamber, and intervening air significantly increases the width of the penumbra. Such beams have profiles close to Gaussian and are therefore easy to match at 50% intensity points to produce a homogeneous dose distribution across a large field. The wide penumbra of such beams makes the dose distribution at the junction relatively insensitive to positional errors.

The matching of electron beams collimated only with photon collimators is relatively simple and can result in a uniform dose in the junction region. However, the flatness of the electron field itself is compromised by the lack of electron cones. Moreover, cumbersome tertiary collimation placed directly on the patient to determine the field edges outside the junction region is required. Because of these problems this approach to electron field matching is not used clinically.

Based on this idea, however, Boyer et al. introduced an electron beam pseudoarc technique for irradiation of curved surfaces. The electron arc therapy is a technique used to treat superficial volumes that follow curved surfaces such as the chest wall. Arc therapy is performed either with continuous beam-on rotation of the electron beam (continuous arc) or with a series of overlapping isocentric stationary electron beams (without electron applicator). The latter approach is known as the pseudoarc technique. The pseudoarc approach has been thoroughly evaluated by Pla et al. The electron arc techniques are complex and therefore not widely used. Furthermore, the dosimetric gain comes at the expense of the setup time in the treatment room because of the necessary lead shielding on the patient to provide a sharp dose cut-off at the edges of the composite treatment field.
2.6 The New Penumbra Generator.

The technique we are proposing in this thesis can be pictured as an hybrid between the beam-edge modifier approach and the field matching without secondary collimation approach. As a matter of fact, our technique modifies the penumbra at the edge of abutment, but it does so by means of a "trimmer" defining the field edge at a higher level with respect to the patient's surface. Thus, we keep the electron applicator in place, but the edge of abutment is defined closer to the source to generate electron scattering in air in a matter similar to the field matching without secondary collimation approach.

2.7 Conclusions.

This chapter has reviewed the different approaches that have been used or are currently used to solve the electron field matching problem. The new technique we propose will be described in detail in Chapter V. Before this, Chapters III and IV will review important aspects of the physics of electron beams and electron dosimetry.

2.8 References.


CHAPTER III

INTRODUCTION TO THE PHYSICS OF ELECTRON BEAM THERAPY.

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3.1 Introduction

This chapter introduces the basic information required to understand the physics of electron beam therapy. The different interactions of electrons with matter in the energy range of interest in radiation therapy are discussed. Furthermore, the parameters used to characterize the energy of an electron beam in clinical practice are presented. Finally, different aspects of clinical electron beams such as the percent depth dose curve, beam flatness and symmetry, or the output factor are reviewed.

3.2 Electron Interactions with Matter.

The transport and penetration of electrons in matter involve interactions which result either in direct energy losses or scattering, or a combination of both. Being surrounded by its Coulomb electric force field, the electron will interact with one or more orbital electrons or with the nucleus of practically every atom it passes. Most of these interactions individually transfer only a small fraction of the incident electron energy, and it is convenient to think of the electron as losing its energy gradually and continuously. This is often referred to as the "continuous
slowing-down approximation". Typically, a 10 MeV electron will undergo some $10^5$ interactions before coming to rest.

The different interactions can be characterized in terms of the size of the classical impact parameter $b$ of the electron trajectory. When $b$ is much larger than the atomic radius $a_e$, the electron force field affects the atom as a whole, thereby distorting it, exciting it to a higher energy level, and sometimes ionizing it by ejecting a valence-shell electron. The net effect is the transfer of a very small amount of energy to an atom of the absorbing medium. Since large values of $b$ are more probable than near collisions on individual atoms, these types of interactions are predominant and account for roughly half of the energy transferred to the medium from an energetic electron. When $b$ is of the order of the atomic dimensions, the probability that the electron interacts primarily with a single atomic electron is appreciable. These kinds of interactions account for roughly the other half of the energy transferred to the medium, even though they are less probable. Finally, when the impact parameter $b$ is much smaller than the atomic radius, the Coulomb force interaction takes place mainly with the nucleus, resulting in bremsstrahlung loss of energy and change in trajectory (scattering).

The electrons can experience either elastic or inelastic Coulomb interactions with the nuclei and orbital electrons of the medium. These four different interactions will now be discussed.

1) **Inelastic Electron-Electron Collision ($b \sim a_e$).**

The incident electron is scattered by an orbital electron while also ionizing the atom. Since part of the incident electron energy has been used to liberate the atomic electron from its bound state, the event is inelastic. In such an interaction, the incident electron can suffer up to a 50% energy loss and any possible change in direction (scattering). If the energy transfer is sufficiently large, the secondary electron may have enough energy to produce a track of ionization of its own. Such a secondary electron is referred to as a delta ray.
2) **Elastic Electron-Electron Collision** \((b \sim a)\).

The incident electron is scattered by an atomic electron without ionizing it. The scattered electron suffers a small change in direction, and little or no loss of energy.

3) **Elastic Electron-Nucleus Collision** \((b \ll a)\).

In about 97-98% of electron-nucleus encounters, the incident electron is scattered elastically. In these interactions, the electron loses just an insignificant amount of kinetic energy necessary to satisfy the conservation of momentum for the collision. Even though it is not a mechanism for transfer of energy to the absorbing medium, it is an important means of deflecting electrons, and is the principal reason why electrons follow very tortuous paths, especially in high atomic number media.

4) **Inelastic Electron-Nucleus Collision** \((b \ll a)\).

The incident electron is strongly decelerated by the interaction with the electric field of the nucleus and the kinetic energy lost is emitted in the form of photons, as described by the Larmor relationship. Such x rays are referred to as bremsstrahlung, the German word for "braking radiation". In such interactions, the electron can give up to 100% of its kinetic energy to the photon in one single interaction.

Unlike photons, which undergo one major event eliminating them from the beam, the typical electron undergoes a large number of minor interactions while remaining in the beam. Thus, a simple characterization of an electron beam's behavior in a medium in terms of an attenuation coefficient is impossible. However, we know that the average energy of the electrons in the beam decreases with penetration, since they lose energy in a "continuous" manner. It is this feature which is used to describe the interaction of the electrons with matter. The quantity used for this purpose is the linear stopping power, defined as the quotient \(\frac{dE}{dl}\), where \(dE\) is the fraction of energy which an electron loses
during its passage through a medium along an increment of path-length $dl$. The stopping powers are statistical quantities, and do not represent the behavior of any single electron. In many applications, the linear stopping power is divided by the density of the medium to obtain the mass stopping power. This removes the dependence on the physical density of the medium. The linear stopping power $S$ has units of MeV/cm, while the mass stopping power $S/\rho$ has units of MeVcm²/g. Another useful quantity is the stopping power ratio of two media, which describes the rate of energy loss of electrons in one medium compared with another.

It is customary to consider separately two components of the stopping power: a) the collisional stopping power $S_{col}$ due to interactions with the atom as a whole characterized by interactions 1-3 described above, and b) the radiative stopping power $S_{rad}$ due to the electron interactions with the electric field of the nucleus resulting in the production of bremsstrahlung. The first set of interactions are grouped under the name of collisional interactions, while interaction 4 described above is termed a radiative interaction. Hence, the total mass stopping power is given by:

$$(1/\rho)S_{tot} = (1/\rho)S_{col} + (1/\rho)S_{rad}$$

The division of the total stopping power into two components emphasizes the difference in the mechanisms of energy losses and energy absorption in the medium. The energy lost in ionization and excitation of atoms is absorbed close to the electron track, whereas the energy carried in the form of bremsstrahlung might travel far before it is absorbed in the medium, or it might even escape from the medium. Figure 3.1 gives the collisional and radiative stopping powers for electrons in water, copper, and lead as a function of their kinetic energy.

Although stopping powers are widely used in radiation dosimetry, they are rarely measured, and have to be calculated from the stopping power theory. A detailed description of the theories involved have been covered in several texts. For the needs of clinical radiation dosimetry we make use of extensive tables of stopping powers for electrons which have been calculated and published. The theories involved in stopping power calculations are beyond the scope of this text, however, a few important generalizations will be provided here.
Collisional Losses (Ionization and Excitation)

As shown in Fig. 3.1, the energy loss rate first decreases and then slowly increases with an increase in electron kinetic energy, with a minimum occurring at about 1 MeV for all materials. This dependence on energy is the result of the combination of two effects. As the electron kinetic energy decreases, the electron travels more slowly and spends more time in the vicinity of each atom. The result is an increase in the probability of interaction and a concurrent rise in \((S/p)_{\text{col}}\) at low electron energies. For high electron energies, the electron velocity approaches the speed of light, and its electric field lines warp increasingly into the plane perpendicular to the direction of motion prolonging the spatial range of possible interactions. Thus \((S/p)_{\text{col}}\) rises at high electron energies. The collisional mass stopping power is greater for low atomic number \(Z\) materials than for high \(Z\) materials. There are two reasons for this \(Z\) dependence: a) the electronic density decreases with increasing atomic number and b) high \(Z\) materials have more tightly bound electrons, making them less available for this type of interaction.
Radiation Losses (Bremsstrahlung)

The radiative mass stopping power is approximately proportional to the electron energy E and to the square of the atomic number Z of the absorber. This comes from the fact that, according to the Larmor relationship, the intensity of bremsstrahlung radiation is proportional to the square of the electron acceleration. The $Z^2$ dependence and the $E$ dependence make the x-ray production more efficient for higher energy electrons impinging on high atomic number absorbers. We can see on Fig. 3.1 that $(S/p)_{\text{rad}}$ intersects $(S/p)_{\text{col}}$ at energies of tens of MeV in high Z media, but not until 100 MeV in water (and tissue). Thus at energies used in radiotherapy $(S/p)_{\text{rad}}$ is less than 1% of $(S/p)_{\text{col}}$ in tissue and can in many circumstances be overlooked without consequence.

3.3 Energy Specification and Measurement.

The term "beam quality" is often used as a general description of the ability of a radiation beam to penetrate through matter. With electrons this concept is related to the specification of the energy and the energy spectrum of the beam. There are many instances requiring the knowledge of the quality of a high-energy beam. Many treatment parameters such as the percent depth doses can only be correctly selected if the beam energy is known accurately. To allow a meaningful comparison of characteristics of electron beams from various treatment units, or comparisons of data and clinical results from different centers, the knowledge of the beam energy is necessary. In dosimetry, knowledge of the beam quality is essential for the choice of the energy dependent factors such as stopping power ratios used in dose calculations.

At the end of its travel through the waveguide, the electron beam reaches its maximum kinetic energy. At this stage, the beam has a very small spread in energy and angle. The energy spectrum exhibits a narrow peak which can be adequately characterized by a single energy value, referred to as the nominal energy of an electron beam. For instance, when one speaks of a 9 MeV electron beam, 9 MeV refers to the energy of the beam inside the accelerator just before hitting the accelerator window. Figure 3.2 shows the distribution of the electron energy fluence in relation to the various points in the accelerator beam geometry.
As the electron beam passes through the exit window, scattering foils, monitor chamber, air, and other materials in reaching the phantom or patient surface, it suffers energy losses which cause a shift of the electron spectrum to energies lower than the nominal energy and cause a broadening of the spectrum (see Fig. 3.2). At the phantom surface it is therefore necessary to use several parameters to characterize the distribution of the energy fluence. Two parameters are of particular interest: the energy value corresponding to the peak of the curve, which is called the "most probable energy" $E_{p,o}$ and the "mean energy" $\overline{E}_o$, which is slightly smaller than $E_{p,o}$. As seen in the previous section, when an electron beam traverses thick layers of matter, the energy losses are large. Therefore, with an increasing depth in phantom, the beam energy becomes progressively lower and the spectrum becomes broader (see Fig. 3.2). The difference between the most probable energy $E_{p,o}$ and the mean energy $\overline{E}_o$ thus increases. Inside the phantom or the body of the patient, the parameters of interest are again the
most probable energy $E_{p,z}$ and the mean energy $\bar{E}_z$, by analogy with the definitions given for the energy at the surface, where $z=0$.

In clinical practice, an electron beam is usually characterized by the energy at the body (phantom) surface. There are several methods to determine this energy, most notably: (1) measurement of threshold energy for nuclear reactions, (2) range measurements in phantoms, and (3) the measurement of Cerenkov radiation threshold. The range method is the most practical for clinical use.

**Most Probable Energy.**

The most probable kinetic energy at the phantom surface, $E_{p,0}$, in MeV is related to the practical range $R_p$ in cm by the simple relationship:

$$E_{p,0} = C_1 + C_2 R_p + C_3 R_p^2,$$

(3.1)

where $C_1$, $C_2$, and $C_3$ are constants. The practical range $R_p$ will be defined in the next section. For water, NACP\textsuperscript{8} and ICRU\textsuperscript{9} found that Eq. (3.1) with $C_1 = 0.22$ MeV, $C_2 = 1.98$ MeV/cm, and $C_3 = 0.0025$ MeV/cm\textsuperscript{2} reproduces the measured data with an accuracy of 2% from a few MeV to 50 MeV.

**Mean Energy.**

It has been shown\textsuperscript{10} that the mean energy $\bar{E}_0$ of the electron beam at the phantom surface is related to $R_{50}$ (the depth in cm at which the dose is 50% of the maximum dose) by the following relationship:

$$\bar{E}_0 = C_4 R_{50},$$

(3.2)

where $C_4$ is a constant. The value of $C_4$ has been controversial and has been studied by Rogers and Bielajew\textsuperscript{11}. The AAPM\textsuperscript{12} and NACP\textsuperscript{8} protocols recommended the value of $C_4 = 2.33$ MeV/cm independent of electron kinetic energy. However, more recent Monte-Carlo calculations\textsuperscript{11} have shown that the value of $C_4$ in the energy range of clinical interest is closer to 2.4 MeV/cm. This small change in the value of $C_4$ has little impact on clinical dosimetry.
Energy at Depth.

As the electron beam penetrates the phantom, the most probable energy and, approximately, the mean energy of the spectrum decrease linearly with depth. This can be expressed by the so-called Harder relationships:

\[ E_{p,z} = E_{p,0} [1-(z/R_p)] \] \hspace{1cm} (3.3)

and

\[ E_z = E_0 [1-(z/R_p)] \, \] \hspace{1cm} (3.4)

where \( z \) is the depth in phantom. Equation (3.4) is important in electron dosimetry, because for absorbed dose measurements it is necessary to know the mean energy at the location of the chamber.

3.4 Characteristics of Clinical Electron Beams.

Central Axis Depth Dose Curves.

The particular feature of the electron beam that makes it a unique therapeutic tool in radiotherapy application is the shape of the depth dose curve. Depth dose curves for different electron energies were presented in Fig. 1.3. Figure 3.3 below shows a typical electron beam percent depth dose curve on which all the relevant parameters used to characterize the curve are indicated.

The curve displays a moderately flat plateau in the first few centimeters of tissue, followed by a rapid fall in the absorbed dose. This distribution makes electrons eminently suitable for treatment of tumors from the skin surface up to depths of about 6 cm. The electron beam offers a more uniform dose over the target volume in comparison with a superficial x-ray treatment technique. The sharp dose cut-off beyond the 80% depth dose is present for all energies up to about 20 MeV; higher energy beams have depth dose curves with a long tail resembling depth doses for low-energy x-rays. Thus, the possibility of treating superficial tissues to relatively uniform doses while sparing underlying healthy tissues is offered by electron beams in the energy range of about 6-20 MeV.
Higher energy beams are not clinically useful, except in some special circumstances.

![Graph showing central-axis depth doses for an electron beam with parameters indicated that can be used to characterize electron beams. These parameters are discussed in ICRU Report 35.](image)

From Fig. 3.3, we can see that there is a build-up of dose which occurs between the surface and the depth $d_{\text{max}}$ where the dose reaches its maximum value $D_{\text{max}}$. The physical basis for this effect is different from the one causing it in high energy photon beams. In the case of a photon beam, absorption of energy cannot take place until secondary electrons are first produced by photon interactions with matter. At the surface, the electron fluence is small and it increases progressively as the beam penetrates beneath the surface, until it reaches the depth where the fluence of the forward moving secondary electrons reaches an equilibrium with the exponentially declining photon beam. For fast electron beams, deposition of energy starts from the very beginning of the beam entry into tissue. Consequently, the surface dose $D_s$ is much higher than it is with a high energy photon beam. For electron beams, the primary cause of the build-up is the oblique scattering of electrons as they penetrate into the tissue layers. At the time of entering the surface, the electron beam can be considered as a
parallel beam. As the electrons penetrate into tissue, their path becomes progressively more oblique owing to multiple scattering effects. This results in an increase in electron fluence and thus, an increase in the energy deposited per unit length along the beam axis. Since electron scattering is more important at lower energies, the build-up region becomes more and more pronounced as the energy of the electron beam is decreased (see Fig 1.3).

At the end of the depth dose curve, there is a long tail which is due to x-ray contamination of the electron beam. The x-ray component $D_x$ can be determined by reading off the dose value at the point where the tail becomes relatively straight (see Fig. 3.3). This x-ray dose is contributed by bremsstrahlung interactions of the electrons with the collimation system (scattering foils, chambers, collimator jaws, etc.) and the body tissues. As discussed in Section 3.2, bremsstrahlung production is more efficient at higher electron energies, thus the x-ray component $D_x$ increases with increasing electron energies.

The practical range $R_p$, introduced in the previous section, is determined from the depth dose curve as the depth at the point where the tangent at the inflection point of the falloff portion of the dose curve intersects the bremsstrahlung background. Figure 3.3 also shows the depth at which the dose is 50% of the maximum dose $R_{50}$ and the therapeutic range $R_t$. The therapeutic range is a measure of the clinically useful portion of the electron depth dose curve. It is the recommendation of the AAPM TG-25 protocol that the therapeutic range should be taken as the depth of the 90% dose level. Thus, we can see that a depth-dose (or depth-ionization) measurement allows one to obtain all the necessary parameters ($R_p$ and $R_{50}$) for the energy specification of an electron beam.

**Field Flatness and Symmetry.**

Uniformity of intensity across the electron field is a requirement for the therapeutic use of electron fields. Both the symmetry and flatness of an electron beam depend on the design of the scattering system and the electron collimation apparatus. The AAPM recommends that the flatness of an electron beam be specified in a reference plane perpendicular to the central axis, at the depth of the 95% isodose beyond the depth of dose maximum. The variation in dose relative to the central axis dose should not exceed ±5% (optimally to be within
±3%) over an area confined within lines 2 cm inside the geometric edge of fields equal to or larger than 10x10 cm².

The beam symmetry compares a dose profile on one side of the central axis to that of the other. The AAPM recommends that a beam profile in the reference plane should not differ by more than 2% at any pair of points located symmetrically on opposite sides of the central axis.

Output Factors and Depth Dose Variations with Beam Area.

As seen in Chapter 1, the dose per monitor unit (output) is a function of field size. The output factor OF(A) is defined as the ratio of the dose D per monitor unit U at d_{max} for a given field size A to that for the reference field size A_o at its own d_{max,o}. Then we have:

\[ \text{OF}(A) = \frac{D(U, d_{\text{max}})}{D(U, d_{\text{max,o}})} \]  \hspace{1cm} (3.5)

The output factor decreases with decreasing field size. Attention must be given to the fact that, for a given energy, the depth of maximum dose may change with field size. In general, if the beam area is large, above about 6x6 cm², the depth doses show very small dependence on field size. However, if the cross section of the beam is small compared with the electron range in the medium, the depth doses may vary considerably with field size, and the build-up region tends to shrink. This may be explained by the fact that in a narrow beam the electrons are scattered away from the beam central area with no compensation by scattering toward the axis from adjacent regions, since these regions are not irradiated under narrow beam conditions. This is an important consideration when the electron beam is further collimated by custom made cut-outs, as discussed below.

Beam Shaping.

The collimation of electron beams is primarily accomplished by means of applicators (cones) supplied with the accelerator. These cones define a range of square or circular beams. However, electron fields of irregular shapes are often required and these can be produced by making shielding masks or cut-outs.
which are attached to the end of the applicator. Two materials are employed most frequently for the fabrication of these cut-outs: lead and Lipowitz’s metal. Lipowitz’s metal is a low melting point alloy known under a variety of trade names, including Cerrobend and Ostalloy 158. Cerrobend is the material of choice for electron beam shaping because it may easily be re-melted after use and moulded into a new field-defining shield. The use of these shielding masks may significantly modify the output of the beam, especially when the aperture is small. Thus, in addition to the output factor associated with a given electron applicator, one has to measure an output factor associated with the output reduction introduced by the use of the shield.

3.5 Conclusions.

This chapter has discussed some important topics related to the physics of electron beams. The different interactions of electrons with matter have been reviewed. We have seen that the energy spectrum of an electron beam changes gradually and continuously with increasing depth in the medium. This feature of electron beams is of great importance when performing electron beam dosimetry as will be seen in the next chapter.

3.6 References.


4.1 Introduction.

This chapter provides an introduction to the basic concepts of absolute and relative electron dosimetry. The most important methods employed in electron dosimetry are reviewed and the theory behind absolute dose measurements with an ionization chamber is discussed. Particular attention is given to film dosimetry because of the extensive use of this modality in the work presented in this thesis. Many important aspects of film dosimetry are discussed by either theoretical considerations or through experiments performed by the author. The reliability of film dosimetry for the present application is demonstrated.

4.2 Definition of Exposure and Dose.

The ICRU defines exposure $X$ as the quotient $dQ/dm$, where $dQ$ is the absolute value of the total charge of the ions of one sign produced in air when all the electrons liberated by photons in air of mass $dm$ are completely stopped in air. Originally, the exposure was measured in esu per cm$^2$ of air (at STP, standard...
temperature 0°C, and standard pressure 101.3 kPa), but in 1928, 1 esu per cm³ of air at STP was assigned the name Roentgen (R). The SI units for exposure are coulomb per kilogram (C/kg), and we have:

$$1 \text{ R} = 2.58 \times 10^{-4} \text{ C/kg air.}$$

As stated in the definition, exposure applies only to photons, and, for reasons of a practical nature, it is a concept used only for photon energies below 3 MeV. Although not directly applicable to electron beams, the exposure is an important concept in electron dosimetry because most dosimetry protocols start from an exposure calibration factor to determine the absorbed dose in a medium with an ionization chamber. This will be discussed in more detail later.

As defined by the ICRU¹, the dose D, or absorbed dose, is the quotient dE/dm, where dE is the amount of energy imparted by ionizing radiation per unit mass dm of the absorbing medium. The size of the mass that is considered in the definition is small enough to represent the dose at a point, but large enough to average out statistical fluctuations. The dose is the most important quantity in radiation dosimetry because it describes the quantity of radiation for all types of ionizing radiation, including indirectly and directly ionizing radiations; all materials; and all energies. It is also the best predictor of the biologically significant effects produced by ionizing radiation. The old unit of dose is the rad (an acronym for radiation absorbed dose) and represents the absorption of 100 ergs of energy per gram of absorbing material:

$$1 \text{ rad} = 100 \text{ erg/g} = 0.01 \text{ J/kg.}$$

The SI dose units are J/kg, and a special unit, the Gray (Gy), has been introduced because of its fundamental role in radiation physics and medicine. Thus:

$$1 \text{ Gy} = 1 \text{ J/kg}$$

and

$$1 \text{ Gy} = 100 \text{ rad with } 1 \text{ rad} = 0.01 \text{ Gy} = 1 \text{ cGy.}$$
4.3 Relative vs. Absolute Dose Measurements.

There are two distinct areas of interest in electron dosimetry which require somewhat different approaches: determination of the spatial beam distribution which is based on relative measurements and absorbed dose calibration which requires absolute methods. In relative dosimetry one measures the dose at any point in the medium relative to the dose at some other point which serves as reference. Relative dosimetry is employed in the measurement of PDDs or beam profiles, as discussed in Chapter I. When performing relative electron dosimetry with an ion chamber, one has to account for the change in the energy spectrum of electrons with depth. This point will be further discussed latter. All of the work presented in this thesis is based on relative dose measurements.

Although relative dosimetry is suitable for many purposes, the treatment of patients requires the exact knowledge of the output of the treatment unit. This means that we must be able to determine accurately the absolute dose at the reference point in units that are clearly understood and reproducible. This is a necessary requirement for the probabilities for tumor control and radiation induced complications to be recorded in a universal manner, and to use this experience to improve the success of radiotherapy. The following sections describe the most important methods used in absolute and relative electron dosimetry.

4.4 Methods Employed in Electron Dosimetry.

There are three methods for determining the absolute dose in a medium: 1) calorimetry, 2) chemical (Fricke) dosimetry, and 3) measurement of the ionization produced in an air-filled chamber. The most important relative dosimetry techniques are: 1) film dosimetry, 2) thermoluminescent dosimetry, and 3) dosimetry using ionization chambers or semiconductor detectors. Ionization chambers and film dosimetry were used extensively in our measurements so they are the subject of sections of their own below. Several other less important techniques will now be briefly reviewed.

Calorimetry.

Calorimetry is based on the principle that the energy absorbed in a medium
when exposed to radiation appears ultimately as heat, while a small amount may appear in the form of a chemical change. This results in a small increase in the temperature of the absorbing medium which, if measured accurately, can be related to the absorbed dose. Calorimetry is the most fundamental means of determining the absolute dose to a medium, but this conceptually simple method is difficult to use in practice because of the very small temperature changes involved. As an example, the absorption of 1 Gy in water (which has a specific heat of $C_p = 4185 \text{ J/kg}^\circ\text{C}$) will produce a rise in temperature ($\Delta T$) of:

$$\Delta T = \frac{D}{C_p} = \frac{1 \text{ J/kg}}{4185 \text{ J/kg}^\circ\text{C}} = 2.39 \times 10^{-4} \circ\text{C}.$$ 

Thus, more than 4000 Gy would be required to produce a temperature rise in water of 1 °C. Considering that routine clinical measurements require the dose to be precisely determined down to less than a fraction of one Gy, it is obvious that calorimetry is extremely difficult to use in clinical practice.

**Chemical Dosimetry (Fricke dosimetry).**

Chemical dosimetry makes use of compounds which, when irradiated, undergo some easily detectable chemical change. If this change can be precisely quantified, it can be used as a measure of absorbed dose. Many systems of chemical dosimetry have been proposed, but by far the most widely used is the Fricke dosimeter, due to the high accuracy and reliability offered compared with other systems. The method is based on the determination of the yield of ferric ions produced during irradiation of the Fricke ferrous sulphate solution:

$$\text{Fe}^{2+} \rightarrow \text{Fe}^{3+}.$$ 

The ferric ion concentration is determined by spectrophotometry of the dosimeter solution (or gel), which shows absorption peaks in the ultraviolet (UV) light at wavelengths of 224 and 304 nm. Although not visible to the eye, a dose of a few Gy will give a detectable change in UV transmission. To make the UV transmission method absolute, the only parameter which has to be known is the radiation chemical yield ($G$ value) defined as the number of ferric molecules produced per 100 eV of absorbed energy. Recently, magnetic resonance has also been used for imaging the relative $\text{Fe}^{3+}$ concentration in a Fricke gel, but this approach cannot yet be used for absolute dosimetry.
Fricke dosimetry has the advantage that the measuring medium has radiological properties very similar to those of tissue, and that it can be put into any desired shape. However, this method, like calorimetry, requires sophisticated experimental procedures as well as high radiation doses to achieve a sufficient reproducibility and accuracy. Therefore, in clinical practice, the only practical option for absolute dosimetry is the measurement of the ionization produced in an ionization chamber.

Thermoluminescent Dosimetry.
Thermoluminescent dosimetry (TLD) makes use of solid crystals (phosphors) which, when irradiated, store a small fraction of the absorbed energy in the crystal lattice. Some of this energy can be recovered later as visible or UV light if the material is heated. Measurements are performed by placing the irradiated material in a heater cup where it is heated through a reproducible heating cycle. The thermoluminescent light emission is measured by a photomultiplier tube which converts visible or UV photons into an electrical current.

Several thermoluminescent phosphors are commercially available but the most widely used is lithium fluoride (LiF). In its purest form LiF exhibits relatively little thermoluminescence, but the presence of a small amount of impurities (e.g., magnesium) gives rise to imperfections in its lattice structure and provides the radiation induced thermoluminescence. These impurities create energy traps in the forbidden region between the valence band and the conduction band of the crystal, thus creating metastable states for charge carriers freed by the ionizing radiation.

Under irradiation, some electrons in the valence band receive sufficient energy to be raised to the conduction band. These electrons can either fall directly back into the valence band and recombine with a positive hole or they can get trapped in the metastable states. The instantaneous emission of light associated with the immediate transitions to the valence band is called fluorescence. The electrons trapped in the metastable states require energy to get out of the trap and fall back to the valence band. The light emitted in this case is called phosphorescence (delayed fluorescence). In TLD, the energy needed to move the electrons out of the metastable states is provided by the reproducible heating cycle through which the irradiated material is heated. LiF phosphorescence is
very low at room temperature, but it is speeded up significantly with a moderate amount of heating, generating the so-called thermoluminescence (phosphorescence induced by thermal energy). The total amount of light emitted is proportional to the number of electrons that were trapped; this in turn is proportional to the amount of energy absorbed from the radiation. Thus, the measured thermoluminescence is directly proportional to the dose absorbed in the TLD crystal.

Since the response of thermoluminescent dosimeters (TLDs) is affected by their previous radiation and thermal history, the phosphor material must be suitably annealed to remove residual effects. The standard preirradiation annealing procedure for LiF is 1 h of heating at 400°C and then 24 h at 80°C. When considerable care is used, precision of approximately 3% may be obtained using TLDs. Although not as precise as the ion chamber, the main advantage of TLD is in measuring doses in regions where an ion chamber cannot be used. For example, TLD is extremely useful for in vivo dosimetry by direct insertion into tissues or body cavities. Since TLD material is available in many forms and sizes, it can be used for special dosimetry situations such as for measuring surface dose.

4.5 Ionization Chambers for Dose Measurements.

An ionization chamber is an instrument designed to measure the quantity of radiation at a certain location by collecting charges released through the action of the radiation field. These charges are mainly released in an air cavity within the chamber which constitutes the sensitive volume of the chamber. Ion chambers are routinely used in measurements of photon or electron beams in the energy range from 1 keV to 50 MeV. An ion chamber is used with an electrometer and a power supply which serve the dual purpose of polarizing the electrodes surrounding the air cavity and of measuring the collected charge which is proportional to the total radiation received by the chamber. The ion chamber and electrometer form a closely-linked pair and are seldom used separately. The popularity of ion chambers is due to the stability of performance and high precision of measurement attainable in comparison with other dosimeters. An ion chamber, connected to a high-quality electrometer, can be expected under identical irradiation conditions to have a response reproducible
to within ±0.05%.

**Ionization Chamber Design.**

There are two main types of chamber designs widely used in electron beam dosimetry: the Farmer-type chamber and the parallel-plate chamber. The Farmer chamber has a cylindrical wall (outer electrode) which encloses an unsealed air cavity volume of typically 0.6 cm³. The collecting electrode is a rod of low atomic number material held in the center of the cavity. The wall, air cavity, and inner electrode constitute what is known as the thimble. This is the collecting portion of the chamber. The design of the commercially available Farmer chambers varies with respect to the composition of the wall or the central electrode material, but the chambers are often made of tissue-equivalent conducting plastics or graphite. High potential (typically 300 V) is applied between the chamber electrodes and it is essential that the insulation between electrodes is of the highest quality to minimize any possible leakage current.

The parallel-plate chamber has an air cavity in the shape of a circular disk. Typically, the parallel electrodes have diameters on the order of 1-2 cm, and they are spaced by about 1-2 mm. The upper electrode (facing the source) is also called the beam entrance window. This electrode is usually very thin (e.g., foils of 0.01 to 0.03 mm thick Mylar, polystyrene, or mica) and allows measurements practically at the surface of a phantom without significant wall attenuation. The lower electrode is the collecting electrode. In some parallel-plate chambers the collecting electrode is mounted on a thin layer of insulating material to reduce direct electron capture, thus giving a negligible polarity effect. An important component of the chamber is the guard ring, which is a grounded conducting ring placed so that it surrounds the collecting electrode. The guard ring reduces the voltage drop occurring across the insulator and thus minimizes the measured leakage currents. It also assists in better definition of the chamber sensitive volume.

4.6 Absolute Dosimetry with Ionization Chambers.

There are many dosimetry protocols which have been produced over the years to provide radiological physicists with a systematic method for determining the absolute dose to water from high-energy photon or electron beams used for
radiation therapy. Most current protocols, such as the AAPM TG-21 protocol\textsuperscript{2} for photon beams or the AAPM TG-25 protocol\textsuperscript{3} for electron beams, recommend using ionization chambers as the measuring instrument and start from an exposure calibration factor. These chambers are used as cavity chambers in a phantom and Spencer-Attix formulation of the Bragg-Gray cavity theory is applied to determine the absorbed dose. This constitutes a major conceptual change with respect to older protocols\textsuperscript{4,5,6} where the ionization chambers were considered to be exposure meters.

As stated previously, all of the work presented in this thesis is based on relative dose measurements. Thus, the intention here is not to provide an in-depth discussion of dosimetry protocols, but to present the fundamentals of radiation dosimetry and their application to electron dosimetry according to the AAPM TG-25 protocol\textsuperscript{3}. In the AAPM protocols, calibration is achieved with an ionization chamber of either cylindrical or parallel-plate design used at depth in water, polystyrene, or Lucite phantoms in a four step procedure. The first step is to measure the ionization in the chamber air. The next three steps are accomplished with the aid of cavity theory. The absorbed dose to air in the chamber is then calculated. From that, the absorbed dose to the phantom in which the ion chamber resides is calculated, and finally the absorbed dose to the reference medium (water) is calculated.

\textit{Bragg-Gray Cavity Theory.}

The theoretical basis for the AAPM protocols is the Spencer-Attix formalism\textsuperscript{7} of the Bragg-Gray cavity theory\textsuperscript{8,9}. The theory itself is quite simple but complications arise because most chambers are not ideal Bragg-Gray cavities. Let us first consider a region of otherwise homogeneous medium w containing a thin layer, or cavity filled with another medium g (see Fig. 4.1).

Two Bragg-Gray conditions apply:
1) The thickness of the g-layer is assumed to be so small in comparison with the range of the electrons striking it that its presence does not perturb the electron fluence.

2) The absorbed dose in the cavity is assumed to be deposited entirely by the electrons crossing it.
Figure 4.1  Electrons of fluence \( \Phi \) crossing a thin layer of material \( g \) in an otherwise uniform medium \( w \).

When the two Bragg-Gray conditions hold, the electron fluence \( \Phi \) does not change and the ratio of the dose to medium \( w \) to that in medium \( g \) is given by:

\[
\frac{D_w}{D_g} = \frac{\overline{\frac{S}{\rho}}^w}{\overline{\frac{S}{\rho}}^g} = \left( \frac{\overline{S}}{\rho} \right)^w_g ,
\]

where the bar over the collision stopping powers implies that an average value of the collision stopping power over the incident electron spectrum is taken.

Now, if we consider an ion chamber with its walls placed within a homogeneous medium, the specific charge \( J_{\text{gas}} \) in the air will be given by the charge \( Q \) produced in air divided by the mass \( m_{\text{air}} \) of the air in the chamber cavity:

\[
J_{\text{gas}} = \frac{Q}{m_{\text{air}}}.
\]

The dose to air is then the specific charge multiplied by the energy that is
required, on average, to create an ion pair in air. This factor \( (W_{\text{air}}) \) was determined experimentally and is constant over a great range of air pressures and electron/photon energies (at least from 10 keV to 50 MeV). The AAPM protocol uses a value of 33.7 J/C for air at 50% humidity, but more recent evaluations\(^{10}\) suggest that \( W_{\text{air}} \) is equal to 33.97 J/C. Thus,

\[
D_{\text{air}} = J_{\text{ea}} W_{\text{air}} \quad \text{(4.3)}
\]

If we assume that the cavity of the chamber is small, and that the walls are very thin, then we can use Eq. (4.1) to write:

\[
\frac{D_{\text{med}}}{(\bar{S}/\rho)_{\text{med}}} = \frac{D_{\text{wall}}}{(\bar{S}/\rho)_{\text{wall}}} = \frac{D_{\text{air}}}{(\bar{S}/\rho)_{\text{air}}} \quad \text{(4.4)}
\]

and therefore, using Eq. (4.3) and Eq. (4.4),

\[
D_{\text{med}} = D_{\text{air}} (\bar{S}/\rho)_{\text{med}}^{\text{med}} = J_{\text{ea}} W_{\text{air}} (\bar{S}/\rho)_{\text{air}}^{\text{med}} \quad \text{(4.5)}
\]

Equation (4.5) represents the Bragg-Gray formulation of the dose to a medium.

**Spencer-Attix Formulation of the Bragg-Gray Relationship.**

The Bragg-Gray cavity theory fails to account for the very energetic secondary electrons (delta rays) generated by the primary electrons. An ideal Bragg cavity must be small so that it does not perturb the radiation field within the medium, but it should be large enough so that the secondary electrons created can lose all of their energy in the cavity. In fact, there is an appreciable fraction of the delta rays generated in the cavity that will have ranges in air that are large compared with the cavity size, and will dissipate some of their energy in the medium rather than in air. In order to resolve this apparent dilemma Spencer and
Attix have divided the secondary electrons in two groups, separated at a cutoff energy $\Delta$ that corresponds approximately to the energy of an electron that can just cross the cavity. A "slow" secondary electron, with energy less than $\Delta$, is considered to dissipate its energy at the point where it is generated, while a "fast" secondary electron, with energy greater than $\Delta$, is included in the spectrum of primary electrons. The stopping powers calculated with this added condition are called restricted stopping powers, and they are used in the AAPM protocols to relate the dose in the air to the dose in the medium. In the AAPM protocols, $\Delta \approx 10$ keV was chosen. Thus, with the Spencer-Attix formalism, Eq. (4.5) is written as:

$$D_{med} = D_{air} \frac{\overline{I}}{\rho}_{air}^{med} = J_{sat} W_{air} \frac{\overline{I}}{\rho}_{air}^{med}, \quad (4.6)$$

where $\frac{\overline{I}}{\rho}$ represents the restricted mean mass collision stopping power.

Chamber Specific Corrections and Influence of Environmental Conditions.

The theory for conversion of ionization measured in an air cavity to absorbed dose to the medium is fairly simple. The complications come in trying to introduce traceability to national standards and in trying to apply the theory to the practical situations where cavities have finite dimensions and non-homogeneous construction (i.e., walls) and phantom materials other than water are used. Conversion to dose to water involves phantom-specific corrections that will be considered later.

For electron beams, the AAPM protocol recommends two chamber-specific corrections: $P_{ion}$ and $P_{repl}$. If all of the charge generated in the ionization chamber is not collected, the dose calculated from Eq. (4.6) will be too small. Therefore, a correction is required for the electrons and positive ions which recombine prior to being swept through the chamber and measured as part of the charge released. Techniques for determining $P_{ion}$, the correction factor for this effect, are based on measuring the chamber response at two different voltages. $P_{ion}$ varies with dose rate, chamber geometry, and collection voltage and must be established for each calibration situation. For pulsed beams, which are characteristic of linacs,
Weinhous and Meli\textsuperscript{11} proposed the following relationship for $P_{\text{ion}}$:

$$P_{\text{ion}} = a_0 + a_1 \frac{Q_1}{Q_2} + a_2 \left( \frac{Q_1}{Q_2} \right)^2,$$  \hspace{1cm} (4.7)

where $Q_1$ and $Q_2$ are the charges collected for two different applied voltages $V_1$ and $V_2$, respectively. The $a_i$ coefficients are given for several ratios of applied biases and can be interpolated for other values.

The second chamber-specific correction ($P_{\text{repl}}$) accounts for the changes in the electron spectrum caused by the insertion of the cavity into the medium. The replacement factor ($P_{\text{repl}}$) is less than unity and its magnitude should depend on the size of the chamber and the gradient of the dose at the point of measurement. For electrons, since the AAPM protocol only applies for measurements at $d_{\text{max}}$ where, by definition, the gradient of the dose is zero, the component of $P_{\text{repl}}$ which accounts for the dose gradient is unity, and there is only an electron fluence component to the correction.

Since the cavity of ionization chambers is unsealed, the mass of air within the cavity will vary according to environmental conditions, affecting the charge released within the chamber. However, the chamber is calibrated at a standards laboratory for specific conditions: 22 °C and 101.3 kPa. It is therefore necessary to correct the mass of the air within the chamber to these conditions when performing absolute dosimetry. Thus, the readings taken with the chamber must be multiplied by a temperature and pressure correction factor:

$$K_{TP} = \left( \frac{273 + T}{295} \right) \left( \frac{1013}{P} \right),$$  \hspace{1cm} (4.8)

where $T$ and $P$ are the ambient air temperature (in °C) and pressure (in kPa), respectively at the moment of performing the measurement.

Incorporating all the correction factors described above into Eq. (4.6), we now have:

$$D_{\text{med}} = J_{\text{gas}} W_{\text{air}} \left( \frac{\bar{E}}{\rho} \right)_{\text{med}} P_{\text{repl}} P_{\text{ion}}.$$  \hspace{1cm} (4.9)
The correction factor for environmental conditions \(K_{TP}\) is integrated into the specific charge \(J_{gas}\).

**Calibration of the Chamber and the \(N_{gas}\) Concept.**

A serious obstacle in the use of Eq. (4.9) is that the mass of air \(m_{air}\) must be accurately known in order to calculate the specific charge \(J_{gas}\). The problem arises because one needs the collection volume of the chamber to compute \(m_{air}\). The collection volume is the one encompassed by the electric field lines between the electrodes. This volume, which might differ from the physical volume, is very difficult to assess accurately in the small, practical chambers used for clinical dosimetry.

In practice the chamber is calibrated against another chamber whose construction lends itself to an accurate determination of the effective mass of air from which charges are collected. This is performed in national calibration laboratories, and institutions send one chamber with or without an electrometer to these laboratories for calibration. The calibration laboratory issues an exposure calibration factor \(N_x\) in R/(scale division) or R/nC. The first applies to a combined chamber/electrometer calibration, the second to a calibration of the chamber alone. The calibration laboratory performs the calibration in air with a cobalt-60 beam, and the calibration factor is valid only for these conditions. In Canada, the calibration service is provided by the National Research Council located in Ottawa. Each institution performing radiotherapy usually has one chamber and electrometer designated to serve as a dosimetry reference and referred to as the institutional secondary standard. The other instruments used in daily clinical practice are calibrated against the secondary standard, which in turn is periodically calibrated in a standards laboratory (typically every two years).

The strategy for absolute dose determination is to obtain from a calibration laboratory the \(N_x\) factor at cobalt energy, and extend its use to higher energy photon beams and to electron beams. This procedure is based on the reasonable assumption that the dose to air per electrometer reading is a property of the chamber/electrometer combination and does not depend on the energy or spectrum of the beam. The AAPM protocol defines this quantity to be \(N_{gas}\):
where $M$ is the electrometer reading corrected to standard conditions ($22^\circ\text{C}$ and 101.3 kPa). According to the previous discussion, we now have:

$$D_{\text{air}} = J_{\text{air}} W_{\text{air}} = MN_{\text{gas}} W_{\text{air}} P_{\text{ion}} .$$

(4.11)

in a cobalt-60 beam. Considering the Equations (4.9), (4.10) and (4.11) we obtain that the dose to the medium is given by:

$$D_{\text{med}} = MN_{\text{gas}} \frac{L}{\rho_{\text{air}}} P_{\text{med}} P_{\text{ion}} .$$

(4.12)

Thus, $N_{\text{gas}}$ is an energy independent air-cavity calibration factor. It is the method chosen by the AAPM protocol to allow the user to trace the chamber calibration to the national standards for exposure. $N_{\text{gas}}$ is evaluated at the Co-60 energy, since that is the only energy for which we have an exposure calibration factor. A rigorous derivation of $N_{\text{gas}}$ has been given by Attix\textsuperscript{3}. The formula for $N_{\text{gas}}$ is given by Equation #6 in the AAPM TG-21 protocol\textsuperscript{2}.

4.7 Phantom Materials.

The aim of clinical dosimetry is to determine the dose delivered to a variety of tissues. However, since it is not possible in practice to measure dose distributions directly in patients, alternative materials are required for routine measurements. It is important that the radiological properties of these materials be accurately known: and that they are as similar as possible to those of tissue.

Basic dose distribution data are usually measured in a water phantom which closely approximates the radiation absorption and scattering properties of muscle and other soft tissues. Water has the advantage of being readily available and has reproducible radiation properties. It holds a special place in dosimetry, since all protocols recommend it as the medium to which dose calibration be referred. However, a water phantom poses some practical problems when used in conjunction with ion chambers and other detectors (such
as films) which are affected by water, unless they are designed to be waterproof. Furthermore, the set-up of a water phantom is time-consuming and solid materials are more convenient for use in measurements at a given point. For these reasons, solid dry phantoms have been developed as substitutes for water. Ideally, for a given material to be tissue or water equivalent, it must have the effective atomic number, number of electrons per gram, and mass density identical to those of tissue or water. For electron beam dosimetry, matching of mass stopping powers and angular scattering powers is necessary.

The most widely used solid phantom materials are polystyrene \((\text{C}_9\text{H}_8)_n\) and polymethylmethacrylate \((\text{C}_5\text{H}_8\text{O}_2)_n\) abbreviated as PMMA and available commercially as Acrylic, Lucite, Plexiglas, or Perspex. Polystyrene is more popular due to its close tissue-equivalence and outstanding resistance to radiation damage. Polystyrene and Acrylic are the two substitute materials recommended by the AAPM TG-21 and AAPM TG-25 protocols.

Plastic phantoms are not perfectly water equivalent. Thus, when converting dose distributions measured in solid phantoms to dose in water, it is necessary to apply some phantom-specific corrections. According to the AAPM TG-25 protocol, in electron dosimetry the conversion of dose measured in a given medium \((D_{\text{med}})\) to dose in water \((D_{\text{water}})\) is accomplished through the following relationship:

\[
D_{\text{water}}(d_{\text{water}}) = D_{\text{med}}(d_{\text{med}}) \left\{ \frac{3}{\rho} \right\}_{\text{coll}} \left\{ \phi \right\}_{\text{med}} \left\{ \phi \right\}_{\text{water}} ,
\]

where \(\left\{ \frac{3}{\rho} \right\}_{\text{coll}} \) is the ratio of the mean unrestricted mass collision stopping power in water to that in the medium and \(\left\{ \phi \right\}_{\text{water}}\) is the ratio of the electron fluence in water to that in the medium. \(\left\{ \phi \right\}_{\text{med}}\) is an energy-dependent factor with values ranging from 1 to 1.047 for clear polystyrene\(^3\). The water-equivalent depth can be approximated using a density determined from the ratio of \(R_{50}\) penetrations by:

\[
d_{\text{water}} = d_{\text{med}} \times \rho_{\text{eff}} = d_{\text{med}} \times \left( \frac{R_{\text{water}}}{R_{50}} \right) .
\]
The recommended values for $p_{eff}$ are 0.975 for polystyrene and 1.115 for PMMA\textsuperscript{3}.

When performing relative dosimetry, the corrections given in Eq. (4.13) can be neglected since their values change slowly with energy (and thus with depth). However, in order to achieve high accuracy with calibration measurements, the dose to the medium ($D_{med}$) given by Eq. (4.12) must be corrected according to Eq. (4.13) when calibration is performed in a non-water phantom.

4.8 Relative Dosimetry with Ionization Chambers.

There is a major difference between photon and electron beams with respect to relative dosimetry using an ion chamber. For photon beams, the electrons set in motion in the medium have basically the same average energy at all depths. Therefore, the energy dependent parameters relating ionization to dose are constant with depth in the medium. Thus, for photon beams, all that is required is the ratio of the charge collected by the chamber at the point of interest to the charge collected at the reference point. The ratio of the charges collected gives the ratio of the doses.

In the case of electron beams, as we have seen in Section 3.3 (see Eq. (3.4)), the mean energy of the electrons decreases rapidly with depth. Thus, in the measurement of depth dose curves, for example, one has to account for the change with depth of the energy dependent parameters relating ionization to dose. For electron beams, a general equation for obtaining PDD in water from measurements in any medium is given by\textsuperscript{3}:

$$\frac{PDD}{100} = \left( \frac{\{M \times \left( \frac{\bar{E}}{\rho} \right)_{\text{water}} \times \phi_{\text{water}} \times \text{P}_{\text{repl}} \} \ d}{\{M \times \left( \frac{\bar{E}}{\rho} \right)_{\text{air}} \times \phi_{\text{water}} \times \text{P}_{\text{repl}} \}_{\text{MAX}}} \right), \ (4.15)$$

where the denominator equals the value of the numerator at the depth of maximum dose. However, as long as two measurements are taken at the same depth, where the mean energy of the electrons is the same, the ratio of the charges collected gives the ratio of the doses.
4.9 Film Dosimetry.

Film dosimetry is used extensively as a convenient and rapid means of measuring dose distributions of therapeutic electron beams. This method has been described in detail by several authors. Some of the characteristics which make film attractive in dosimetry are its high spatial resolution, its ability to provide a permanent record of dose distributions, and the short measurement time it requires. Indeed, film has the capability of recording a large volume of dosimetric information in one exposure. Furthermore, film offers the possibility of dose measurements in full planes in heterogeneously composed solid phantoms, and in solid phantoms with irregular or curved surfaces.

Although film dosimetry seems to be the method of choice for obtaining relative dose distributions of clinical electron beams, the whole sequence of dose distribution acquisition by means of film comprises many pitfalls. In order to obtain accurate and reproducible results, many precautions have to be taken. This section will provide guidelines to achieve high accuracy in relative dose measurements with film dosimetry.

**Physical Characteristics of Film and Film Processing.**

X-ray film commonly consists of a thin layer of radiation sensitive and light sensitive emulsion (~0.01 mm) coated on both sides of a cellulose or polyester base (~0.2 mm). Firm attachment between the emulsion layer and the film base is achieved by a thin layer of adhesive. The delicate emulsion is protected from mechanical damage by gelatin layers of negligible thickness known as the supercoating. The emulsion consists of small grains (1-2 μm) of silver halide suspended in gelatin. The gelatin keeps the silver halide grains well dispersed and prevents their clumping. The halide is mainly composed of silver bromide (AgBr), and small amounts of silver iodide (AgI) are added to raise the sensitivity of the emulsion.

During irradiation, electrons transfer energy to the emulsion and cause the freeing of electrons from bromide ions. These electrons move in the crystal lattice until they are captured and temporarily fixed. The negative charges thus created attract the mobile interstitial Ag⁺ ions and neutralize them to form silver atoms. These atoms then act as electron traps for a second electron which will
cause the migration of a second silver ion that will also be neutralized. Growth of silver atoms continues by repeated trapping of electrons, followed by their neutralization with interstitial silver ions. The negative bromine ions that have lost electrons are converted into neutral bromine atoms, which leave the crystal and are taken up by the gelatin of the emulsion. The small clumps of silver atoms thus created form latent image centers, which are sites at which the developing process will cause visible amounts of black metallic silver to be deposited.

The difference between an emulsion grain that will react with the developing solution and thus become a visible silver deposit and a grain that will not be developed is the presence of one or more latent image centers in the grain. The more silver atoms that exist at a latent image center, the greater the probability that the grain will be developed. Development amplifies the latent image by a factor of millions by reduction of the silver ions, which change into black metallic silver. An entire grain is developed (reduced) once the process begins. The silver in a grain that does not contain a latent image center can be reduced, but at a much slower rate. Thus, the degree of blackening of an irradiated film is not only dependent on the amount of energy (dose) the film has received by the electron beam, but it also depends in a complicated manner on the development or processing conditions. Many parameters, such as the development time or the temperature and the efficiency (activity) of the developing solution, have a direct influence on the degree of blackening that the film acquires. In order to achieve high accuracy with film dosimetry, these parameters have to be well controlled. We will see later how a quality control program can be implemented to ensure that the film processor will provide reproducible results.

Optical Density and the Sensitometric Curve.
The dosimetric information contained in an irradiated film (its degree of blackening) is analyzed by light transmission measurements. Optical density (OD), or simply density, expresses quantitatively the degree of blackening due to radiation exposure. It is defined as:

$$OD = \log_{10}\left(\frac{I_0}{I_T}\right),$$  \hspace{1cm} (4.16)
where $I_0$ is the unattenuated light intensity from the light source and $I_T$ is the light intensity transmitted through the film. A processed film that has not been exposed to radiation will still have a small OD. This background (fog) density is due in part to the small amount of unexposed grains of silver halide which are developed during film processing and to the opacity of the film base layer. This fog density must always be subtracted when performing film dosimetry to avoid large errors in regions of low optical density. Thus, the quantity of interest in film dosimetry is the net optical density which is defined as:

$$OD_{NET} = OD - FogDensity$$  \hspace{1cm} (4.17)

It is not recommended to acquire the background density from the edge of an exposed film outside the irradiated region, since leakage radiation may lead to an overestimation of the fog. For better accuracy in fog subtraction, an unirradiated film from the same batch should be processed. From here on in this thesis unless otherwise specified, the optical density quoted is always meant to be the net optical density, i.e., optical density with fog subtracted.

The dependence of the optical density on the absorbed dose to the medium in the absence of film is expressed by a sensitometric curve (see Fig. 4.2). The shape of the sensitometric curve depends mainly on the film emulsion, but also depends on the processing conditions and the applied film batch. The sensitivity can vary slightly from batch to batch even for the same brand of film. For this reason, a sensitometric curve was measured for each batch of film used in the work presented in this thesis. This was achieved by irradiating pieces of film at the depth of maximum dose in a polystyrene phantom with different numbers of monitor units. The film used in this study was the Kodak XV-2 therapy localization film. The film pieces were placed perpendicularly to the beam axis at the center of the field. Irradiations were performed with the phantom placed at the nominal SSD of 100 cm and with a 15x15 cm$^2$ electron applicator (which is the reference field size used for electron beam calibration at our institution). The electron beam energy was 12 MeV. For these measurements, the calculation of the dose received by the film pieces was based on the calibration of the linear accelerator which gives 1 cGy/MU at a depth $d_{max}$ of 2.5 cm in water. The dose in polystyrene was calculated according to Eq. (4.13). The optical density for the film pieces was measured with the densitometer provided with the Wellhöfer
dosimetry system (WP600) to be described later. To ensure compatibility with other measurements performed in this study, the film pieces were irradiated within their envelope.

Sensitometric curves for two different film batches are presented in Fig. 4.2, showing that the response of our film is linear with dose up to about 45 cGy. In this linear portion of the curve the net optical density can be used for relative dose measurements without any corrections. Hence, unless otherwise specified, 40 monitor units (40 cGy at the depth of maximum dose) were used to irradiate films in this study. This ensures that isodensity curves can be considered as isodose curves.

![Figure 4.2 Measured sensitometric curves for two different batches of Kodak XV-2 films. The straight line is a linear fit to the points up to a dose of 40 cGy.](image)

To be able to rely on the measured sensitometric curve, the processing conditions have to be well controlled. This was verified by monitoring the film processor (AGFA Curix 160) during the course of this study. Verification was achieved with the help of a sensitometer (Nuclear Associates, Model 07-417), which is a device that exposes film with a known quantity of light through a 21 step light modulator. The sensitometer is designed to make highly reproducible
exposures on x-ray film. By processing exposed films at periodic intervals and comparing the measured optical densities of the different steps confidence was obtained that the processor provides reproducible results. The reproducibility of the processor functioning is illustrated in Table 4.1 which gives the measured optical densities of the different steps for a typical period of monitoring.

<table>
<thead>
<tr>
<th>Step #</th>
<th>21-Jun-95</th>
<th>22-Jun-95</th>
<th>25-Jun-95</th>
<th>26-Jun-95</th>
<th>27-Jun-95</th>
<th>29-Jun-95</th>
<th>4-Jul-95</th>
</tr>
</thead>
<tbody>
<tr>
<td>9</td>
<td>0.16</td>
<td>0.17</td>
<td>0.16</td>
<td>0.16</td>
<td>0.17</td>
<td>0.15</td>
<td>0.17</td>
</tr>
<tr>
<td>10</td>
<td>0.17</td>
<td>0.18</td>
<td>0.17</td>
<td>0.17</td>
<td>0.18</td>
<td>0.16</td>
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</tr>
<tr>
<td>11</td>
<td>0.20</td>
<td>0.18</td>
<td>0.18</td>
<td>0.19</td>
<td>0.19</td>
<td>0.18</td>
<td>0.19</td>
</tr>
<tr>
<td>12</td>
<td>0.21</td>
<td>0.22</td>
<td>0.21</td>
<td>0.22</td>
<td>0.22</td>
<td>0.20</td>
<td>0.22</td>
</tr>
<tr>
<td>13</td>
<td>0.25</td>
<td>0.26</td>
<td>0.25</td>
<td>0.26</td>
<td>0.27</td>
<td>0.25</td>
<td>0.26</td>
</tr>
<tr>
<td>14</td>
<td>0.33</td>
<td>0.33</td>
<td>0.32</td>
<td>0.33</td>
<td>0.34</td>
<td>0.32</td>
<td>0.33</td>
</tr>
<tr>
<td>15</td>
<td>0.40</td>
<td>0.43</td>
<td>0.43</td>
<td>0.44</td>
<td>0.44</td>
<td>0.42</td>
<td>0.43</td>
</tr>
<tr>
<td>16</td>
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<td>0.59</td>
<td>0.59</td>
<td>0.61</td>
<td>0.61</td>
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<td>17</td>
<td>0.78</td>
<td>0.79</td>
<td>0.80</td>
<td>0.82</td>
<td>0.81</td>
<td>0.80</td>
<td>0.81</td>
</tr>
<tr>
<td>18</td>
<td>1.07</td>
<td>1.08</td>
<td>1.08</td>
<td>1.10</td>
<td>1.11</td>
<td>1.09</td>
<td>1.10</td>
</tr>
<tr>
<td>19</td>
<td>1.43</td>
<td>1.44</td>
<td>1.45</td>
<td>1.47</td>
<td>1.48</td>
<td>1.47</td>
<td>1.47</td>
</tr>
<tr>
<td>20</td>
<td>1.83</td>
<td>1.84</td>
<td>1.85</td>
<td>1.88</td>
<td>1.88</td>
<td>1.88</td>
<td>1.88</td>
</tr>
<tr>
<td>21</td>
<td>2.25</td>
<td>2.24</td>
<td>2.27</td>
<td>2.27</td>
<td>2.27</td>
<td>2.28</td>
<td>2.27</td>
</tr>
</tbody>
</table>

| Fog Density | 0.15 | 0.15 | 0.15 | 0.14 | 0.15 | 0.14 | 0.15 |

Table 4.1 Optical densities for the different sensitometer steps for a typical period of processor monitoring. The measured fog densities are also indicated.

Similar results to the ones presented in Table 4.1 were obtained throughout the course of this study.

Energy Response of the Kodak XV-2 Film.

It is well known that, because the photoelectric effect depends on the cube of the atomic number, the silver (Z = 45) in the film emulsion absorbs photons below 150 keV very strongly by the photoelectric process. Since most clinical photon beams contain a scatter component of low-energy photons, the correlation between optical density and dose becomes tenuous. For this reason, the usefulness of film for low-energy photon dosimetry is relatively limited. In the case of electron beams, however, the experimental information available on the energy response of film is sparse. Data reported by Wachsmann and Dudley indicate very small sensitivity changes for energies above 1 MeV. Markus and
Paul\textsuperscript{20} estimated that the film response varies by 5\% between 0.3 and 10 MeV. Dutreix and Dutreix\textsuperscript{14} estimated 2\% between 0.3 and 20 MeV. To obtain a clearer picture, the energy response of the film used in this study was investigated.

In general, the energy response of the film will be determined by the emulsion stopping power vs. energy relationship and the capability of the electrons to penetrate the emulsion thickness. In practical dosimetry, the film is exposed in a phantom with physical characteristics (mass stopping powers and angular scattering powers) different from the ones of the film emulsion and the cellulose or polyester base. Thus, in a first approach, a theoretical energy response can be calculated based on the stopping power ratio of the film emulsion and the dosimetry phantom used. Such a theoretical response was calculated with the mass collision stopping powers\textsuperscript{21} of AgBr and clear polystyrene (the phantom material used in this study) and the following equation:

\[
\left( \frac{OD}{D_{\text{Poly}}} \right)_{E} = \left( \frac{(S/\rho)_{\text{AgBr}}}{(S/\rho)_{\text{Poly}}} \right)_{10 \text{MeV}} \left( \frac{(S/\rho)_{\text{AgBr}}}{(S/\rho)_{\text{Poly}}} \right)_{10 \text{MeV}} E,
\]

where the subscript $E$ designates electron energy $E$ and 10 MeV has been chosen as the energy of normalization. The mass collision stopping power ratio of AgBr/polystyrene as a function of electron energy is presented in Fig. 4.3.

Once normalized, the curve in Fig. 4.3 provides a theoretical response for film exposed in polystyrene. Stopping power is not, however, the exclusive factor affecting the energy response since the capability of low-energy electrons to penetrate the emulsion and scattering effects are also important. Indeed, the change in scattering with electron energy will modify the obliquity and path length of the electrons in the emulsion. Also, the wrapping surrounding the film, when present, may influence the performance.
An experimental investigation of the energy response of the Kodak XV-2 film was performed by irradiating pieces of film with different electron energies. The basic setup used for measurements was the same as the one described for obtaining the sensitometric curve. In this case, the film pieces were placed at different depths in phantom and several beam energies were used. The mean energy of the electrons striking the film pieces was calculated using Eq. (3.4) with the known values of $E_0$ and $R_p$ for the accelerator used. Using the PDD of the different beam energies, the dose in water received by the film was calculated, and the dose in polystyrene was obtained with Eq. (4.13). The number of monitor units was set such that every piece of film received approximately 35 cGy. The film sensitivity was obtained by dividing the measured optical densities of the film pieces by the dose they received. All the results are presented in Table 4.2 and Fig. 4.4.

The theoretical film sensitivity calculated with Eq. (4.18) corresponds quite well with the experimental data presented in Fig. 4.4. This suggests that the mass collisional stopping power ratio is the main factor determining the film energy response. According to the experimental data, the film energy response is

Figure 4.3 Ratio of mass collisional stopping powers of silver bromide and polystyrene as a function of electron energy. Data from ref. 21.
### Table 4.2

Results of the film energy response investigation. The normalized film sensitivity is obtained by dividing the value of OD/Dose for a given energy by its value for 9.6 MeV.

<table>
<thead>
<tr>
<th>Mean Energy (MeV ± 0.2)</th>
<th>Dose Received (cGy)</th>
<th>Optical Density (± 0.005)</th>
<th>O.D./Dose (1/cGy)</th>
<th>Normalized Sensitivity</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2</td>
<td>35.3 ± 0.9</td>
<td>1.160</td>
<td>0.033 ± 0.001</td>
<td>0.89 ± 0.03</td>
</tr>
<tr>
<td>0.9</td>
<td>34.5 ± 0.5</td>
<td>1.130</td>
<td>0.0328 ± 0.0006</td>
<td>0.89 ± 0.02</td>
</tr>
<tr>
<td>1.1</td>
<td>35.3 ± 0.5</td>
<td>1.250</td>
<td>0.0354 ± 0.0007</td>
<td>0.96 ± 0.02</td>
</tr>
<tr>
<td>2.0</td>
<td>35.0 ± 0.5</td>
<td>1.230</td>
<td>0.0351 ± 0.0007</td>
<td>0.95 ± 0.02</td>
</tr>
<tr>
<td>3.7</td>
<td>35.4 ± 0.4</td>
<td>1.290</td>
<td>0.0364 ± 0.0005</td>
<td>0.98 ± 0.02</td>
</tr>
<tr>
<td>5.1</td>
<td>34.6 ± 0.4</td>
<td>1.260</td>
<td>0.0364 ± 0.0005</td>
<td>0.98 ± 0.02</td>
</tr>
<tr>
<td>7.7</td>
<td>35.0 ± 0.4</td>
<td>1.280</td>
<td>0.0366 ± 0.0005</td>
<td>0.99 ± 0.02</td>
</tr>
<tr>
<td>9.6</td>
<td>34.6 ± 0.4</td>
<td>1.280</td>
<td>0.0370 ± 0.0005</td>
<td>1.00 ± 0.02</td>
</tr>
<tr>
<td>13.2</td>
<td>35.4 ± 0.4</td>
<td>1.310</td>
<td>0.0371 ± 0.0005</td>
<td>1.00 ± 0.02</td>
</tr>
<tr>
<td>15.1</td>
<td>34.6 ± 0.4</td>
<td>1.290</td>
<td>0.0372 ± 0.0005</td>
<td>1.01 ± 0.02</td>
</tr>
</tbody>
</table>

### Figure 4.4

The measured sensitivity of the Kodak XV-2 film as a function of the mean energy of the electrons. The solid curve is a theoretical sensitivity based on the normalized mass collision stopping power ratio presented in Fig. 4.3.
relatively constant in the energy range from 2 to 15 MeV, but decreases for lower electron energies. However, the film sensitivity does not drop below 95% (of its value at 9.56 MeV) until the mean electron energy falls to about 1 MeV. In practical measurements, usually a limited variation of the electron energy is involved, particularly within the depth dose distribution of clinical importance, and film can be considered a reliable detector. In this thesis, the electron beam energies investigated were 9 MeV and 12 MeV. Thus, the energy response of the film is not a problem since the decrease in film sensitivity is only important for depths near the practical range. Furthermore, the results of this experiment give us the confidence that film can be irradiated with matched electron beams of different energies and provide reliable data. Also, the measured sensitometrie curves and the extent of their linear portion are generally energy independent.

Some Important Precautions.
Film can be used either perpendicularly or parallelly to the beam axis. For measurements with film parallel to the beam axis, precautions must be taken to carefully align the film edge in the phantom slabs. Indeed, Dutreix and Dutreix have demonstrated that film edge sticking out or pushed into the phantom produces significant distortions of the dose at shallow depths. Dutreix and Dutreix also demonstrated that air gaps between the film and the phantom slabs produce considerable distortions to the dose distribution. Thus, the film must be compressed well enough between the slabs to extrude all air.

4.10 Radiochromic Film Dosimeters.

Surface dose measurements were carried out with radiochromic film dosimeters. The radiochromic film is a thin (100 μm) detector whose sensitive layer (6 μm thick) changes from colorless to blue by dye polymerization upon exposure to ionizing radiation. An attractive feature of the radiochromic film is that it does not require post-irradiation processing. Its use and operational characteristics have been described by several authors.

---

* GafChromic dosimetry media, Type MD-55, distributed by Nuclear Associates.
Figure 4.5  Dose response curve of the radiochromic films used for measurements of surface doses. The dose is measured in polystyrene.

For optimal results, the radiochromic films must be analyzed with a light source emitting in the red portion of the visible spectrum. A home made densitometer was built from a diode laser ($\lambda = 670$ nm) and a photodiode detector connected to an electrometer. Extensive series of measurements demonstrated that relative dose measurements can be performed with a 3.5% precision with the dosimetry system consisting of radiochromic films and our densitometer. The dose response curve used to convert optical density to dose is presented in Fig. 4.5. This calibration curve was obtained in much the same way as sensitometric curves for Kodak XV-2 films (refer to Section 4.9).

4.11 Conclusions.

This chapter has provided an introduction to the basic concepts of absolute and relative electron dosimetry. The most important methods employed in electron dosimetry have been reviewed. Many important aspects of film dosimetry have been discussed and the reliability of this technique for the present application has been demonstrated.
Chapters III and IV have been devoted to the discussion of fundamental aspects of the physics and dosimetry of electron beams. Now that this necessary knowledge has been introduced, we turn our attention in the final chapters to our new technique for electron field matching.

4.12 References.


20B. Markus and W. Paul, Strahlentherapie 92, 612 (1953).


CHAPTER V

THE NEW PENUMBRA GENERATOR: METHODS AND MATERIALS.

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5.1 Introduction.

This chapter describes the geometric and dosimetric characteristics of the new approach proposed to modify the penumbra of an electron beam and thereby improve the dose uniformity in the overlap region when two electron fields are abutted. The experimental procedure used to determine the optimal field separation and to quantify the uniformity of the dose distribution in the junction region is also presented.

5.2 The Penumbra Generator and its Modification of the Electron Beam.

The new penumbra generator consists of a metal block made of Lipowitz metal and placed on the electron applicator insertion plate to stop part of the electron beam on the side of the field abutment (see Figures 5.1 (A) and 5.2). In this configuration the electrons of the unblocked part of the beam have a large distance to be scattered in air under the block before reaching the patient. The result is a wide and smooth penumbra, ideal for field matching (see Fig. 5.3). The penumbra broadening occurs only at the edge of abutment, while the remaining field boundaries are defined near the patient in the usual manner (see
Fig. 5.2) with the electron cone. This technique can be considered as a compromise between a regular stationary electron field collimated with an electron cone and an electron beam without the secondary collimation. Indeed, by defining the edge of abutment further away from the patient surface, the penumbra generator mimics the effect of removing the applicator (refer to Chapter II), but the applicator is kept in place to define the remainder of the field.

The penumbra generator and the setup used for measurements are depicted in Fig. 5.1. The Cerrobend block is placed on top of the insertion plate as shown in Fig. 5.1 (A). The approach taken is to completely stop the electrons in the blocked part of the 9 MeV and 12 MeV electron beams used in this study. Purdy and Choi have performed transmission measurements with Cerrobend shields and have shown that for a 12 MeV beam a Cerrobend thickness of 2 cm is needed to attenuate 97% of the open beam dose. Therefore, a 2 cm thick block was used in our experiments ensuring that only a weak bremsstrahlung contamination is transmitted through the block.

The insertion plate of the applicator and the penumbra generator are situated 44 cm above the surface of the phantom placed at nominal SSD (100 cm). The penumbra generator is therefore at 56 cm from the nominal source. The block is aligned parallel to the edge of the beam to be matched as shown in Fig. 5.2. Its relative position with respect to the center of the field is located with the help of the alignment laser (to locate the central axis) and the light localization field. We define the distance X as the distance between the central axis of the electron beam and the position of the edge of the light field under the block (Fig. 5.1 (A)). This distance is measured on top of the custom-made cutouts insert, which is 6.3 cm above the phantom surface. This measurement point was chosen for convenience, and the corresponding distance at the surface of the phantom can be found from a simple scaling factor (1.07 times the distance X). The gap between adjacent modified electron beams is measured between the light field edges under the blocks at the surface of the phantom (see Fig. 5.1 (B)).

The effect of the penumbra generator on the electron beam is illustrated by a typical example in Fig. 5.3. The modified isodose distribution was obtained with a

* Cerrobend is a trade name for Lipowitz metal: 50% bismuth, 27% lead, 13% cadmium and 10% tin.
9 MeV beam and a 15x15 cm² applicator with a distance X of 3 cm. For comparison, the dose distribution obtained with the same beam in the absence of the penumbra generator is also presented. The penumbra is substantially increased at the edge to be abutted, and the isodose curves are essentially parallel to the beam axis. These points will be further discussed in the next chapter. For the moment, it is important to note that the light field edge position (under the block) at the surface of the phantom corresponds approximately to the 50% isodose curve. Thus, the optimal field separation (as measured between the light field edges) will always be relatively small. It is clear that, by varying the distance X from zero up to a maximum value which will be discussed in the next chapter, a wide range of field sizes can be obtained with a given applicator.

Figure 5.2 illustrates the use of the penumbra generator in a clinical application with custom-made shields defining the composite field boundaries. To use the penumbra generator in conjunction with shields, the shields have to be sufficiently wide open at the edge of abutment to avoid interference with the air-scattered electrons under the block. This is depicted in the upper portion of Figures 5.2 (A) and 5.2 (B), where we see that the opening in the two complementary shields goes well beyond the junction plane indicated by the dotted lines. The fact that the penumbra generator can be used easily with custom-made shields is a clear advantage over other beam-edge modifying devices, described in Section 2.4.

For clinical application, thin aluminum plates are used to support the Cerrobend block. Four holes were drilled in the insertion plate, and the aluminum plates are screwed to the applicator. The penumbra generator is taped in its appropriate position on the aluminum plates during the first treatment fraction. For subsequent fractions, one simply has to screw back in place the whole assembly consisting of the block taped on its supporting plates.

5.3 Film Irradiation Setup.

Film dosimetry was used in this study to obtain isodose distributions and beam profiles at various depths in phantom. The main characteristics of film dosimetry which made it the method of choice in this study are its short measurement time to obtain two-dimensional distributions and the fact that film
Figure 5.1  Experimental setup showing: A) The Siemens electron applicator with the penumbra generator placed on the insertion plate. The measurement point of the distance X is indicated. SI indicates the Shield Insert where collimation is provided by insertion of custom-made cutouts which define field boundaries. B) The two matched fields with the gap measured at the phantom surface. (In both sketches, the light fields are also shown).
Figure 5.2 A)

When the Penumbra Generator is used with custom made Shields, they must be Wide Open at the Edge Under the Block to avoid Interference with the Air-Scattered Electrons.

Thin Aluminum Plates to Facilitate the Positioning of the Block.

The Siemens Electron Applicator

Custom Made Shield

The Penumbra is Enlarged through the action of the Penumbra Generator.

Electron Field with Modified Penumbra.

Figure 5.2 (A) Sketches showing how two complementary modified electron fields can be matched to obtain a composite treatment field when the penumbra generator is used with custom-made cutouts. (See text for details).
Figure 5.2 (B) Sketches showing how two complementary modified electron fields can be matched to obtain a composite treatment field when the penumbra generator is used with custom-made cutouts. (See text for details).
Figure 5.3  Illustration of the effect of the penumbra generator on the electron beam. The upper isodose distribution was obtained with a 9 MeV beam and a 15x15 cm² applicator with a distance X of 3 cm. The lower dose distribution was obtained with the same beam in the absence of the penumbra generator. (The light field is also shown).
dosimetry constitutes the only practical means of recording the composite dose distribution of two matched fields. Ready-pack Kodak XV-2 films were placed in phantom in a plane parallel to the central axis of the beam. The films were aligned in the crossplane direction at the center of the field. The crossplane direction is the one parallel to the plane in which the gantry rotates. A 30x30x30 cm³ polystyrene phantom consisting of 2.5 cm slabs stacked together was used. The setup was designed to minimize the effects of the potential film artifacts, discussed at the end of Chapter IV. The film jacket was punctured at each corner and the entire assembly, consisting of the phantom slabs and the aligned film, was tightly clamped together with a large carpenter cramp to avoid any perturbation to the dose distributions caused by air trapped in the film jacket. The films were kept in their jacket during irradiation to exclude light and Cerenkov emissions. The film edge facing the source was carefully aligned with the phantom surface. The excess wrapper extending beyond the film edge was folded over to one side and taped to the phantom. All irradiations were carried out with the phantom surface placed at the nominal SSD of 100 cm.

5.4 Materials Used in this Study.

All measurements were made using the 9 MeV and 12 MeV electron beams from a Siemens Mevatron KD-2 linear accelerator. Experiments were performed with the following clinically useful electron cones: 10x10 cm², 15x15 cm², and 20x20 cm². Film dosimetry was performed with a Wellhöfer dosimetry system (WP600). The Wellhöfer is a complete system for radiotherapy beam analysis consisting of a controller unit interfaced to a personal computer (with accompanying software), a water tank with a remotely driven arm to allow beam scanning in three dimensions with an ionization chamber, a two channel electrometer, and an automated densitometer for film scanning.

The densitometer is a fully automated device allowing two dimensional scanning of exposed films. The light source is a regulated and pulsed diode, emitting in the infra-red region (950 nm). Before entering the film, the light emitted by the transmitter is collimated to produce a light beam with a diameter of 0.8 mm. The diameter of this beam defines the spatial resolution obtained in this study.
5.5 Experimental Procedure.

Our first task was to investigate the abutment of two electron beams of the same energy on a flat surface. This involves optimizing the field separation and obtaining a quantitative indicator of the dose uniformity in the junction region. A flowchart of the experimental procedure is given on page 79. The initial data were taken by irradiating the films with a single modified beam for various distances $X$ and for all the electron cones used. Beam profiles at seven different depths were extracted from these films and saved in an ASCII format from the Wellhöfer. The depth of the saved profiles was chosen to cover uniformly the dose distribution down to the 50% isodose line, thus covering the range of depths of interest in clinical practice. The depths chosen were 0.8, 1.3, 1.8, 2.0, 2.7, 3.2 and 3.6 cm for the 9 MeV electron beams and 0.8, 1.5, 2.0, 2.5, 3.3, 3.8 and 4.5 cm for the 12 MeV beams.

These profiles were then fed into a computer program which generates a mirror image of the profiles and combines them with different gaps between the two fields. The resulting combined profiles were used to find an optimal gap for different distances $X$, for the 9 MeV and 12 MeV beams and for the three electron cones used. Furthermore, the dose distribution in the junction region as a function of the field separation was observed and quantified. The program was designed using a commercial spreadsheet and it could load seven profiles of up to 510 points each. The profiles were saved from the Wellhöfer with points at every half a millimeter so that, once loaded in the program, the matched fields could be reproduced with field separations varying by steps of 0.5 mm. On-line visualization of the composite profiles and quantification of the uniformity of the resulting dose distribution for different gaps between the fields was provided by the program.

Bagne\(^2\) has defined the Depth Dose Ratio (DDR) to describe the composite dose distribution of adjacent fields. This is defined as the ratio of the dose at a depth $z$ on the mid-separation axis to the dose at that depth on the central axis of a

\(^1\)Lotus 1-2-3 v.4.01 for Windows. Lotus Development Corporation.
FLOWCHART OF THE EXPERIMENTAL PROCEDURE.

- Film irradiations with a single modified beam for different X distances and for the 3 cones used.

- Extraction of profiles at seven different depths uniformly covering the dose distribution down to the 50% isodose.

- Computer program generating a mirror image of the profiles and combining them with different field separations.

- Verification films irradiated with two matched fields combined with the optimal gaps prescribed by the program output.

- Extraction of two profiles around Dmax.

- Reproduction of these profiles by the computer program.

- Correction on the First approximation.

- Correlation of the Flatness (AFF) definition:

  Average of the Flatness of the 5 shallowest composite profiles; where the Flatness definition used is the maximum value between:

  \[
  \frac{\text{MaximumDose} - \text{CentralDose}}{\text{CentralDose}}
  \]

  AND

  \[
  \frac{\text{CentralDose} - \text{MinimumDose}}{\text{CentralDose}}
  \]
single field:

\[
\text{DDR}(z) = \frac{PDD(z) \text{ at mid-separation axis}}{PDD(z) \text{ at central axis}}
\]

However, this interesting concept fails to account for the particular shape that the composite beam profiles have for particular distances \(X\) (this will be discussed further in the next chapter). For this reason, another indicator has been defined. In this study, the parameter chosen to describe the goodness of the composite dose distribution for a particular field separation is the *Average Field Flatness* (AFF), defined as the average of the flatness of the five shallowest composite profiles. Here, the flatness definition used for a given composite profile is the largest value between:

\[
\frac{\text{Maximum dose} - \text{Central dose}}{\text{Central dose}} \quad \text{and} \quad \frac{\text{Central dose} - \text{Minimum dose}}{\text{Central dose}}
\]

where the central dose is the dose value at the center of the composite field, and the maximum and minimum dose values are taken from within 80% of the composite field width. This AFF was found to be the best indicator of the uniformity of the dose distribution in the junction region. The optimal gap for a given set of parameters (beam energy, distance \(X\), and electron cone size) is the one which minimizes the AFF.

The precision on the optimal gaps was limited by the positioning of the films with respect to the beam central axis. Any error made in locating the field center on the films is reproduced in the optimal gaps obtained. Verifications were performed by direct irradiations of films with two modified beams matched with field separations prescribed by the output of the program. Two profiles around \(d_{\text{max}}\) were extracted from these films and reproduced by the computer program, giving the correction needed on the previously obtained optimal gaps. The overall procedure gave a \(\pm 0.5\) mm precision on the optimal gaps to be used with a given set of conditions, namely: beam energy, distance \(X\), and electron cone size.
Finally, isodose measurements were made by irradiating films with two modified beams matched with the optimal gaps. Tests were also performed to see how the technique could be used to match beams of both energies. In Chapter VI, results are presented for 9 MeV and 12 MeV beams and for matched beams of both energies. The application of the technique to irradiation of curved surfaces is discussed in Chapter VII.

5.6 Conclusions.

This chapter has described the new device proposed to modify the penumbra of an electron beam. The experimental procedure used to determine the optimal field separation and to quantify the uniformity of the dose distribution in the junction region was also presented. The results of a systematic study of electron fields matching with the penumbra generator are summarized in the next chapter.

5.7 References.


6.1 Introduction.

This chapter summarizes the results of a systematic study of electron field matching with the penumbra generator described in Chapter V. The chapter deals with field matching on a flat surface. Application of the technique to curved surfaces is discussed in the next chapter. Profiles and isodose distributions of modified beams are presented. The variation of the AFF as a function of the field separation when using the penumbra generator is compared to matching of unmodified beams. Results are presented for adjacent fields of 9 MeV and 12 MeV electrons and for matched beams of both energies.

6.2 Modified Beam Profiles and Penumbra Widths.

Figure 6.1 shows two typical sets of beam profiles with the penumbra modified by the penumbra generator (left penumbra). Profiles are shown for the 9 MeV (A) and 12 MeV (B) electron beams with the 15x15 cm² applicator and a distance X of 3 cm. The profiles show a penumbra which is smooth and linear over a broad width at all depths. Such a linear dose gradient around the 50% intensity point is ideal for obtaining a perfect match. Indeed, the linear dose gradient ensures that the penumbra of both fields will sum in a complementary...
fashion at all depths (refer to Section 2.3). Furthermore, the decreased dose gradient makes the overlap region between adjacent fields less sensitive to the separation of lateral fields.

Figure 6.1 Electron beam profiles with the penumbra modified by the penumbra generator. Data were obtained with the 15x15 cm² applicator and a distance X of 3 cm: A) 9 MeV and B) 12 MeV. (Right: unmodified penumbra; Left: penumbra modified by the penumbra generator).
Beam profiles measured at $d_{\text{max}}$ showed that the penumbra width (measured between the 80% and 20% dose values) was increased by the penumbra generator from about 1.4 cm to 3.3-3.4 cm (depending on the distance $X$ and the electron cone size) for 9 MeV electrons. The 12 MeV penumbra was increased from about 1.3 cm to 2.7-2.9 cm under the same conditions. The measured penumbra widths for various values of $X$ and three electron cones are given in Table 6.1. These penumbra widths are comparable to those obtained by Kurup et al. (see Chapter II). It was found that the penumbra width eventually decreases past a certain distance $X$ (see Table 6.1). This corresponds to the point where the edge of the electron applicator's lower part (see Fig. 5.1 (A)) begins to interfere with the air-scattered electrons. For both energies, it becomes clearly observable at about $X = 4, 6$ and 8 cm for the 10x10 cm$^2$, 15x15 cm$^2$, and 20x20 cm$^2$ applicators.

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Table 6.1 Penumbra widths measured at $d_{\text{max}}$ for the 9 MeV and 12 MeV beams. The widths are measured between the 80% and 20% dose values. Left table: modified penumbra widths as a function of the distance $X$; Right table: penumbra widths of unmodified beams.
15x15 cm\(^2\) and 20x20 cm\(^2\) applicators, respectively. The resulting constriction of the isodose curves near the surface results in a degraded uniformity of the dose distribution in the junction region, which is reflected as an increased AFF for large X distances (see Table 6.2). This effect will be further discussed and illustrated in Section 6.4.

6.3 Modified Isodose Distributions.

![Figure 6.2](image)

*Figure 6.2*  Electron isodose distributions in polystyrene for beams with penumbra modified by the penumbra generator: A) 9 MeV, 15x15 cm\(^2\) applicator and distance X = 3 cm; B) 12 MeV, 15x15 cm\(^2\) applicator and distance X = 3 cm. The arrows indicate the position of the light field edge at the phantom surface.
Figure 6.2 (previous page) shows isodose distributions for modified beams at (A) 9 MeV and (B) 12 MeV. Both were obtained with the 15x15 cm² applicator and a distance X of 3 cm. The arrows indicate the position of the light field edge (under the block) at the phantom surface (refer to Fig. 5.3). In Section 2.3 we described what would be an electron penumbra ideal for abutment. Looking at Fig. 6.2, we see that the modified edges correspond in all essential respects to the ideal penumbra described in Chapter II (refer to Fig. 2.1 (B)). Namely, the isodoses are widely and evenly spaced, and they are quite parallel to the beam axis. As discussed in Chapter II, such parallel isodose lines produce a beam edge whose shape is independent of depth, thus providing a template to obtain planes of matched dose throughout the volume of the junction region.

For a given energy, the shape of the isodose distribution in the penumbra region was found to be slightly influenced by the distance X. All block positions (distances X) generate modified edges with the same general shape as the ones presented in Fig. 6.2, except for a small variation in penumbra widths (Table 6.1). This is true up to the point where the effect of the electron cone wall becomes important, as discussed in Section 6.2. Therefore, the effect of displacing the penumbra generator can be simply pictured as stretching (or compressing) the plateau region in the isodose distributions presented in Fig. 6.2. This simple and predictable behavior makes the technique easy to use in clinical practice.


Examples of isodose distributions for abutted modified electron fields are shown in Fig. 6.3 for (A) 9 MeV and (B) 12 MeV electron fields. The isodose curves indicate small (less than 2%) variations in the match region for both energies. The examples presented are more of academic than clinical interest, since these field sizes could be obtained simply (on a flat surface) by using a larger electron applicator with secondary shields defining the field boundaries near the patient's skin. The examples of Figure 6.3 were chosen for reasons of convenience (since it was impossible to record a very large field size entirely on a single film), and they illustrate the dose uniformity achievable with our technique.
Figure 6.3  Electron beam isodose distributions in polystyrene for adjacent matched fields using the penumbra generator: A) both fields are 9 MeV beams with the 10x10 cm$^2$ applicator and distance X of 2 cm; B) both fields are 12 MeV beams with the 15x15 cm$^2$ applicator and distance X of 4 cm. The large arrows indicate the position of the beam central axes while the small arrows indicate the light field edges.

In practice, field sizes of up to 36x20 cm$^2$ (with the 20x20 cm$^2$ applicator) can be obtained using this field matching technique. Such large and uniform fields can
be useful in many clinical applications. An example of such an application is the
treatment of medulloblastoma, where long and narrow electron fields are often
used to irradiate the spinal cord. However, the real potential of this technique is
realized in situations involving curved or irregular surfaces which cannot be
irradiated easily with a single field, or in situations requiring different electron
energies. These points will be further discussed in the next Chapter. In this
chapter we present a systematic study of electron field matching on a flat
surface to understand better the behavior and physical characteristics of the
penumbra generator.

<table>
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<th>AFF (%)</th>
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Table 6.2 Measured optimal gaps and the corresponding calculated AFF values for
different distances X.

AFF values ranging from 1.71% to 5.59% were obtained for distances X where
the effect of the electron cone wall was not a problem. The optimal gaps
measured (see Fig. 6.4) and the corresponding calculated AFF values for
different distances X are given in Table 6.2. Figure 6.4 reveals that the optimal
gaps (mm) to be used as a function of the distance X can be well approximated
by linear functions. Only the slopes of the linear fits are needed to obtain the
gap to be used for a given distance X. Slopes close to 1 were obtained for the
15x15 cm² and 20x20 cm² cones, while for the 10x10 cm² cone the slopes were close to 1.5. This dependence of the optimal gap on the distance X is the result of an increased beam divergence under the penumbra generator for larger values of X. Because of this increased divergence, the penumbra width slightly increases and the 50% isodose curve is displaced away from the light field edge as the distance X is made larger. Therefore, it is necessary to increase the field separation in order to match the 50% isodoses of both fields as the distance X is made larger.

An example of the variation in the AFF as a function of the field separation for different distances X is given in Fig. 6.5. The example is for the 9 MeV beam and the 20x20 cm² electron cone. For a given distance X the AFF is similar to a
parabolic function with the lower part of the curve corresponding to the optimal field separation. We will see in the next section that the AFF of matched modified beams is a slowly varying function compared to the behavior observed when matching regular beams.

The sudden increase in AFF for \( X = 8 \) cm (see Fig. 6.5) is caused by the interference of the lower part of the applicator with the air-scattered electrons. This effect was partly discussed in Section 6.2. For large distances \( X \), there is a slight constriction of the isodose curves near the surface. However, only the low dose portion (0-20% relative dose) of the modified penumbra is affected. Figure 6.6 provides an example of composite profiles for modified beams matched with a large distance \( X \). The peculiar shape of the composite profiles at shallow depths is the result of the constriction of isodoses in the modified penumbra, as discussed above. Even though the AFF is significantly higher for large distances \( X \) (see Table 6.2 and Fig. 6.5), Fig. 6.6 shows that the dose variations are important solely near the surface and only a small volume is affected. Thus, the composite dose distributions obtained are interesting for clinical applications even for large distances \( X \).

\[ \text{Figure 6.5} \quad \text{Variation of the AFF as a function of the field separation for various distances } \ X. \text{ Data obtained with the 9 MeV beam and the } 20\times20 \text{ cm}^2 \text{ applicator.} \]
Figure 6.6 Composite profiles for matched modified beams. Data obtained with the 12 MeV beam and the 15x15 cm$^2$ applicator with a distance X of 6 cm.

6.5 Comparison with Adjacent Unmodified Beams.

Experiments were performed with standard (unmodified) electron beams in order to quantify clearly the gain in dose uniformity provided by the use of the penumbra generator to match adjacent fields. The dosimetry procedure described in Section 5.5 was applied to unmodified electron beams to see how their AFFs would compare to the ones obtained when using the penumbra generator. We found that for optimal field separation the AFFs of matched regular 9 MeV beams are 8.6 %, 9.6 % and 12.9% for the 10x10 cm$^2$, 15x15 cm$^2$ and 20x20 cm$^2$ electron cones, respectively. For the 12 MeV beam, these values are 9.2%, 12.3% and 16.2%. When comparing these values with the ones given in Table 6.2 for the fields matched with the penumbra generator, we see that the dose uniformity in the junction region is significantly improved when using modified beams.

In addition to providing a better dose uniformity in the junction region, the penumbra generator makes the dose uniformity less sensitive to positioning
errors. This is shown for 9 MeV electron beams in Fig. 6.7, where AFF is plotted as a function of field separation for a typical modified beam (square symbols) and for a regular 15x15 cm$^2$ field (diamond symbols). Lateral setup errors at the field junction of the order of 2 mm might cause only about 3.5% variation in the AFF, compared to around 12% in the absence of the penumbra generator. Even when the modified beams are matched with a field separation of 4 mm less than the optimal gap of 5 mm, the dose uniformity is still better than that obtained with regular fields matched at optimal field separations. Results for other distances $X$ at 9 MeV and 12 MeV are similar. This is one of the most important features provided by the penumbra generator since, as discussed in Section 2.3, positioning uncertainties are an inherent part of the radiotherapy process.

![Figure 6.7 Average Field Flatness (AFF) as a function of field separation for adjacent electron beams. Square symbols are for 9 MeV electron beams modified by the penumbra generator, with the 15x15 cm$^2$ applicator and a distance $X$ of 4 cm; diamond symbols are for standard unmodified 9 MeV electron beams with the 15x15 cm$^2$ applicator.](image)
Figure 6.8 A) (Modified 9 MeV Beams)

- Depth (cm)
- Crossplane (cm)
- Dose (%)
Figure 6.8 B) (Unmodified 9 MeV Beams)
Figure 6.8 C) (Unmodified 12 MeV Beams)
Figure 6.8 (Three previous pages) Composite beam profiles for fields matched with various gaps. In each case, the central display is for fields matched with an optimal gap, while the upper and lower displays are for fields matched at optimal gap - 2 mm and optimal gap + 2 mm, respectively. Part (A) is for 9 MeV beams modified by the penumbra generator with the 15x15 cm$^2$ applicator and a distance $X$ of 3 cm (optimal gap = 2 mm); part (B) is for unmodified 9 MeV beams with the 15x15 cm$^2$ applicator (optimal gap = 3 mm). Part (C) is for unmodified 12 MeV beams with the 15x15 cm$^2$ applicator (optimal gap = 4 mm).

Figure 6.8 further illustrates the relative insensitivity of the new field matching technique to lateral positioning uncertainties. Comparison is made between adjacent modified and regular beams with different gaps between the fields (refer to the figure caption above for more details). We can see that a ±2 mm shift with respect to the optimal field separation introduces dose non-uniformities of ±20% when matching regular beams (Fig. 6.8 (B) and (C)). On the other hand, similar lateral shifts leave the dose distribution almost unchanged when using the penumbra generator (Fig. 6.8 (A)). The ±2 mm shift used in these examples is a somewhat conservative figure. Indeed, in daily clinical practice, higher lateral shifts (or positioning uncertainties) may be encountered. As illustrated in Fig. 6.7, deviations from the optimal gap rapidly lead to severe dose non-uniformities when matching regular fields. Thus, it is difficult to know accurately what is the dose distribution from one fraction to another when matching regular beams. On the other hand, the penumbra generator provides the needed insensitivity to lateral positioning uncertainties.

6.6 Output Factors for Modified Electron Beams.

For clinical applications, it is essential to know the effect of the penumbra generator on the output of a given electron beam. For example, the output factor of the modified electron beams must be considered when adjoining fields of different energies. For needs of clarity, we will distinguish between a "primary output factor" and a "composite output factor", where the primary output factor refers to the output for a single modified beam, while the composite output factor refers to the output for the resulting composite field (Fig. 6.9).
Determination of the **primary output factors** was achieved by comparison of ionization measurements performed in water with a Farmer chamber (Nuclear Enterprises, NE 2531). Readings were taken at a depth of $d_{\text{max}}$ at the center of both the modified electron beams and the same energy beam with the unmodified electron applicators. In the case of the modified beams, the field center was defined as the **center of the light field at the phantom surface** for an SSD of 100 cm. This field center does not correspond to the usual central axis of the open field. The field centers of both modified beams are indicated by the arrows on Fig. 6.9. In the example presented the **usual beam central axes** are situated at $1.4 \text{ cm}$ inside the positions marked by the arrows.

![Figure 6.9 Typical example of matched profiles illustrating the measurement of output factors for modified and composite electron beams. The example provided is for 9 MeV beams with the 10x10 cm$^2$ applicator and a distance X of 2 cm. The arrows indicate the centers of the modified beams. Note how the tails of the modified edges contribute to the total dose at the centers of modified fields (arrows).](image)

From Fig. 6.9 we can estimate that, for fields matched with a small distance $X$, there will be an important contribution to the dose at the center of one field.
(arrows) due to the tail of the modified penumbra from the second field. When adding this contribution from the second field, we obtain the composite output factor. These contributions were measured from modified beam profiles scanned at the depth of maximum dose in the water tank of the Wellhöfer 3D plotter.

Thus, in general, the output in cGy/MU of a given composite field will be given by:

$$D_{CF} = D_{o} \times OF_{P}(X,E,F) \times [1 + OAR(X,Gap,E,F)] \quad (6.1)$$

where $D_{CF}$ is the output of the composite field, $D_{o}$ is the output of the related unmodified beam, $OF_{P}(X,E,F)$ is the primary output factor (which depends on the distance $X$, the beam energy $E$, and the applicator size $F$), and $OAR(X,Gap,E,F)$ is the contribution of the tail as depicted on Fig. 6.9. The $OAR(X,Gap,E,F)$ is the Off-Axis Ratio measured on a modified beam profile normalized to 100% at its center (position of an arrow in Fig. 6.9). In Eq. (6.1), the composite output factor is given by $D_{CF}/D_{o}$. The measured output factors are given in the next page (Fig. 6.10), and the corresponding figure caption can be found below.

Figure 6.10  (Next page) Measured output factors for modified and composite beams, for the three electron applicators studied. Solid symbols are for 9 MeV electron beams and open symbols are for 12 MeV electron beams. The primary output factors are given by circle symbols, while the composite output factors are given by square symbols.
Figure 6.10

10x10 cm²

Output Factor

Distance X (cm)

15x15 cm²

Output Factor

Distance X (cm)

20x20 cm²

Output Factor

Distance X (cm)
6.7 Adjoining Fields of Different Electron Beam Energies.

Experiments were performed to see if the penumbra generator is useful for matching electron beams of different energies with good dose uniformity. Direct irradiations of films with matched beams of 9 MeV and 12 MeV were performed. In the present case, the number of monitor units used for each field was adjusted to account for the different output factors presented in the last section. Abutted fields of different energies are shown in Fig. 6.11. They are for (A) 10x10 cm$^2$ electron cone with a distance $X$ of 2 cm, and (B) the 15x15 cm$^2$ cone with a distance $X$ of 3 cm. The number of monitor units used for obtaining the isodose distributions in Fig. 6.11 is given in the figure caption.

The composite dose distributions present small variations ($\pm 3\%$) caused by the unequal penumbra width of the 9 MeV and 12 MeV beams. Being slightly wider (see Table 6.1), the 9 MeV penumbra contributes more on the 12 MeV side than the 12 MeV penumbra does on the 9 MeV side. As a result, beam profiles at a depth of about 2 cm exhibit a hump on the 12 MeV side and a small hollow on the 9 MeV side. However, these variations are acceptably small and the 90% and 80% isodose surfaces are found to be very smooth. Therefore beams of different energies can be matched with good dose uniformity by using the same optimal gaps presented in Table 6.2. The optimal gaps found for 9 MeV and 12 MeV beams are equal (within the experimental uncertainty) for a given distance $X$. Thus, taking the optimal gap value for 9 MeV or 12 MeV beams will ensure good dose uniformity when matching beams of either energy.

Figure 6.11  (Next page) Distributions shown in (A) and (B) are for matched fields of 9 MeV and 12 MeV electron beams with: (A) 10x10 cm$^2$ electron applicator and a distance $X$ of 2 cm, 40 MU for the 9 MeV beam, and 38 MU for the 12 MeV beam; (B) 15x15 cm$^2$ applicator and distance $X$ of 3 cm, 40 MU for the 9 MeV beam and 39 MU for the 12 MeV beam. Distribution shown in (C) was obtained with 12 MeV electron beams and the 20x20 cm$^2$ applicator. A 9 mm thick bolus was used to cover half of the field.
Figure 6.11

A) Distance from mid-separation axis (cm)

B) Distance from mid-separation axis (cm)

C) Crossplane (cm)
Several tests were performed with a single unmodified 12 MeV beam and bolus material covering half of the field. These experiments were done in order to compare their results with isodose distributions obtained with matched modified beams of different energies. The thickness of bolus material used (9 mm of polystyrene) was chosen to bring the 90% isodose for the 12 MeV beam to the same depth as that for a 9 MeV beam. The irradiations were made with the phantom surface positioned at the nominal SSD.

An isodose distribution obtained with bolus is presented in Fig. 6.11 (C). The hot and cold spots (near the central axis) are clearly visible and are characteristic of the dose distribution behind the edge of a small inhomogeneity\(^2\). However, these hot and cold spots are not a major feature of this dose distribution, since they could be decreased by smoothing the edge of the bolus material. On the other hand, the use of bolus has the effect of reducing the dose build-up region (see the right portion of Fig. 6.11 (C)). This attenuation of the "skin-sparing effect" is not always desirable. Otherwise, the distribution of Fig. 6.11 (C) is comparable to the distributions presented in Figures 6.11 (A) and 6.11 (B). However, the use of a single field and bolus material to obtain a smooth distribution like the one of Fig. 6.11 (C) is limited to flat surfaces. When faced with a curved surface, obliquity effects at the edges of the field preclude the use of a single field. In such cases, matched fields of different energies could prove to be a valuable tool to irradiate target volumes of a complicated shape.

6.8 Surface Dose.

The surface dose is an important consideration in electron beam therapy since the maximum dose which can be imparted to the target volume is sometimes limited by the tolerance of the patient's skin. Usually, surface dose is measured with a thin-window plane-parallel ionization chamber or with thin TLD chips. Since none of these techniques were available at our institution, surface dose was investigated with radiochromic film dosimeters whose physical and dosimetric properties have been discussed in Section 4.10.

Surface doses were measured by placing strips of radiochromic film at \(d_{\text{max}}\) and at the surface of a polystyrene phantom. The strips were aligned in the crossplane direction at the center of the electron fields. The surface dose was
evaluated at various off-axis positions by comparing the measured dose at the surface to the corresponding dose at \( d_{\text{max}} \). Irradiations with 9 MeV and 12 MeV electrons were performed with regular beams and matched fields.

Typical results are presented in Fig. 6.12, where comparison is made between a regular 15x15 cm\(^2\) beam and two matched fields (distance \( X = 2 \) cm, field separation = 2 mm) at 12 MeV. In the case of the adjacent fields the vertical axis (\( x = 0 \) cm) corresponds to the mid-separation axis. No error bars are displayed on Fig. 6.12 for reasons of legibility. Results with other geometries and the beam energy of 9 MeV are similar. Thus, within the experimental precision, the surface dose is not affected by the presence of the penumbra generator.

6.9 Conclusions.

This chapter has summarized the results of a systematic study of electron field matching with the penumbra generator. The penumbra generator increases the penumbra width from about 1.4 cm to 3-3.4 cm for 9 MeV electrons and from
about 1.3 cm to 2.7-2.9 cm for 12 MeV electrons. The modified edges correspond in all essential respects to an electron penumbra ideal for abutment. We demonstrated that the penumbra generator provides a better dose uniformity in the junction region between adjacent electron fields and makes the dose uniformity less sensitive to positioning uncertainties.

6.10 References.


CHAPTER VII

IRRADIATION OF CURVED SURFACES.

7.1 Introduction.

This chapter presents the potential of the penumbra generator technique for the irradiation of curved surfaces. A clinical application of the penumbra generator is presented. A brief discussion of the future investigations necessary to make this approach systematic is included in Chapter VIII.

7.2 Application to Irradiation of Curved Surfaces.

In order to try the penumbra generator technique on curved surfaces, a few tests were performed with an anthropomorphic phantom (Rando phantom). Rando phantoms are patient analogs molded about natural male or female human skeletons with plastic materials which are radioequivalent to soft tissues. The phantoms are transected at 2.5 cm intervals for insertion of TLD dosimeters or film.

A piece of film was cut in the dark room to conform to the shape of a Rando slice and sandwiched, light proof, between two slices. A few slices were then added on both sides of the "sandwich" and the whole assembly was tightly clamped together. A given arc along the film edge was chosen as a target area, and this was irradiated with three 9 MeV modified beams. The tests were performed with the 15x15 cm² electron applicator. The technique is similar to the three-field wraparound technique described by Norris¹, except that the dosimetry of the field junctions is improved by the penumbra generators. Here, the central field had to
be modified at both edges to ensure good matching with the two lateral fields (see Fig. 7.1).

An example of achievable dose distributions is presented in Fig. 7.2. This dose distribution was obtained with the following field parameters: Field A (refer to Fig. 7.2) is an AP field (0° gantry angle) with a distance X of 2.8 cm, Field B has a gantry angle of 39° and distances X of 2.35 cm on Field A and Field C sides, Field C has a gantry angle of 78° and a distance X of 0.65 cm. All fields were positioned at nominal SSD and the monitor units used were: 40, 32 and 40 for Fields A, B and C, respectively. The skin gaps used are simple averages of the optimal gaps (Table 6.2) obtained previously for the different distances X to be matched. The curves indicate no important variations near the field junctions. The dose in the central field region is slightly higher (about 2%), but the uniformity of the dose distribution in the whole target volume is excellent. The
isodoses smoothly follow the phantom surface and the 90% isodose curve uniformly covers the selected target area. Such uniform distributions should be obtainable for a wide range of curvatures and target volumes with appropriate positioning of the fields and an adequate selection of distances $X$ and monitor units per beam.

![Diagram of Phantom Contour and Fields A, B, C with isodoses and junctions](image)

*Figure 7.2* An example of isodose distribution obtained in the Rando phantom. The three fields used are indicated and the arrows indicate the field junctions.

The technique described above has the advantage that the field boundaries can be defined in the simple usual way with secondary shields (custom-made cutouts) inserted into the standard electron cone near the patient's skin (at $S_1$ on Figure 5.1 (A)), while the field matching zone is defined by the penumbra generator. No lead shielding on the patient is necessary. However, the cutouts must be wide open at the field junctions to avoid interference with the air-scattered electrons (refer to Chapter V, Fig. 5.2).
7.3 A Clinical Example.

The tests described in Section 7.2 (and the example presented in Fig. 7.2) were designed to tackle the somewhat complicated clinical situations involving large arc angles and necessitating the use of multiple fields. However, even with a two field technique, the penumbra generator can find interesting applications when the curvature of the region to be treated prohibits the use of a single field. This is illustrated by the following clinical example.

A 60 year old woman was previously treated for a lesion of the left breast in 1993. She has been admitted in 1995 to the radiation oncology department for a recurrence on the right anterior chest wall. The region to be treated was impossible to irradiate with a single field because of the grazing incidence at the edges of the field. Thus, we decided to implement the penumbra generator technique to irradiate the target volume with two fields. This case presented some difficulties since the physician wanted 12 MeV and 9 MeV beams for the anterior and lateral fields, respectively. Furthermore, the use of a 5 mm bolus was prescribed to enhance the skin dose.

We used the 20x20 cm² applicator with secondary shields for both fields. In order to optimize the dose distribution, we reproduced the patient contour using Rando slices with the help of bolus material (Superlab). Different tests were performed with films between the Rando slices, until we obtained the dose distribution presented in Fig. 7.3. The 12 MeV anterior beam had a gantry angle of 325°, while the 9 MeV lateral beam was at a 280° gantry angle. Both fields were positioned at an SSD of 100 cm (as measured on the bolus) and an optimal field separation of 4 mm (measured on the patient skin) was used. Both fields had small distances X (1.7 and 1 cm).

The patient contour and the isodose distribution measured in phantom are presented in Fig. 7.3 (the 5 mm bolus is not shown on the figure). The missing part of the distribution corresponds to the region where bolus material was used to reproduce the patient contour (film could only record the dose distribution within the Rando slices).
The target volume is well covered by the 90% isodose and clinically acceptable dose variations (10%) near the field junction are present, as illustrated in Fig. 7.3. This clinical example illustrates that the penumbra generator can be used to irradiate curved surfaces with good dose uniformity. No particular problems were encountered in the actual delivery of the treatment and this special case was well integrated into the daily routine of the department.

7.4 References.

8.1 Future Developments.

Systematization of the Technique presented in Section 7.2.

In the example presented in Fig. 7.2 a relatively small central field was used. In this case, an important part of the dose imparted in the central field region is contributed by the modified field edges of the two lateral fields. This contribution from the lateral fields has to be quantified clearly in order to correctly choose the number of monitor units to be used for the central field. Excellent dose distributions can be obtained by trial and error, but the problem of the central field weight is still left standing.

The rationale for using a small central field was to minimize the angle of incidence at the field edges and to prevent large additional air gaps produced by the extended SSD. When the radius of curvature is not too small (in the central field region), another approach would be to use a larger central field, thus minimizing the problem of the contribution of the lateral fields. This will have to be further investigated to come up with a systematic approach for curved surfaces irradiation. A detailed study of the changes in the modified penumbra under oblique incidence and extended SSD will be required. Also, the way that this influences the optimal skin gaps to be used will have to be investigated. This additional knowledge should allow us to solve the problems related to the use of this technique.
Application of the Penumbra Generator Concept for Matching Electron and Photon Fields.

In this study, the penumbra generator was placed on top of the applicator upper trimmer. Although the modified field edges are ideal for matching electron fields, the penumbra widths are too large to be useful for abutment with a photon field. However, it is clear that a range of penumbra widths could be obtained with the penumbra generator placed at different locations. A closer location of the block relative to the patient skin would result in a narrower penumbra that may facilitate abutment with a linac photon field. For instance, with the Siemens electron applicator, a middle trimmer available for that purpose is located at about 18.5 cm from the patient skin. An example of an isodose distribution for a 9 MeV beam with a penumbra modified by a block placed on this middle trimmer is presented in Fig. 8.1.

Figure 8.1  Electron isodose distribution for a 9 MeV beam, showing a modified penumbra (Left penumbra) which could be useful for abutment with a photon field. See text for details.
Radiotherapy of head and neck tumors frequently involves joining photon and electron fields. In those cases, problems of fields abutment ("hot" and "cold" spots) are frequently encountered. Papiez and Dunscombe\(^1\) have devised penumbra spreading techniques which lead to a relatively uniform dose distribution in the join-up region and reduce the effect of positioning errors on dose uniformity. A stepped wedge attenuator was used to obtain a wider penumbra for the x-ray beam and a Lucite scatterer was used for the electron beam. The scatterer was a 1.25 cm thick Lucite plate, positioned at a distance of about 3 cm from the phantom surface. This approach suffers from the same problems encountered when using plastic wedges\(^2\) (refer to Chapter II), namely, the beam energy is altered and the spreading is not limited to the edge of abutment.

A penumbra generator on the lower trimmer could advantageously replace the Lucite scatterer in the technique devised by Papiez and Dunscombe\(^1\). The modified penumbra presented in Fig. 8.1 should be useful for such an application. Investigations in this direction will be conducted in the future.

8.2 Thesis Summary.

This thesis has described the problems related to abutment of electron fields. A brief review of the different approaches used to solve these problems was provided. Chapters III and IV have introduced fundamental aspects of the physics of electron beams and electron dosimetry. Film dosimetry was thoroughly discussed by either theoretical considerations or through experiments performed by the author.

A new approach for the modification of an electron beam penumbra to improve field abutment has been presented. Beam-edge modification by means of a low melting point alloy block placed on the electron applicator's insertion plate was found to yield a beam penumbra sufficiently wide and linear to ensure good dose uniformity at the field junction. The dose distribution in the junction region was shown to be much less sensitive to positioning errors than when regular fields are matched. Since the device acts mainly by complete absorption of part of the electron beam, it modifies the beam without degrading its energy. Clinical application demonstrated that the technique is easy to implement and can be
introduced into the daily routine of a radiotherapy department. Furthermore, the
device is simple and inexpensive, and no special measurement is needed for its
design. The block has simply to be of a thickness larger than the maximum
electron range in the material, and its length and width have to be correctly
chosen to cover the part of the beam to be blocked.

A wide range of field sizes can be obtained with the existing square electron
applicators by varying the distances X used. This provides the needed flexibility
to irradiate different target areas. Furthermore, the technique was shown to be
useful for matching electron beams of different energy. These two characteristics
can prove to be useful when faced with a target volume of a complicated shape.

The potential of the technique for irradiation of curved surfaces was presented.
More work will be needed to develop a systematic approach, but the technique
could prove to be a simple and acceptable alternative to electron arcs for chest­
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8.3 References.

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