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An Investigation of Radiotherapy Electron Beams Using Monte Carlo Techniques

by

George X. Ding

A dissertation submitted in partial fulfillment of
the requirements for the degree of
Doctor of Philosophy

Department of Physics
Ottawa-Carleton Institute for Physics
Ottawa, Ontario
May, 1995

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"An Investigation of Radiotherapy Electron Beams

Using Monte Carlo Techniques"

submitted by

George Ding, B.Sc., M.Sc.

in partial fulfillment of the requirements

for the degree of Doctor of Philosophy

Chair, Department of Physics

Thesis Supervisor

External Examiner

Carleton University

May 24, 1995

Preface

In the summer of 1991 I came to Ottawa and started work on the OMEGA (Ottawa-Madison Electron Gamma Algorithm) project under the supervision of Dr. Dave Rogers. The OMEGA project is a collaborative project to develop a 3-D electron beam treatment planning system based on Monte Carlo simulation techniques. A major part of the OMEGA project is to develop a general-purpose Monte Carlo user code to simulate realistic clinical electron beams from medical electron accelerators. The detailed knowledge of clinical electron beams is essential to dose calculations for treatment planning.

During the course of the work, abstracts were presented at several meetings. The main results have been summarized in a set of papers. The articles have been incorporated directly into the thesis as following:

- Thesis **chapter 2**, *BEAM: A Monte Carlo code to simulate radiotherapy treatment units*, D.W.O. Rogers, B.A. Faddegon, G.X. Ding, C.-M. Ma, J. Wei, and T.R. Mackie, Medical Physics 22 (5) (1995). My contributions:

I wrote code for most component modules and contributed significantly to the development of the EGS4 user-code BEAM; converted machine specifications to simulation input files for all simulated accelerators; validated BEAM by comparing calculated to measured dose distributions for all simulated beams in water phantoms; measured the geometry of the beam-defining system of an accelerator installed at Ottawa Regional Cancer Center; took part in the experiment on the NRC accelerator for validating the code BEAM; modified the EGS4 user code DOSXYZ and implemented various new features in DOSXYZ: reading a simulated beam phase-space file as an input, point source with a given energy spectrum, point source with its own random-number generator for correlated sampling, repeated use of a

phase-space file with options of redistributing the incident particles for maximizing the use of the phase-space file; selecting a particle according to its history by using a variable LATCH in reading a phase-space file as an incident beam; analyzed data and produced plots.

- Thesis **chapter 3**, *Energy spectra, angular spread and dose distributions of electron beams from various medical accelerators*, G.X. Ding and D.W.O. Rogers, National Research Council of Canada Report PIRS-0439 (1994) which is available on the word-wide-web (www) at <http://www.irs.inms.nrc.ca/inms/irs/papers/PIRS0439/pirs439.html>.

My contributions:

Wrote computer codes, called PHSP_ENXY, PHSP_SPECXY and PHSP_ANG, to analyze the simulated beams (phase-space data files) in order to obtain the energy spectra, angular distributions and other characteristics of simulated beams, modified code DOSXYZ to calculate dose contributions from various components of beam-defining system of a accelerator; analyzed data and produced all figures and tables; drafted the manuscript.

- Thesis **chapter 4**, *Mean energy, energy-range relationships and depth-scaling factors of clinical electron beams*, G.X. Ding, D.W.O. Rogers and T.R. Mackie, submitted to Med. Phys. May, 1995. My contributions:

I wrote an EGS4 user code, ENXYZ, to calculate mean energy as a function of position of electrons in phantoms; analyzed data and produced all figures and tables; drafted the manuscript.

- Thesis **chapter 5**, *Calculation of stopping-power ratios using realistic clinical electron beams*, G.X. Ding, D.W.O. Rogers and T.R. Mackie, Medical Physics 22 (5) (1995). My contributions:

I modified the EGS4 user code, SPRRZ, to read a phase-space data file as an incident realistic beam to calculate stopping-power

ratios and added options to SPRRZ in order to select particles (all electrons, direct electrons, contaminant photons, etc) in the beam at user's choice for detailed study of stopping-power ratios; did all calculations; analyzed data and produced figures; implemented all calculated water/air stopping-power ratios of clinical beams in a code, called SPR; drafted the manuscript.

- Thesis **chapter 6**, *Fluence correction factors used in conversion of dose measured in plastic to dose in water*, G.X. Ding, D.W.O. Rogers, J. Cygler and T.R. Mackie, submitted to Med. Phys. May, 1995 My contributions:

I simulated all beams and did all calculations; wrote a computer code, called DOSFLU, to analyze data; produced all figures and tables; drafted the manuscript.

- Thesis **chapter 7**, *Calculation of backscatter into beam monitor chamber for electron beams from various accelerators*, G.X. Ding, D.W.O. Rogers and T.R. Mackie, to be submitted to Med. Phys. (1995) My contributions:

I simulated all accelerators and did all calculations; analyzed data and produced all figures and tables; drafted the manuscript.

An Investigation of Radiotherapy Electron Beams Using Monte Carlo Techniques

**A dissertation submitted in partial fulfilment of
the requirements for the degree of
Doctor of Philosophy
at Carleton University**

by

George Ding

1995

ABSTRACT

Radiotherapy electron beams are more complicated than photon beams due to variations in the beam production, the scattering of low-energy electrons, and the presence of contaminant photons. The detailed knowledge of a radiotherapy beam is essential to an accurate calculation of dose distribution for a treatment planning system. This investigation aims to enhance our understanding of radiotherapy beams by focusing on electron beams used in radiotherapy. It starts with a description of the Monte Carlo simulation code, BEAM, and a detailed simulation of an accelerator head to obtain realistic radiotherapy beams. The simulation covers electron beams from various accelerators, including the NRC research accelerator, the NPL (UK) accelerator, a Varian Clinac 2100C, a Philips SL75-20, a Siemens KD2, an AECL Therac 20, and a Scanditronix MM50. The beam energies range from 4 to 50 MeV. The EGS4 user code, BEAM, is extensively benchmarked against experiment by comparing calculated dose distributions with measured dose distributions in water. The simulated beams are analyzed to obtain the characteristics of various electron beams from a variety of accelerators. The simulated beams are also used as inputs to calculate the following parameters: the mean electron energy, the most probable energy, the energy-range relationships, the depth-scaling factor to convert depths in plastic to water-equivalent depths, the water-to-air stopping-power ratios, and the electron fluence correction factors used to convert dose measured in plastics to dose in water. These parameters are essential for electron beam dosimetry. The results from this study can be applied in cancer clinics to improve the accuracy of the absolute dosimetry. The simulation also provides information about the backscatter into the beam monitor chamber, and predicts the influence on the beam output factors. This investigation presents comprehensive data on the clinical electron beams, and answers many questions which could not be previously answered without the full simulation of a variety of medical accelerators. This study also demonstrates that the Monte Carlo technique is a powerful tool for the optimal design of accelerator treatment heads.

Acknowledgements

I would like to thank those who have made my education possible. Without the help of others, this thesis could never have been written.

I would like to start by thanking my parents who began my education. Their teachings were the source of my strength. To my sisters who helped me through the years of "the culture revolution" when my parents were in the "re-education camp", thank you!

To my fellow graduate students and research associates at the NRC, thanks for the support and friendship. To Jiansu Wei, my deceased classmate, thanks for the short but memorable experience of working with me on the OMEGA project. I would also like to thank Bruce Faddegon for many helpful discussions on writing the EGS4 user code.

I would like to thank the first rate scientists and support staff in the Ionizing Radiation Standards group. Their commitment to excellence in doing research influenced me greatly. I'm grateful to Carl Ross for carrying out the experiments on the NRC research accelerator which served as a milestone in validating the simulation code. I thank Alex Bielajew for his valuable discussions on Monte Carlo techniques, and for supporting the EGS4 system at NRC. Thanks also go to David Marchington for making slides on numerous occasions for conference presentations and Michel Proulx for teaching me many computer skills.

I especially thank Joanna Cygler for teaching and introducing me to practical clinical radiotherapy.

I would particularly like to thank Rock Mackie for many insightful discussions and the strong support he gave me to complete the project.

Among my teachers, I owe the largest debt to Dave Rogers, my supervisor. I was lucky to be his student. His professional attitude towards science, his vast knowledge,

and his dedication to perfection in publishing results influenced my work. I thank him for the numerous hours he spent suggesting changes to this thesis. I learned many valuable techniques during these busy and very enjoyable years we spent working together. I will always be grateful for his guidance to this thesis, and to my overall growth. His effort and influence will be with me for the rest of my career.

I would like to thank the members of my committee: John R. Cunningham, Paul Johns and Gary Slater who reviewed my thesis plan and gave many helpful suggestions at various stages. My thanks also go to the members of my thesis examining committee: Jerry Battista, John R. Cunningham, Paul Johns, Pat Kalyniak, and Gary Slater for their valuable suggestions to improve the thesis.

I would also like to thank the following individuals for supplying measured data and machine specifications related to various accelerator beams: Iain Bruinvis of the Netherlands Cancer Institute for data related to an SL75-20; Chen Chui of Memorial Sloan Kettering Cancer Center for data related to the Racetrack Microtron (MM50); Joanna Cygler of the Ottawa Cancer Clinic for data related to the KD2 and the Therac 20; Cal Huntzinger of Varian for the information related to the Clinac 2100C; Jack Janssen and Henk Huizenga of the Dr Daniel den Hoed Cancer Clinic for information related to the Siemens KD2; Rock Mackie and Carol Wells of the University of Wisconsin for data related to the Clinac 2100C; Carl Ross of NRC for data related to the NRC accelerator.

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This thesis is dedicated to my wife and best friend, Christine, for her invaluable support, sacrifice, and understanding throughout this long endeavor.

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List of Symbols

α Fraction of ionization due to electrons from the chamber wall

AECL Atomic Energy of Canada Limited

A_{ion} Ion-collection efficiency in the user's chamber at the time of ^{60}Co exposure calibration

A_{wall} Correction for attenuation and scatter in the wall and build up of user's chamber when exposed in air to ^{60}Co gamma rays

csda continuous slowing down approximation

Δ Cutoff energy in Spencer-Attix formulation of the Bragg-Gray equation

D_{max} maximum absorbed dose (usually on a depth-dose curve)

D_m absorbed dose to material m at a point of measurement in a phantom

%DD percentage depth-dose

d_{max} depth at which D_{max} occurs, also called R_{100} .

$\bar{E}_0$ mean energy of electron beam at phantom surface

EGS4 Electron Gamma Shower Version 4, a Monte Carlo code

ECUT charged particle cutoff energies in MeV (total energy)

ETRAN Electron TRANsport, a Monte Carlo code

$E_{0,p}$ most probable energy of electron beam at phantom surface

- ϕ_{med}^{water} A factor that relates electron fluence in water to that at a comparable point in a different medium (AAPM notation)
- h_m Correction factor to be applied when measurement are carried out at the ionization maximum in material m instead of in water (IAEA notation)
- I_{50} The depth at which the ionization falls to 50% of its maximum
- ITS** Integrated Tigger Series, Monte Carlo package of codes
- K_{el} correction factor to account for an electrode which is made of a material different from the wall of the chamber.
- K_h humidity correction factor for ion chambers.
- K_{pn} the point-source non-uniformity correction factor which accounts for the finite size of the chamber in the diverging r^{-2} field of the source.
- K_{st} Stem correction factor to account for the photons scattered into the chamber by the stem.
- K_{wall} A correction which takes into account the lack of charged particle equilibrium in an ion chambers wall near the cavity, the most important aspect of which is photon attenuation and scatter in the walls of the ion chamber. $K_{wall} = A_{wall}^{-1}$ and in some older literature is called k_{wall} .
- $\left(\frac{L}{\rho}\right)_{air}^{med}$ Ratio of averaged stopping powers for material med to air. the average is by the Spencer-Attix theory over the total electron energy spectrum at the point of measurement (AAPM notation)
- M** Reading of an ionization chamber system (IAEA notation)
- MeV** Mega electron-Volts, unit of energy.
- N_D absorbed dose to air chamber factor (this is N_{gas} of AAPM notation)
- N_{gas} Cavity-gas calibration factor (Unit: Gy/C or Gy/scale division)

N_x exposure calibration factor of an ionization chamber (R/C or R/scale division uncorrected for ion recombination)

P_{eff} the effective point of measurement of the ionization chamber (IAEA notation)

PCUT photon cutoff energies in MeV

PGES4 Preprocessor for EGS4

P_{ion} Ion-recombination correction factor applicable to the calibration of the user's beam

P_{repl} A factor that corrects replacement of phantom material by an ionization chamber

P_{wall} A factor that accounts for the walls being different from the medium and thereby changing the electron spectrum in the cavity.

PMMA Poly-methyl-methacrylate, $C_5H_8O_2$, also called lucite

p_s factor to allow for lack of saturation of charge collection (due to ion recombination) in an ionization chamber (IAEA notation)

p_{TP} Temperature and pressure correction factor

p_u factor to allow for non-water equivalence of the ionization chamber (i.e. chamber wall material and air cavity), when in the user's beam. An effective point of measurement is to be used when applying this correction factor (IAEA notation)

$\bar{S}/\rho$ Mean unrestricted collision mass stopping power

R_{100} Depth of dose maximum for electrons.

R_{50} depth at which dose falls to 50% of D_{max} .

R_{80} depth at which dose which is 80% of D_{max} .

R_p practical range of an electron beam; depth of intersection of a line drawn tangent to the depth-dose curve goes at the point of maximum slope and a line extrapolated from the bremsstrahlung tail

R_t therapeutic range, usually corresponding to the 85% level

SDD Source-detector distance

SSD Source-surface distance

$s_{m,air}$ Ratio of averaged stopping powers for material m to air. The average is by the Spencer-Attix theory over the total electron energy spectrum at the point of measurement (IAEA notation)

$\bar{\mu}_{en}/\rho$ Mean mass energy-absorption coefficient

W/e Mean energy expended per unit charge in air at usual humidity

X Exposure at position of chamber center, with chamber replaced by air (unit R)

Chapter 1

Introduction

1.1 Cancer Treatment and Radiotherapy

Cancer patients are treated by radiation, surgery or chemotherapy. A treatment method proving increasingly effective is radiation, used by itself or in combination with other modalities. The principal radiation modality used in cancer clinics is external irradiation with x -ray or electron beams, produced by electron accelerators. There are about 100,000 cancer patients each year in Canada alone and the number is increasing each year due to earlier diagnosis and aging population. About 50% of cancer patients are treated by radiation. Radiotherapy consists of brachytherapy (10%) and teletherapy (external irradiation, 90%). Approximately 10% of teletherapy patients are treated by electron beams.

The first electron linear accelerator for medical use was developed in the early 1960's in Great Britain, followed closely by developments in the United States. Nowadays, there are some 3500 medical linear accelerators in operation around the world.

When an x -ray beam or electron beam passes into an absorbing medium such as body tissues, some of the energy carried by the beam is transferred to the medium.

The measure of energy absorbed by the the medium is “dose” which is defined as the expectation value of energy imparted per unit mass. The transferred energy gives rise to a variety of highly reactive chemical species which in turn damage molecules of biological importance such as DNA, and so lead to cell death. The aim of radiotherapy in treating cancer patients is to deliver enough radiation to the tumor to destroy it without irradiating normal tissue to a dose that will lead to serious complications. Before a patient is treated with radiation a process called treatment planning is performed to determine the location and the amount of radiation to be delivered to the patient. For successful cancer radiotherapy, the uncertainty in the absorbed dose delivered to the tumor should be no larger than 3 – 5%. In the radiation treatment of cancer a treatment plan can only be made correctly if characteristics of the beam are known.

Radiotherapy treatment planning in general, and electron beam planning in particular, is very complex because scattering is strongly affected by changes in density and composition of the patient and the incident energy and angular distribution are rarely known. Contemporary pencil beam algorithms cannot properly account for electron scatter. The underlying Fermi-Eyges model assumes that all heterogeneities are infinite in lateral extent. Three-dimensional implementations of pencil beam algorithms, of which the Hogstrom Code is the most widely used, often deviate from measurement by more than 10% [1, 2] for even relatively simple geometries. More complex pencil beam algorithms [3, 1, 4, 5, 6] remove some of the consequences of infinite lateral extent of heterogeneities. Although more accurate results are achieved, significant approximations remain and some take almost as long to calculate as the Monte Carlo method. These pencil-beam models are all semi-empirical and there has been a great deal of effort to reduce assumptions, often employing Monte Carlo calculations to do so (*e.g.* Werner *et al.* 1982 [7], Lax and Brahme 1985 [8], Jette 1988 [9], Bruinvis *et al.* 1989 [10]).

1.2 The Monte Carlo technique

The Monte Carlo technique of radiation transport consists of using knowledge of probability distributions governing the individual interactions of electrons and photons to simulate their passage through matter. Scoring physical quantities of interest for a large number of simulated particles ("histories") provides information about average transport properties and the associated distributions such as the dose deposition. Since the Monte Carlo method uses well-established physics describing the nature of the interactions, we can expect the technique to yield accurate results. The results are statistical in nature but the statistical uncertainty can be reduced by running more histories. With increasing computing power available, many complicated Monte Carlo simulations become possible. With the development of new imaging technology, such as the CT (computer tomography) scanner, we have detailed knowledge of the geometry of the body to be irradiated. This could lead to more accurate determination of absorbed-dose distributions in patients by using Monte Carlo simulations. Furthermore, Monte Carlo simulations make it possible to perform "an experiment" under "ideal" conditions and this helps to understand the underlying physics behind the observations and guide further developments.

Monte Carlo calculations have been considered the "*correct*" way of doing many radiation dosimetry calculations and as a result they have been used extensively in the databases used for radiation dosimetry protocols such as those of the AAPM[11, 12] and the IAEA[13].

A Monte Carlo simulation code has four major components[14]:

1. the cross-section data for the particle's interactions for all the processes being considered in the simulation,
2. the algorithms used for the particle transport,
3. the methods used to specify the geometry of the problem and to determine the

physical quantities of interest,

4. the analysis of the information obtained during the simulation.

The first two components define the underlying physics of the simulation. When we speak of the EGS4 code system[15] we mean only the first two of these components. The EGS4 user codes which vary with applications are mainly written to deal with the last two components. The BEAM code developed in this project is an example of the EGS4 user code. The BEAM code will be described in detail in chapter 2. The history and features of the EGS4 code system have been described in detail in the SLAC-Report-265[15] and a recent review[16] of improvements is given in. The particles that are transported in the EGS4 system are photons and electrons (positrons). The following physics processes taken into account by the EGS4 code system[15]:

- bremsstrahlung production
- positron annihilation in flight and at rest
- Molière multiple scattering (i.e., Coulomb scattering from nuclei)
- Møller ($e^- e^-$) and Bhabha ($e^+ e^-$) scattering
- continuous energy loss applied to charged particle tracks between discrete interactions
- pair production
- Compton scattering
- Rayleigh (coherent) scattering can be included by means of an option
- photoelectric effect

In Monte Carlo simulation, as a photon passing through matter each discrete interaction is simulated while for an electron passing through matter this is very

difficult to do because of the large number of scatterings from atomic nuclei. For example, it is estimated[17] that electrons slowing from 500 to 250 keV undergo 4000 elastic scatterings in aluminum or 7000 scatterings in gold. To handle this situation, most electron transport algorithms use a condensed-history technique in which the electron is followed in a series of steps, and a multiple scattering theory is used to group individual elastic scattering events which occurred during the step.

As an example to illustrate the sampling in the Monte Carlo method, let's consider a simple photon transport model which includes only Compton scattering and pair production events, with the total interaction cross section at a given energy:

$$\Sigma_{tot} = \Sigma_{Compton} + \Sigma_{pair} \quad (\text{cm}^{-1}) \quad (1.1)$$

The questions are how far this photon can go before interacting and which interaction occurs? In order to determine them, the sample procedure begins with the selection of two random numbers R_1 and R_2 evenly distributed between 0 and 1. Since we know that the path lengths of the photons are exponentially distributed, we must sample from an exponential distribution having a mean path length given by $1/\Sigma_{tot}$. It has been proved that the variable

$$x = -\frac{1}{\Sigma_{tot}} \ln R_1 \quad (\text{cm}) \quad (1.2)$$

is exponentially distributed between zero and infinity with a mean value $1/\Sigma_{tot}$. Therefore using the random numbers R_1 we have determined that this photon will go a distance x cm and then interact. Next, we can determine which type interaction occurs by using the random numbers R_2 . If

$$R_2 \leq \frac{\Sigma_{Compton}}{\Sigma_{tot}} \quad (1.3)$$

it will be a Compton interaction and a pair production interaction otherwise. These are the two simplest but most often used sampling routines in Monte Carlo simulations.

The underlying principle of Monte Carlo method is the use of random sampling. It should be emphasized that the use of random sampling is not an approximation. It actually simulates physical reality since the inherent nature of virtually all the processes involved in particle transport is random. At best, the knowledge of each process amounts to knowing the probability distributions governing the event. Therefore the accuracy of the Monte Carlo method rests on the input of the various probability distributions which describe the physical processes involved and on any approximation which is introduced to sample these probability distributions.

1.3 Purpose of this investigation

The knowledge of the dose distributions from clinical beams is essential to accurate treatment planning for patients. This investigation aims to enhance our understanding of radiotherapy beams by focusing on electron beams used in radiotherapy.

The OMEGA (Ottawa-Madison Electron Gamma Algorithm) Project aims to develop an accurate electron beam treatment planning system. A major part of the project is to develop a general-purpose user code, called BEAM, to accurately simulate electron beams emerging from medical accelerators. A general-purpose system for simulating radiotherapy beams will also have many applications in absolute dosimetry. In the past, many dosimetry parameters were calculated using simple models for the incident clinical electron beams. For example, Andreo *et al.* [18] have demonstrated that the stopping-power ratios needed for electron beam dosimetry vary by as much as 2% at dose maximum because of *assumed* energy and angular distributions of the beam. With data of simulated realistic beams now available, we can eliminate the uncertainties of this kind. Similarly, the international IAEA dosimetry protocol bases the correction for use of plastic phantoms on experimental data whereas the AAPM (U.S.A) uses an analytic calculation. Huq and Nath[19] have shown that there is a significant discrepancy in dose (up to 4%) between these two approaches.

There are many other aspects of electron dosimetry, such as the mean electron energy, the energy-range relationships, the depth-scaling factors *etc.* which can be improved using the simulated clinical beams. As beam energy and field size are changed on a clinical electron accelerator, the output factor (the central-axis dose per monitor unit) changes significantly. This effect is due to radiation arising from collimators of an accelerator and backscattering into the beam monitor chamber. Experimentally it is difficult to determine the origin of the backscatter due to the complicated accelerator treatment head. The capability of simulating realistical clinical beams should allow those long standing issues to be resolved definitively.

1.4 Outline of the research project

Chapter 2 describes BEAM, a general purpose Monte Carlo code to simulate the radiation beams from radiotherapy units including high-energy electron and photon beams, ^{60}Co beams and ortho-voltage units and presents a variety of calculated results to demonstrate the code's capabilities.

Chapter 3 presents detailed information (i.e., energy and angular spectra) and characteristics of various electron beams from a variety of medical accelerators. The accuracy of the simulated clinical beams is demonstrated by comparing the calculated dose distributions with measured dose distributions in water phantoms.

Chapters 4 to 6 are devoted to the calculations of parameters used in electron beam dosimetry. Chapter 4 presents calculated mean energies at the surface of the phantom, the mean energy as a function of depth in phantoms, energy-range relationships and depth-scaling factors of a variety of electron beams used in radiotherapy. Data of calculated depth-scaling factors are provided for clear polystyrene, white polystyrene and PMMA phantoms.

Chapter 5 focuses on the calculation of Spencer-Attix water/air restricted mass

collision stopping-power ratios using the simulated realistic clinical beams and provides detailed data of calculated stopping-power ratios for realistic clinical incident beams. It also provides data for corrections to absolute dose in cancer clinics when current dosimetry protocols (AAPM[11] or IAEA[13]) are applied.

Chapter 6 presents calculated electron fluence correction factors ϕ_p^w as a function of depth for a variety of radiotherapy electron beams for converting the dose measured in plastics to the dose in water.

Chapter 7 provides the detailed information about backscatter into the beam monitor chamber and predicts the influence of the backscatter on the beam output factors.

Chapter 2

BEAM: A Monte Carlo code to simulate radiotherapy units

2.1 Introduction

The OMEGA project is a collaborative project to develop a full 3-D electron beam treatment planning system based on Monte Carlo simulation techniques to calculate the dose distributions in the patient[20, 21]. The rationale is that analytic methods of calculating dose in an electron beam suffer from various shortcomings which leave residual uncertainties or errors of 10% or more, even when using 3-D implementations of pencil beam algorithms[2, 22]. Advanced analytic models may remove some of these problems[5, 23, 24], but at the expense of very considerable computation time. However, in the end, the Monte Carlo technique for dose calculation is the gold standard, capable, in principle, of accurately computing the dose under almost all circumstances[25, 14, 26]. In particular Monte Carlo techniques can handle backscatter from prostheses, bone, internal shields, or scatter perturbations by air cavities much more accurately than any of the other current models.

The major drawbacks of using the Monte Carlo technique to predict dose distributions are computing time and the need for a detailed knowledge of the incident radiation beam. The actual code to compute the dose in the patient is relatively simple and a completely general code (DOSXYZ) capable of using CT data in each voxel was written in 1987 and has been used as a standard timing benchmark for the EGS4 code on a large number of computers[14, 27]. Using this code it was shown that a realistic treatment plan for a 20 MeV electron beam with $\pm 1\%$ statistical uncertainty in voxels with 2.5 mm sides would take about 600 hours on a VAX11/780, although the actual timing was dependent on a large number of parameters such as field size, voxel size, beam energy and accuracy required[14]. In the mid-90s, computers in many radiotherapy centers are tens or even hundreds of times faster than the VAX11/780 and the Monte Carlo code can be run in parallel on several machines, thus reducing the brute force calculation to a few hours. We have also been investigating ways to speed up the calculation using variance reduction techniques such

as correlated sampling[28] which provides a definite speed up in most situations or use of the macro Monte Carlo technique which is capable of increases in computing speed of about a factor of 10[29]. With future increases in computing speed and the development of routine procedures for farming out calculations to large numbers of computers, the Monte Carlo calculation of dose delivered to a patient will be achievable in a few minutes. This will still be longer than required for advanced analytic techniques, but, at minimum, the Monte Carlo technique will be of direct clinical use for "final" dose calculations or verification of other, simpler techniques which will be used in optimization or other exploratory calculations.

The problem addressed in this paper is thus that of calculating accurate information about the incident electron beam in order to initiate the transport of particles in the patient model. In another paper we address the issue of how to characterize these beams with sufficient accuracy for routine use[30]. It has been shown that to match accurately the measured central-axis depth-dose curves in a water phantom irradiated by "clean" 10 and 20 MeV electron beams[25], it is essential to include the complete energy distributions of the electron beam rather than just using the correct average energy from a point source[14]. Udale has further shown the need to include the correlation between energy and angle in the incident electron beam if good agreement is to be obtained on the central axis[31, 32, 33]. Our results confirm this result[30] although Udale's work concentrated on the 10 MeV beam from an SL75-20 which represents the worst case we have studied, and in general angular effects are small on the central axis (up to about 3% near the surface and less at depth). However, if the doses in the wings or penumbral regions are to be calculated accurately, we find one needs a detailed knowledge of the spatial distribution of electron fluence and to a lesser extent, the angular information about the beam. Other groups have demonstrated that Monte Carlo calculations can be done in 3-D patient models[34, 35, 36] but they have always used very simple models of the incident beam. In some cases this may be accurate, but in general it is not, and full models of the beam are necessary to assess the accuracy of the simple models.

The major emphasis in this project has been on simulating electron beams. These are more difficult to simulate accurately than photon beams. However, the code BEAM is equally applicable to photon beams from accelerators, ^{60}Co beams or even low-energy x-ray beams and a variety of capabilities related specifically to photon beams have been included. The code is also a very versatile, general purpose Monte Carlo transport package which can be used in a wide variety of applications other than simulating radiotherapy beams.

As well as being useful as the first step in accurate 3-D treatment planning[37, 38], detailed information about radiotherapy beams has a wide variety of applications in clinical physics and radiation dosimetry. For example, Andreo *et al.* [18] have shown that the energy and angular distributions of electron beams affect the stopping-power ratios need for reference dosimetry. Detailed simulations using realistic beams allow a more systematic study of these effects (see Chapter 5). Calculated information on photon beam characteristics can be very useful for a wide variety of photon dosimetry problems such as studies of electron contamination in photon beams[40, 41].

2.1.1 Previous Monte Carlo Models of Radiotherapy Beams

Application of Monte Carlo techniques to calculations of radiotherapy beams has a long history. ICRU Report 18 made extensive use of various calculations of the spectra from ^{60}Co sources[42]. More recently various groups have done sophisticated simulations of both ^{60}Co units[43, 44, 45] and photon beams from linear accelerators[40, 41, 46, 47, 48, 49] although in all cases these studies were restricted to models with cylindrical symmetry. Recent studies have shown that neither the EGS nor the ITS (ETRAN) codes can be used in default mode to calculate the angular distribution of bremsstrahlung photons near 0° [50, 51, 52]. Although both codes can give accurate results, the default versions have roughly 40% errors which are related to different problems (see section 2.3.5). Thus any previous calculations must be repeated if the

detail of these angular distributions plays a role in the results obtained.

The more difficult task of simulating electron beams has been studied by Udale[31, 32, 33]. Using the EGS4 code, she developed very extensive models which are no longer restricted to cylindrical symmetry. Udale studied 3 different accelerators and although her code was well structured, it could not be applied to other accelerators without a large effort. One interesting feature of her code's design was that it considers a series of more or less independent slabs, each slab containing a separate accelerator component. We have adopted and extended this modular approach.

2.1.2 Design Philosophy

To meet the goals of the OMEGA project, the BEAM code has been designed to simulate the radiation beams from any radiotherapy source, including low-energy x-rays, ^{60}Co units and both electron and photon beams from accelerators.

BEAM is based on the PRESTA extension[53] of the EGS4 Monte Carlo system for simulating radiation transport[15, 54, 14]. This code system is extensively documented and validated. EGS4 can be tailored to produce the results required within a required statistical uncertainty (i.e., with a wide variety of variance reduction techniques). The code is written in MORTRAN3, a Fortran77 preprocessor which is used for the EGS4 system[55]. Extensive use is made of the powerful macro capability of this pre-processor. The BEAM system runs under the Unix operating system using nearly 2000 lines of scripts.

One element of the basic design philosophy is that the BEAM code produces a phase-space output of the beam (*i.e.* the position, energy, direction, charge and history tag for each particle) at any specified plane. This phase-space file can either be re-used by the BEAM code itself (see below), used as an input to the patient dose computation algorithm directly, analyzed using various data analysis programs (such as the CERN package called PAW[56]) or used to characterize the beam in a

more compact way[30]. This approach retains complete flexibility without requiring a definitive *a priori* model to characterize the beam, albeit at the expense of large data files (≈ 100 Mbytes is common even in a compressed format). Once a beam characterization approach is developed, it may be added to the BEAM code so that the large data files may be avoided.

Another element of the design philosophy of BEAM is that the model of the accelerator head is built up from a series of individual component modules (CMs), each of which operates completely independently of the other component models and occupies a slab at right angles to the beam axis. Each CM deals with a specific class of geometric shape within the horizontal band, *e.g.* a simple set of slabs or a complex set of nested truncated cones. A CM does not specify explicit values related to the geometric shape (such as the number of slabs, their thickness or the materials they are made with) but allows all such possibilities and handles input of all the necessary data for a specific run. The independence of each component of the geometry package allows each CM to be tested and debugged in isolation. It also allows many very different designs of accelerators to be simulated because the various CMs can be used in a wide variety of configurations.

The first step in modeling a particular machine is to “specify” the machine by selecting, in order, the CMs to be used and also associating a unique name with each so that a particular CM can be used more than once. For example, a pair of collimator jaws and a pair of electron applicator scrapers may have the identical geometric requirements and thus they can both use the same CM coding. However the coding is used twice with a unique name attached to each use. Once the machine is “specified”, the Unix script “builds” the machine by pulling together all the relevant source code and automatically editing it appropriately to avoid redundant variable names and then compiling the code. Once compiled, this “machine” model can still model a wide variety of configurations since each simulation reads a user-supplied input file which contains all the specific information related to this run (*e.g.* the

initial electron energy distribution, the dimensions and materials of all parts of the machine, the location of output planes, how to track a particle's history, which variance reduction techniques to apply, what transport parameters to use, *etc*). This file can be interactively created. This approach was adopted because it maintains the ability to add new CMs and model new configurations easily, at the expense of having a very complex series of scripts to run the code. The other alternative was to write a single general purpose code which allowed any machine to be specified solely from a user-supplied input file. While this might be easier to use, it would inevitably require compromises in generality if unexpected components were required to simulate a new machine (a common experience) and it would be much harder to debug the geometry. Furthermore, by automating our scripts one step further (a straightforward task), we could achieve the goal of having a system driven by one input file. The machine specification and run-time input files would be concatenated and the accelerator code recompiled for each run. Fig. 2.1 shows the overall structure of the BEAM system and the steps involved in using it.

Another element of the design philosophy was to make the code so it could be used accurately by others. We have made a significant effort to document the code well, especially in the source code. For example, all variables and input requirements are defined within the source code and scripts are available to extract stand-alone files containing these descriptions. This was essential since for each machine "specified" the input file could be different. The code also outputs a file which can be used to produce a 3-D display of the machine geometry and particle histories using EGS_windows, a standard graphics package for EGS4 simulations[57]. This feature is very important for verifying the input of any user-supplied geometry.

Figure 2.1: The steps involved in using the BEAM system. To specify an accelerator means to define an ordered set of component modules (CMs) to be used in the simulation. To build an accelerator consists of gathering all the source code, automatically editing it to avoid duplicate names and compiling the resulting MORTAN and FORTRAN code. To do the simulation, BEAM reads in a cross section data set (prepared by the PEGS package from the EGS4 system[15]) and an input file which contains all the information related to that specific run: the complete geometry; the initial beam (either from the accelerator vacuum or from a previous run); the output required from this run; the simulation control parameters (cutoffs, forcing regions, *etc*). BEAM outputs phase-space data and graphics if requested and produces an output listing. The phase-space data can be used in off-line analysis programs or fed directly into patient or other simulation codes.

2.2 BEAM Capabilities

2.2.1 General Characteristics

The BEAM code models the therapy source with the z-axis taken as the beam-axis and usually the origin is defined as the center of the beam as it exits from the accelerator vacuum. The model consists of a series of component modules (CMs), each of which is contained between two planes which are perpendicular to the z axis and which can not overlap.

There can be an arbitrary number of scoring planes which are at the back plane of a CM and thus perpendicular to the z-axis. At these planes the particle fluences, average energies and average angles of all crossing particles are scored in an arbitrary number of circular rings or concentric square regions of arbitrary width, and a phase-space data file can be created at each plane.

Dose can be scored in an arbitrary number of regions, and the user can arrange to average the dose in an arbitrary group of regions. These regions may span several CMs.

As well as a selection of standard radiation source routines (pencil beam, point source, circular or rectangular beams), the simulations can be started using a phase-space data file as input. This allows the user, for example, to simulate the head of an accelerator down to the applicator and then reuse the same data with a variety of different applicators in place. When a phase-space file is re-used, particles which have crossed the scoring plane previously are not used to prevent double counting. Re-use of the phase-space data also allows the simulation of a ^{60}Co capsule in a separate calculation and then the use of the phase-space data file as an input to a model of the ^{60}Co therapy unit.

BEAM has a restart facility which allows a run with inadequate statistics to be restarted, or a run which was terminated before completion for some reason, to have

the data analyzed. This is done by writing all scoring results to disk at the end of each “batch” which is normally taken as 1/10-th of the total number of histories in a run. As well, full information about the state of the random number generator is stored at this point. Based on these data, the run can be re-started or analyzed up to the end of the last completed batch. Information about how many records are written to the phase-space file are also written out so that the phase-space data can be properly appended to the end of the previous batch (for the case of an interrupted run).

For all scored quantities, a statistical estimate of the uncertainty on the quantity is given by breaking the calculation into an arbitrary number of separate runs (usually 10) and then calculating the mean and variance of the mean to estimate the uncertainty (the method is discussed in detail in [14]). To allow for a restart facility, each batch in the run is broken into multiple separate calculations so that the statistical analysis is available after each batch.

The CMs are not allowed to overlap each other and any user request to do so causes an error message.

Each CM prepares output which can be used by the EGS_windows package to display the geometry. The actual display is set up to help understand an accelerator simulation rather than give an accurate representation of the physical structure. For example, when a primary collimator is simulated by a truncated cone, all that is displayed is the inner surface of the cone, not the exterior.

2.2.2 Geometric Considerations

A recurring problem in Monte Carlo codes is numerical roundoff problems as particles cross boundaries. The problem is that one must calculate the distance to the boundary and if numerical accuracy problems cause the particle to be transported just short of the boundary, or too far, the tracking algorithm can fail. To avoid this, we have used

various techniques: such as making obvious assumptions in some cases (*e.g.* when crossing a plane); or using "boundary buffers" which cause the boundary to always be slightly further away than it really is so that the calculation will always ensure the particle is just across the actual boundary. The tolerances are set to be reasonable for an accelerator simulation, but may fail if much smaller geometries are being simulated.

2.2.3 Component Modules (CMs) Available

The code has been written as a series of independent component modules. These are described in detail in a user's manual which is being developed with specific information about input requirements and details about restrictions. There is also a report being developed which describes the detailed specifications for writing a CM and doing the quality assurance on the CM. The following brief descriptions are meant to give an indication of the range of geometries which can be handled currently. For each CM, all the physical dimensions and materials involved are set by the user at run time. Thus in one use of a given CM the central region may be air with lead shielding on the outside, *e.g.* to represent jaws and in another application the central core may be lead and the outer region air to model a flattening filter or bolus. In most cases the CM extends to either an outer cylinder or a square box. The outer boundary is set separately for each CM and is constant for the CM. The CMs do not have to be contiguous, and any gaps between them are automatically filled with air throughout the CM.

The CMs have been named after the components they were originally written to model, but they can be applied to many more structures.

SLABS

This is the simplest CM but has many applications. It models parallel slabs in the x-y plane. They can have different materials and be of arbitrary thickness. The outer

boundary is square.

CONS3R

Fig. 2.2a) shows the component module CONS3R which models a stack of truncated cones (which can also be cylinders) treated as three regions. It is useful for modeling flattening filters where the inner region is a heavy material and the outer region is air, or conversely, for modeling a conical collimator where the central region is air. It can be used for many situations if there is cylindrical symmetry and if there are only two radial regions. If the cones are specified by a set of paired depth and radius values, the depth values are constrained to increase constantly *i.e.* they can not "double" back on themselves in the Z direction but they may in the radial direction. The outer boundary is a cylinder.

Although the CONESTAK and FLATFILT CMs can handle this geometry as a subset of more general cases, this particular CM executes much more quickly because there are no internal boundaries for particles to cross and range rejection (see below) is more efficient in large regions.

CONESTAK

Fig. 2.2b) shows the CONESTAK CM which models a series of stacked truncated cones (which can be cylinders). A cylindrical wall of arbitrary material surrounds the entire CM, just inside its outer cylindrical boundary. The radii of the cones must not decrease as the depth increases. The inner and outer conical regions of each layer have their materials specified separately.

This CM can be used to model scattering foils, primary collimators and many other components. This CM runs more slowly than CONS3R because there are far more boundaries for a complex case. However, it is applicable in many more situations than CONS3R. It is also a subset of FLATFILT, but the required inputs are somewhat

simpler to handle.

FLATFILT

Fig. 2.2c) shows the FLATFILT CM which is the most complex CM and is designed to handle some of the very complex beam flattening filters required in photon beam simulations. It is basically a generalization of CONESTAK to an arbitrary number of truncated conical sections on each layer and with the removal of the restriction about radii increasing with Z. In particular, FLATFILT can handle a flattening filter inside a conical collimator and the central component of the flattening filter can consist of a different material from the outside component.

CHAMBER

This CM models an ionization chamber and many other structures with cylindrical symmetry and centered on the beam axis (fig. 2.2d). For a monitor chamber, both the front and back walls may have several layers of different materials. The side walls consist of three layers, a chamber wall, a gap and a container wall. The central part of the chamber may consist of several air cavities separated by layers of other materials (serving as electrodes). This CM is also useful for central-axis depth-dose calculations in a phantom. Because there are fewer boundaries for the particles to cross in the side walls of this CM (i.e., the regions adjacent to the dose-scoring regions) than in other cylindrical-planar or voxel geometries, the computing efficiency can be greatly improved by using the electron range rejection technique (see section 2.3.4). A major feature of this CM is that it allows the scoring of dose which has been broken down into components based on the settings of the parameter LATCH (see discussion below on LATCH and discussion of fig. 2.12).

Figure 2.2: a) Component module CONS3R defined by 5 node points. Z_{MIN} is the distance from the reference plane ($Z = 0$) to the front plane of the CM (excluding the air gap).

b) Component module CONESTAK shown with 4 layers of stacked coaxial truncated cones, one pair per layer, with a cylindrical wall surrounding.

c) Component module FLATFILT shown with an outer conical shell and 4 layers of stacked truncated coaxial cones, with an arbitrary number on each level. The number of coaxial cones on a given layer and the material in each cone need not be the same. Both the number of cones and the radii of cones in each layer can be specified independently.

d) CHAMBER component module for an ionization chamber or any symmetric cylindrical-planar geometries. The CM consists of up to three parts. Both the top and the bottom part may have several flat layers of different materials; the materials may be different for the inner (circular) region and the outer (annular) region in the same layer. The central part is for the chamber, which may have several air cavities separated by layers of different materials and encircled by a chamber wall, a gap and a container wall.

PYRAMIDS

Fig. 2.2e) shows the PYRAMIDS component module which models a series of stacked, truncated pyramids. In this figure, the central region is the air gap which is partially indicated by the dotted lines. The four faces of each layer are parallel to the x or y axes but can have arbitrary orientations specified by the positive and negative x and y values of the front and back edges of each face (XFN, YBP, *etc*). The layers must have an air gap of at least 0.01 cm between them. Each layer can have different materials while the material inside the pyramid defaults to air.

The materials extend to the square outer boundary of the CM. This CM may be used for modeling rectangular jaws or collimators or beveled electron applicators.

APPSQ

Fig. 2.2f) shows the APPSQ component module which models a series of square applicators or scrapers. There can be air on the outside of the scrapers. The inner faces are all parallel to the z-axis as well as their respective x and y axes. The outer boundary of this CM is a square. There is an air gap of at least 0.01 cm required between each layer of applicator and surrounding materials.

This CM may be used for modeling the square scrapers found in electron beams although it does not allow for a beveled edge. This CM allows for leakage outside the scrapers, unlike the PYRAMIDS CM.

JAWS

Fig. 2.2g) shows the JAWS component module which models pairs of opposing flat surfaces which are perpendicular to the x or y axis. The X-jaws are defined as those which move in the direction of the x-axis. The jaw surfaces may be at an arbitrary angle with respect to the z-axis. This CM is primarily intended to model

Figure 2.2: e) The PYRAMIDS component module shown with two stacked, truncated pyramids. The positive and negative x and y values of the front and back edges of each face are defined by XFP, XFN, XBP, XBN and the same for Y, *etc.* The layers must have an air gap of at least 0.01 cm between them.

f) The APPSQ component module shown with two layers of truncated square columns or scrapers. Each scraper may have specified material and dimension. The minimum space between two layers of scraper is 0.01 cm of air.

g) The JAWS component module for case of paired x and y jaws which can be asymmetric (hence the need to specify the XFP, XBN, *etc.* Front and Back x values in the Positive and Negative directions). The distances to the top and bottom surfaces of each pair of jaws are given relative to the reference plane ($Z = 0$). The CM can model multiple sets of x or y jaws.

the movable collimator jaws in an accelerator. One simplification is that the upper and lower surfaces are perpendicular to the axis of the beam whereas in some clinical accelerators the jaws rotate about the source to maintain a sharp edge to the beam. This simplification is not expected to have any effect since the front face can be properly rotated. If found to be critical, thin JAWS above and below each main jaw could be used to model these surfaces accurately .

The outer boundary of the accelerator is square for the JAWS component module. It can not have jaws on all 4 sides at once, but can be set up with multiple sets of jaws in the x or y direction, one set behind the next. Each side of the jaw can be positioned independently and hence can be used to model a wedge.

MIRROR

Fig. 2.2h) shows the MIRROR CM used to simulate mirrors in the accelerator. The mirror is assumed to be one or more flat layers of different materials. It is taken to be uniform in the y-direction and varies with x (*i.e.* it is like the face of an x-jaw with multiple layers). Although accelerator mirrors are usually centered in the beam, the flat surface can be positioned arbitrarily in the x direction. It always extends to the outer y-boundary. The outer boundary is square.

XTUBE

This CM is needed to model an x-ray tube as the photon source and thus must be the first CM in any given simulation configuration. It simulates an x-ray target (possibly several layers) in air or vacuum backed by a target holder (backing material). The target angle is fixed by the target surface and z-axis (see fig 2.2i). Particles reaching the boundaries of the CM are assumed to be absorbed by the shielding except for those entering the second CM. Special source configurations have been written for this CM with the incident rectangular or circular electron beam coming in normal

Figure 2.2: h) The MIRROR component module shown with two layers of material which are perpendicular to the z axis, but rotated about a line parallel to the y axis and truncated by two planes perpendicular to the z axis. The material in each layer may be specified by user.

i) The XTUBE component module models an x-ray tube. The target may consist of several flat layers of different materials backed by a target holder (backing material). The target angle is defined by the target surface and the z-axis.

to the z-axis. It should be emphasized that although this allows an x-ray tube to be simulated, the EGS4 code does not handle impact ionization and thus is expected to underestimate the fluorescent lines from the target[58].

2.3 Other CMs

The above CMs are capable of simulating most standard accelerators, including accessories such as wedges. However, other CMs will need to be developed for applications such as multi-leaf collimators or non-rectangular cutouts or beam blocks. For example, a CM modeling a wire mesh has been written and applied in one study [59]. It is also possible to include bolus and other similar modifiers (e.g., compensator) in the patient simulation code, DOSXYZ.

2.3.1 Outputs from BEAM

BEAM has three major forms of output. The first, the listing file, is always present and optionally a phase-space data file can be produced and/or a graphics file can be produced which shows both the accelerator geometry in a stylized manner and every step in the particle's history.

Listing File

The listing file is the primary record of a simulation and is designed to make it possible to reproduce a given run by completely specifying all the input information plus an accurate specification of exactly what accelerator model was used. In particular, the entire system is handled under the Unix sccs library system and the exact version number of each piece of code used is echoed to the output listing and can be recovered using the sccs system if needed. In addition, all the other input control and geometric

parameters are listed. There is a summary of the accelerator regions which includes absolute region numbers for every region in the accelerator plus the local region numbers of each region in its component module since these are more easily understood. In this same table the mappings from geometric regions into dose zones are defined (*i.e.* the geometric regions belonging to each dose scoring zone are specified) along with the mapping between geometric regions and bits set in LATCH (see section 2.3.2).

The listing file contains several standard outputs. For each scoring plane defined in the accelerator model, a variety of fluence related parameters are scored in an arbitrary number of nested squares or circles. The number and widths of the scoring regions are selected on input, as are the locations of the scoring planes. For each particle type (electrons, positrons and photons) the total number of particles, the proper average particle fluence (cm^{-2}), the average energy and the average angle with respect to the z-axis are scored along with the estimated uncertainties on each. The particle fluence is scored by summing the inverse of $\cos(\theta)$ where θ is the angle with respect to the z-axis as the particle crosses the plane (capped at 85° , see [14]). This method of scoring fluence has some limitations in the case of low-energy electrons especially and is only meant to give a reasonable estimate. The final output is the dose and energy deposited in an arbitrary number of dose zones (which can include multiple regions).

Phase-Space Data File

The phase-space data files contain the data about each particle crossing the scoring planes. These are usually the most important output. The exact contents of the file depend on the application, but a general purpose FORTRAN subroutine for reading and writing the phase-space files has been written and can be used by other programs. The standard files contain the variables LATCH, E, X, Y, U, V, SIGN(W), WT, IQ, and NPASS. LATCH is the history variable discussed below, E is the particle's energy, X and Y are the position coordinates in the scoring plane (the Z value is a constant for

a given scoring plane and not stored), U and V are the direction cosines with respect to the X and Y axes respectively, $\text{SIGN}(W)$ is the sign of W , the direction cosine of the angle with respect to the Z axis (where the magnitude of W is recovered from the fact that $W = \sqrt{1 - (U^2 + V^2)}$), WT is the weight of the particle, IQ is the charge of the particle (-1 , 0 or $+1$) and $NPASS$ is a variable specifying whether this is the first or later time that the particle has crossed this scoring plane. A slight compression is achieved by storing $\text{SIGN}(W)$ and $NPASS$ in 1 bit each and IQ in two bits. Each record contains 28 bytes and thus a file with 1 million particles requires 28 Mbytes of disk space. Rather surprisingly, these files do not grow much as lower-energy electrons are tracked in an electron beam because there are so few of these electrons (see section 2.4).

Graphics Output File

Most component modules are set up to output a simplified representation of the geometry for input to the EGS_windows graphics package[57]. On request the user can output a complete history file which tracks every step in the simulation or output a data file which can be read by EGS_windows and displayed in three dimensional space. This feature can be very instructive and useful in debugging. The data files can become very large and only a few histories at a time should be followed.

Fig. 2.3 shows representations of four of the accelerators studied in this work. Even though the representations are somewhat stylized, these figures immediately give considerable insight into the models being used. When viewing a color terminal with the capability of rotating and zooming the image, this tool becomes invaluable.

2.3.2 Tracking each particle's history

One of the major advantages of the Monte Carlo technique is that it allows detailed information about each particle's history to be known. To allow flexible access to

this information, BEAM includes a general technique built upon the **LATCH** feature of EGS4[15]. **LATCH** is a variable which is passed on to a particle's descendants and which the user can interrogate or modify. In BEAM, each bit of **LATCH** is manipulated separately. Bits 1 to 23 are set whenever a particle interacts in a region (the region-to-bit correspondence is set by the user in the individual input files). For an electron, passage through a region automatically constitutes an interaction on within the region. For a photon, there are two options; the photon is either tagged the same way as an electron or only when an interaction actually occurs within the region. The **LATCH** technique allows one to separate the effects of particles which have hit the applicator, or the jaws *etc.*

Bit zero of the variable **LATCH** is set whenever a bremsstrahlung photon is created. Thus it is possible to interrogate any particle about whether there was a bremsstrahlung photon created anywhere along its history trajectory.

The last piece of information which is stored is the region number in which the particle was created(bits 24–28). This identifies the sources of secondary electrons in an electron beam and is useful for determining the region in which bremsstrahlung photons are created.

The **LATCH** feature is an option and requires setting up a bit-to-region correspondence in the input file. This feature requires some extra computing time, in practice the extra time is negligible(<1%).

2.3.3 Calculation of Charge from Ion Chambers

One of the features of BEAM, which no other accelerator model has included, is the direct calculation of the dose deposited in the air in the internal monitor chamber. This feature will allow studies to be made of the variation in output factors with field size and energy. The accurate calculation of absolute ion chamber response is perhaps the most difficult problem one can attempt with Monte Carlo simulations

in the sense that at the 1% level, present calculation techniques don't work very well[14, 60]. This is not a problem for radiotherapy beam calculations, but it can be critical for ion chamber response calculations used for absolute standards. However, even if the absolute accuracy is somewhat uncertain, relative currents(absorbed dose to air in the ion chamber) should be calculated accurately. This feature has not been fully exploited yet, but is based on use of the CHAMBER component module which allows separation of the dose to the air into a variety of components caused by particles which have followed different types of histories. For example, studies have been done which show that the amount of backscatter into the current monitor decreases substantially as the electron beam energy is increased but is independent of the jaw settings for a Clinac 2100C (see chapter 7).

2.3.4 Variance Reduction Features

Range Rejection

Range rejection can save significant quantities of computing time for electron transport calculations[14]. The basic method is to calculate the residual range of a charged particle and terminate its history if it cannot escape in any direction from the current region. The BEAM code includes a subroutine, *MXRNGE*, which is used to pre-compute the residual ranges to the threshold energy (*AE* MeV, total energy) in each medium as a function of electron energy. From these tables, the residual range to the lowest energy for which an electron is transported (*ECUT* MeV, total energy) can be worked out since *ECUT* values may vary in different regions. These residual ranges represent the pathlength traveled by electrons slowing to *ECUT* if they undergo no discrete interactions and thus are calculated using the restricted stopping powers, *i.e.* they represent the maximum possible pathlength the particle could go in the simulation, and are, in general, larger than the residual CSDA ranges calculated with the larger unrestricted stopping powers. This is a conservative approach since very few particles

actually go anywhere near these maximum possible residual ranges, both because of the discrete interactions, which cause significant local energy loss, and because of the multiple scattering of the electrons which cause torture paths.

Range rejection involves an approximation, *viz.* any bremsstrahlung photons that would have been generated by the electrons as they slow down, are assumed not to escape from the region. The effects of this approximation can be minimized by setting a maximum energy for which a history can be terminated (called **ESAVE** in **BEAM**) and thereby allowing the higher energy electrons a possibility of creating bremsstrahlung which escapes from the region, even if the electron itself could not escape from the region. Another option is to disable range rejection in a region in which there is significant bremsstrahlung production (such as the bremsstrahlung target) but do it everywhere else.

The appropriate selection of **ESAVE** is dependent on situation. For initial electrons of 20 MeV incident on Tungsten, simple radiative yield arguments show that about 11% of the bremsstrahlung energy fluence is created by electrons with energies below 5 MeV and only 2% by electrons below 2 MeV. In this case, **ESAVE** of 2 MeV is acceptable, especially since attenuation of low-energy photons will reduce the amount of bremsstrahlung from low-energy electrons even further. For initial electrons of 10 MeV, the corresponding 5 and 2 MeV values are 32% and 7% and thus one would need to use an **ESAVE** value of 1 MeV to reduce the error to 2% or less.

The above range rejection approach is very general but in **BEAM** we can save further time by terminating any history which cannot reach a scoring region. This is hard to specify in general but as a partial answer to the problem, a threshold energy, called **ECUTRR**, is automatically calculated for each region such that an electron must be able to escape the region with energy greater than **ECUTRR**. If its energy is less than **ECUTRR** as it leaves the region, the particle cannot reach the scoring region of interest. For example, say we are interested in all electrons with energies down to 10 keV at the base of the accelerator. As an electron leaves a scattering foil 50 cm from the base,

assuming only air in the way, an electron needs a kinetic energy of at least 200 keV to get to the base of the accelerator. Thus, there is no need to track electrons inside the scattering foil unless they will leave it with at least 200 keV. Thus when doing range rejection in the foil, BEAM terminates the history if the range of the electron as it slows down to 200 keV is less than the distance to the nearest boundary. Taking into account even thin monitor chambers and other materials in the beam saves time if electrons are being tracked to low energies. This is because, to first order, it takes as long to track an electron from 200 keV to 100 keV as from 20 MeV to 10 MeV. BEAM can automatically compute the minimum energy needed by electrons as they leave each CM if they are to reach the base of the accelerator and this can be the energy used in the range rejection routines. However, this option must not be used if, for example, one wants to calculate the charge created in the machine's monitor chamber and thus needs to track low-energy electrons inside the accelerator.

Table 2.1 shows a variety of timing results for simulations of an 18 MeV beam from the Clinac 2100C studied extensively in this paper. If we take case 1 as the base case, a variety of comments are in order. Firstly, these times are for an SGI Indigo with R4400 cpu but when run on a SUN Sparcstation 2, the case 1 calculation takes exactly 5 times as long. Thus, for this very different type of EGS4 simulation, we find that timing comparisons based on the standard EGS4 benchmark code [27] are applicable here. However, absolute timing comparisons only apply for calculations done on otherwise un-used machines and we find variations in cpu time per history of up to 50% depending on the computer's load. Case 2 *vs* case 1 shows that for $ECUT = 0.700$ MeV, use of the automatically increased values of the low-energy cutoff as one moves further away from the scoring plane (as in case 1) saves only a few percent compared to leaving the low-energy cutoff fixed. Case 3 *vs* case 1 shows that allowing electrons with energies between 5 and 2 MeV to be transported, even when they cannot escape from their current regions, adds very little to the computing time but it also slightly increases the overall number of bremsstrahlung photons in the beam, reflecting that these electrons between 5 and 2 MeV are producing some

photons but far fewer than the 9% one might expect based on the simple arguments given above. As expected, the number of electrons in the beam is unchanged. Case 4 shows that turning off range rejection altogether (by setting $\text{ESAVE} = 0.0$) increases the computing time by about a factor of two, with virtually no change in the number of photons or electrons in the beam compared to case 3. Cases 7, 8 and 9 indicate a similar trend if electrons are being tracked to even lower energies except that in this case (0.521 MeV instead of 0.700 MeV), range rejection improves the efficiency by a factor of 6 and tracking electrons from 5 to 2 MeV adds an extra 17% to the already longer calculation times. Comparison of case 7 and case 7a (where the values of the low-energy cutoff are not automatically increased) shows that this feature can lead to considerable improvement (30%) in computing speed when ECUT is low.

When simulating photon beams we find that large numbers of bremsstrahlung photons are generated in the target and subsequently absorbed deep in the primary collimator. Very few of these escape except those “near” the beam. To avoid senselessly tracking these photons in the collimator, we find it useful to introduce a very thin region on the upstream side of the collimator. This region prevents all particles from entering the top of the collimator except for those actually in the beam or nearby. This form of range rejection is very crude but can save up to 20% in computing time for photon beam calculations.

Bremsstrahlung Splitting

For the simulation of clinical photon beams from a therapy machine, a variance reduction technique called “particle splitting” [50, 14, 61] can be used in BEAM to increase the number of bremsstrahlung photons created in the simulation geometry. At each bremsstrahlung interaction site, N photons are sampled individually and their weights are reduced by a factor of $1/N$. The statistical uncertainty in the photon energy fluence spectrum scored for a given number of incident electrons can be reduced dramatically since most of the time in these simulations is taken up track-

ing electron histories. Since one is normally interested in photon spectra when using bremsstrahlung splitting, BEAM can use Russian Roulette[62, 14] on the electrons created by the multiplicity of bremsstrahlung photons. In essence, Russian Roulette means the number of electrons followed is restricted to the same number as would occur if bremsstrahlung splitting was not used (*i.e.* if N photons are created in each bremsstrahlung event, BEAM kills off $N-1$ out of N secondary electrons created by these photons and increases the weight of the survivors by a factor of N).

The increase in computing efficiency obtained using bremsstrahlung splitting is very dependent on the parameters of interest, as is the choice of an optimal splitting number. In realistic photon beam calculations we find it useful to use a splitting of 10 to 20 and can see increases in efficiency of up to a factor of 10.

Forced Interactions

One variance reduction technique, forcing photon interactions, has been shown to be very helpful in some cases[63, 14]. It is particularly helpful when studying electron contamination in accelerator beams since virtually no photons interact in the air and yet that is a major source of the electron contamination[64, 43]. The way around this is to force photons to interact in the geometry. This can be very efficiently implemented in any EGS4 user code because of the general structure of the code[65, 14]. However, one must take care to ensure that after the photon is forced to interact the original photon continues with a reduced weight so that the fluence of particles is not overscored anywhere in the geometry. In the implementation in BEAM, photons can be forced to interact in any subset of component modules.

2.3.5 Bremsstrahlung Angular Distributions

The default version of EGS4 deflects a bremsstrahlung photon by a fixed angle with respect to the electron generating the photon. This angle is the mean deflection angle

and is about 1 to 2° for 15 MeV electrons. Since multiple scattering of electrons in high-Z materials is generally much larger than this, the default version of EGS4 assumes that the small effect of the bremsstrahlung angular distribution is washed out by the electron multiple scattering. It is, except on the first step. This can cause significant problems for the angular distribution of bremsstrahlung photons from clinical accelerators. An extension to EGS4 which removes this problem has been utilized here[50]. This extension has up to a 40% effect near the central axis. The angular distribution of bremsstrahlung photons from high and low-Z materials has been shown to be in excellent agreement with measured angular distributions, but only if the extension is used[51].

It may be worth noting that a problem of a similar magnitude has been identified with the default version of the ITS system and great care must be taken with that system as well to get the angular distribution of bremsstrahlung photons correct near the central axis[52].

2.3.6 Incident Beams

BEAM can handle input beams of electrons, photons or positrons. The selection of energies and geometries of the sources is generally decoupled except when a phase-space data file is being used as an input. These phase-space data files can be read in at any plane between CMs. In general this is to allow restarting calculations part way through the machine. This facility could also be used to input an absolutely arbitrary source by creating a phase-space file. The user also has the option of using a generalized source routine by writing a subroutine (with well defined requirements) which is used by BEAM. This option is used to allow characterized beams from earlier runs to be used as input sources[30].

If phase-space data or the generalized source routine are not used, the user has the option of specifying an arbitrary energy spectrum for the incident beam.

BEAM handles a wide variety of input source geometries, most of which are assumed to start at the top of the accelerator. The most straightforward is a pencil beam of particles (usually electrons) incident along the z-axis at the origin. A slight extension is a parallel circular or rectangular beam incident on the first plane at an arbitrary angle. Another possibility is a point source on the accelerator axis at some arbitrary distance, uniformly irradiating a circle on the first plane.

There are three source geometries which simulate moving electron beams. The first models a beam which is used at NRC to achieve flat photon beams without the need for a field flattening filter. The electron beam is incident on the first plane of the accelerator in a small circle centered on the axis and is swept to generate an incident cone of electrons. The geometry is defined by the radius of the circle and the half-angle of the cone. In the absence of any scatter in the accelerator, the electrons would sweep out a narrow ring about the beam axis. The second moving beam simulates the scanned beam from an AECL Therac 20 in which the beam is swept in a saw-tooth manner across the field of interest. The pattern is defined by the size of the region scanned at 100 cm from the accelerator exit window and the ratio of x and y scanning frequencies. Fig. 2.4 shows the fundamental scanning pattern, once with no air present so that the underlying swept-beam characteristics are clear, and also with 100 cm of air present to demonstrate that just the air scattering in this swept 10 MeV beam is adequate to give a flat beam. The model simplifies reality because all the angles for sweeping the beam are selected prior to the beam entering the accelerator whereas in reality, the beam is swept after it has passed through the vacuum exit window. The third moving beam is for modeling the scanned electron beam from MM50 accelerator and is arranged so it would sweep out a uniformly irradiated circle 100 cm from the start of the accelerator if there were no materials in the way.

For completeness the BEAM code has a source routine for simulating an orthovoltage x-ray unit which is described in section 2.2.3.

2.4 Selection of Run parameters

Selection of various run-time parameters for any Monte Carlo calculation can be very complex[14, 15, 66, 53] and often depends on what aspect of the results are most important and how many resources are to be consumed by the calculation.

The BEAM code uses the PRESTA algorithm for electron transport[53]. For calculations down to $ECUT = 700$ keV (total energy), comparison of cases 1, 11 and 12 in table 2.1 suggests that utilizing the default PRESTA algorithm produces comparable results (to the 1% uncertainties) to calculations with steps sizes also restricted to ESTEPE values of 4% and 1%. For other calculations it may be important to use ESTEPE of 1% or even less[60] but at the expense of 35% or more computing time.

Another parameter of interest is AE, the lowest total energy of secondary electrons. Lower values simulate more interactions and create more realistic energy-loss straggling distributions. If studying energy spectra *per se*, it is important to use a low value of AE (0.521 MeV), but if calculating depth-dose curves, much higher values are completely accurate[65]. Case 10 compared to case 1 in table 2.1 shows that using an AE value of 0.521 MeV instead of 0.700 MeV when using a value of $ECUT = 0.700$ MeV has virtually no effect on the results shown, but increases the computing time by 43%. It also significantly affects the high-energy electron fluence spectrum and thus the lower value should only be used if energy spectra are of interest.

Comparison of cases 1, 5 and 6 for $AE = 0.700$ MeV and cases 7 and 10 for AE of 0.521 MeV show that increasing ECUT, the low-energy cutoff for electron transport can reduce computing times substantially, especially from lower values (*e.g.* going from $ECUT = 0.521$ to 0.700 MeV with $AE = 0.521$ MeV, reduces the computing time by a factor of about 3). However the rather surprising result is that the number of particles exiting from the accelerator actually decreases somewhat for the lower values of ECUT whereas one would expect the inclusion of the small number of low-energy electrons to increase the total number of electrons. Although the effect is

just barely statistically significant, the unexpected result must come via a very weak dependence of the PRESTA algorithm on the value of ECUT[53]. We have chosen to do our calculations for $ECUT = 0.700$ MeV, but based on this table, it might often be wise to use a much higher value unless effects of low energy electron reaching on the patient skin are an important part of the study.

2.5 Validating the Code for Electron Beams

The problem of checking a code of this complexity and size is formidable. Just the Unix scripts to run the system are longer than the original EGS4 source code and with 20,000 lines of FORTRAN (with all comments removed), the code to simulate a typical accelerator is bound to have some mistakes in it. The issue is to demonstrate that they are not critical to the results. This has been done by extensively testing each individual CM in a wide variety of both typical and highly unusual situations against the results of widely used NRC EGS4 user codes (DOSRZ, FLURZ and DOSXYZ) which have been extensively benchmarked[14]. The first two codes are restricted to cases with cylindrical symmetry and the latter to rectilinear voxel geometries. These obviously do not test all possible situations, but because each component module can be tested separately, one can cover a wide variety of cases. Much of the code has also been checked by simply having more than one person read the code and its documentation. The final test is to use BEAM to simulate the beams from a wide variety of accelerators and calculate the dose distributions in a homogeneous water phantom. We then compare calculated results to measurements. Comparisons to measured data in phantoms with heterogeneities are being presented elsewhere [67].

Comparison to experiment is complicated by the fact that clinical accelerators have many physical parameters which users usually do not know. The most important of these may be the energy spectrum and direction of the electrons leaving the accelerator vacuum. This is discussed in section 2.8 and for the moment all beams

in vacuum are assumed to be monoenergetic and on-axis. For many machines this is a good approximation. For example the Varian 2100C specifications indicate it delivers a beam with a maximum energy width of $\pm 3\%$. The incident beam energies for simulations of clinical electron beams are usually selected by matching the value of the calculated and measured values of R_{50} , the depth at which the dose falls to 50% of the dose maximum. Another option would be to match R_p , the projected range. We have calculated depth-dose curves for electron beams with a mean energy of 20 MeV and symmetric energy widths varying from 5 to 20%. For widths up to 10%, the energy width had little effect on the values of R_{50} or R_p but above that width, the value of R_p started to change since it has some sensitivity to the highest energy in the spectrum. In other words, the final depth-dose curve contains little usable information about the details of the *incident* electron spectrum unless the spectrum is very broad. We thus chose to match R_{50} since this is the easiest procedure in practice. It is interesting to note that we get very good agreement with the measured depth-dose curves by using this procedure (see below) but this may only reflect the insensitivity of the final depth-dose curves to the *incident* spectrum. In the case of our worst agreement (the SL75-20, 20 MeV beam, see below) we calculate a depth-dose curve which falls off too slowly past the dose maximum and a broad incident spectrum would only make it fall-off more slowly. Thus the discrepancy is not caused by the width of the incident spectrum although Deasy has shown that electron energy spectra from these machines do have a considerable width[68].

We have obtained detailed information about a variety of accelerators. Often, when we failed to get agreement between the experiments and calculations we found that we were given incorrect information about a given accelerator because not all accelerators are identical, even for the same models. Thus the data calculated here apply to the specific machines discussed. Many accelerators are adjusted in the clinic to match previous accelerator beams (by selection of scattering foils or tuning of the waveguide) and manufacturers keep improving the hardware (e.g. developing thinner monitor chambers). Thus this BEAM code (as opposed to phase-space data sets) is

of great value in predicting the effects of such customization.

In the following sections we will describe the simulation of electron beams from 5 different electron accelerators and comparison of the measured dose distributions in homogeneous water phantoms with those calculated using the NRC EGS4 user-code DOSXYZ[14, 27] which has been modified to use the full phase-space data file from BEAM as an input. DOSXYZ is a very simple Monte Carlo code which calculates the dose in rectangular 3-D voxels when irradiated by a beam described by the phase-space output from BEAM. One option which is available in DOSXYZ is to "re-use" the phase-space data as often as required to improve the statistical precision of calculated depth-dose curves. The phase-space file always contain enough particles (several million) to ensure accurate representation of the incident beam. While re-using the beam seems counter productive, we find it leads to a distinct improvement in the precision of the calculated dose distributions. The random numbers ensure each particle follows a different path in the phantom. This should be compared to simpler situations in which incident particles have identical values for some or all parameters (*e.g.* when calculating the dose in a monoenergetic broad parallel beam). In DOSXYZ, if the incident beam is assumed to be symmetric about the x and y axes (a common situation), then when re-using the phase space the particles are shifted into the 4 symmetric positions ((x,y) to (x,-y) with (u,v) to (u,-v) *etc*).

It should be emphasized that these comparisons require no approximations or models of the output obtained from the BEAM code. For disk space and other reasons, most previous comparisons of electron depth-dose curves broke the calculations into two steps[25, 32, 33, 69]. The first step was to simulate the accelerator and to score the energy, spatial and sometimes angular distributions of the electrons from the accelerator. The second step was to calculate the central-axis depth-dose curve in a phantom using various approximations to generate the electron beam incident on the phantom, such as independence of energy, angular and/or spatial distributions. Udale[32, 33] examined the effects of using different levels of complexity and Andreo

and Fransson[69] demonstrated explicitly that ignoring correlations between energy and angular distributions leads to inaccuracies in the reconstructed incident beam. All these concerns are overcome in our case by using a brute force approach since the calculation retains all the phase-space information for every particle. In related work, the issue of accurate representation of these beams is studied in detail[30].

These comparisons are restricted to electron beams since they are generally more difficult to simulate accurately than photon beams. Furthermore, various previous studies have demonstrated that EGS4 simulations predict accurate dose distributions for photon beams, even with relatively simple models[46, 43, 44, 47, 40, 45].

2.5.1 Electron beam simulations and comparisons with measured dose distributions

NRC Accelerator

The NRC accelerator is a Vickers research accelerator with no head assembly. The energy of the electron beam exiting the vacuum window is known independently with an accuracy of $\pm 1\%$ [70]. The accelerator has a 0.127 mm titanium exit window, a 0.092 mm thick tungsten scattering foil, a very thin mylar-walled monitor ion chamber, square steel jaws which define a field size of $8 \times 8 \text{ cm}^2$ on the phantom surface at an SSD of 96 cm. The water phantom has a 9 mm wall of PMMA and the calculations explicitly account for this. The measurements have been done with a small diode detector, a parallel plate ion chamber and a cylindrical Farmer chamber. After correction of the ion chamber results for stopping-power ratios, all three measurement techniques gave the same results well within 1%.

The upper panel of fig. 2.5 shows a comparison between the calculated and measured depth-dose curves when the incident accelerator energy in the simulation is that determined independently (*viz.* 20 MeV). There is a slight discrepancy in the

values of R_{50} but otherwise the agreement is excellent. The figure also shows a similar comparison with the incident beam energy increased in the simulation by 1.5%. In this case even the value of R_{50} is in close agreement with experiment. The required incident beam energy is only 1.5 standard deviations from the measured value and thus the measured energy may be the problem. Alternatively, a 1.5% error in the water electron stopping power would not be too surprising and this would affect the calculated depth-dose curve, although it would be somewhat surprising if the stopping power is 1.5% too small at all energies[71].

Note the close agreement for the bremsstrahlung tail.

The lower panel of fig. 2.5 also presents a comparison of the calculated and measured dose profiles in the NRC beam measured with a diode detector. All data are normalized to the peak of the depth-dose curve. The agreement is remarkable. To our knowledge this represents the first ab-initio calculation which properly reproduces the dose profiles in a realistic electron beam.

Varian Clinac 2100C Accelerator

Fig. 2.6 compares the calculated and measured central-axis depth-dose curves for the 6 and 18 MeV beams from a Clinac 2100C at University of Wisconsin. These machines have dual scattering foil systems and this particular accelerator has the scattering foils of a Clinac 1800 in order to match the depth-dose curves of an earlier machine. It also has the relatively thick walled gold-plated mica ion chamber and a Type II applicator cone with 4 relatively thin scrapers. In these calculations the incident electron energies in the simulation are adjusted to give the measured values of R_{50} . The agreement is very good, well within 3% of R_{max} everywhere. At 6 MeV the calculated surface dose appears to be somewhat high and at 18 MeV the calculated dose fall-off may be somewhat too steep and the bremsstrahlung tail slightly underestimated. The 18 MeV results suggest there is something in the beam which is not

included in our model or the scattering foils are slightly thicker than their nominal values since more material would increase the bremsstrahlung tail and increase the straggle which would make the fall-off less abrupt. A comparison of measured and calculated dose profiles (in this case at $d_{\max} = 1.9$ cm in the 18 MeV beam) is shown in fig. 2.7. The calculated and measured data are in good agreement, even outside the beam (see section 2.7 for a discussion of what causes the dose outside the beam).

AECL Therac 20 Accelerator

Fig. 2.8 compares the calculated and measured central-axis depth-dose curves for the 9 and 20 MeV beams from the AECL Therac 20 at the Ottawa Regional Cancer Centre. This accelerator scans the electron beams to achieve beam flatness and, as can be seen from fig 2.3, has much thicker applicator scrapers than the Clinac 2100C. Both these features lead to a much cleaner beam and very different central axis depth-dose curves. The agreement between calculation and measurement is excellent. This indicates that BEAM is capable of accurately modeling clinical accelerators as different as the Clinac and Therac in their electron spreading technique. Note that for the Therac 20 the bremsstrahlung tail is much less than for the Clinac and other scattering foil accelerators although the calculations also underestimate it slightly in the 20 MeV beam.

Philips SL75-20 Accelerator

Fig. 2.9 compares the calculated and measured central-axis depth-dose curves for the 10 and 20 MeV electron beams from a Philips SL75-20 accelerator which uses only a single set of scattering foils at any particular energy (measured values and machine specifications from Udale's thesis[31]). The applicator on this accelerator has solid flat walls compared to the series of scrapers in the applicators of the AECL Therac 20 and Clinac 2100C. This accelerator is again very different from the previous

ones but the agreement between the calculations and measurements is very good. Udale also obtained good agreement between calculations and experiment for the 10 MeV beam[33]. For these beams the bremsstrahlung tail is very accurately predicted although, as mentioned above, the calculated fall-off in the 20 MeV beam is slower than experiment.

Racetrack MM50 Accelerator

Fig. 2.10 compares the calculated and measured central-axis depth-dose curves for the 25 and 50 MeV beams from the Scanditronix MM50 at the Memorial Sloan Kettering Cancer Center. The agreement is excellent. The MM50 uses both beam scanning and a very thin scattering foil to give a uniform beam distribution. The beam sizes are larger than for the lower-energy beams to ensure broad-beam conditions. Unlike other accelerators, this machine does not have electron applicators at these beam energies. The beam is shaped by a primary collimator and a multileaf collimator which for the present case is set for a square. The beams from this accelerator are close to monoenergetic beams because of two factors: the scanned beam and the large air space between last part of the beam defining system and the phantom surface. This air gap means many of the electrons scattered from the beam defining system leave the field.

Summary of Experimental Comparisons

The above comparisons show that the BEAM code is capable of calculating the electron beams from a wide variety of accelerators very accurately. The depth-dose curves can be calculated with an accuracy which is certainly better than 2% of D_{max} on average. To achieve a tighter comparison would require more accurate measurements since most measurements ignore changes with depth of various factors and these may have a 1 or 2% effect. It is also worth noting that our calculations appear more

accurate for the scanned beams. This may just reflect that they are much simpler beams or the fact that it is easier to get complete information about the accelerator because there is so much less information required.

2.6 Electron Beam Spectra and Other Characteristics

BEAM allows the energy distributions of electrons and photons to be studied in great detail. Variations in spectra based on location in the beam or last point of scatter can be important and are being investigated [30].

Fig. 2.11 presents the energy spectra for the standard electron beam energies for each of the 4 accelerators studied in detail (a non-standard Clinac 2100C with a Type II applicator, SL75-20 with a solid walled applicator and the scanned beams from the Therac 20 and MM50 Racetrack Microtron). These are planar fluence spectra averaged over the open beam which is usually $10 \times 10 \text{ cm}^2$ and at an SSD of 100 cm (details in previous section). The plots are shown on a logarithmic scale to display the data at low energies but it must be emphasized that on a linear plot the scanned beams appear to be mono-energetic whereas the machines with either single or dual scattering foils exhibit significant low-energy tails.

When calculating energy spectra, it is essential to use a lower threshold for the production of secondary electrons ($AE = 0.521 \text{ MeV}$) since this ensures adequate energy-loss straggling. If a threshold of 700 keV is used, as done for the depth-dose calculations, there is a distinct artifact in the energy spectrum (see [66] for a complete discussion). In the Clinac 2100C spectra, the lower-energy peak about 2 MeV below the main peak is caused by electrons passing through one applicator scraper and remaining in the beam. This is discussed further below. For the SL75-20, there is a distinct double peak for the higher energy beams. The scattering foils consist of

circular foils of varying radii and the first of these does not intercept the entire beam. In a separate run we have established that the upper, smaller peak comes from those electrons which "miss" this initial foil and thus lose less energy.

As well as comparing our results to experiment, we have compared our results for the SL75/20 to those of Udale[32, 33]. When we compare mean energies at the phantom surface for $10 \times 10 \text{ cm}^2$ applicators, we get reasonable agreement at 5 MeV but differ by nearly 400 keV at 10 MeV. Our calculations include electrons inside the field down to 189 keV kinetic energy whereas hers appear to include the entire phantom surface and include all electrons down to 50 keV. While these differences may explain the differences in calculated mean energies, they are not important when we compare the electron energy spectra near the peak energy. Here we see significant differences which may be related to her use of an AE value of 50 keV (kinetic energy) compared to our use of 10 keV which gives more accurate energy loss straggling in the BEAM calculations. We have done calculations with $\text{AE} = 700$ (189 kinetic energy) and 10 keV and found significant differences due to the artifacts mentioned in the previous paragraph, but it is not clear that these would explain all of our differences with Udale's results.

Table 2.2 presents a variety of data related to the 8 accelerator beams for which calculated and measured depth-dose curves have been compared. The third column relates the number of electrons hitting the patient per cm^2 to the number of incident electrons. For fields of 100 cm^2 these numbers give the percentage of electrons getting to the patient. The percentages vary greatly, from around 2% for the low-energy beam from the Clinac to around 13% for the high-energy beams from the MM50. These variations are exactly what is expected since electrons in machines with scattering foils and lower-energy electrons scatter more.

The number of photons per cm^2 given in column 4 is seen to vary even more, being highest for high-energy beams from machines with scattering foils and being lowest for scanned beams from machines which do not define the beam with scrapers

(viz. the MM50).

Column 5 shows the cpu time on a Sun Sparcstation 2 per (electron/cm²) at the patient surface since this takes into account different field sizes and the fact that the precision of calculations for dose distributions depends directly on the incident fluence. These timing figures vary considerably and are very sensitive to the details of each calculation (see the discussion related to table 2.1) and the exact machine model being used. Several general observations can be made. A major determinant of computing time is the number of electrons reaching the patient per particle incident (see col 2). The 6 MeV beam from the Clinac 2100C requires the longest computing time. In contrast, the 50 MeV beam from the MM50 also requires a long time, mostly because the high-energy electrons which get stopped in the beam defining system take a very long time to track (their long ranges make the range rejection algorithms ineffective)

Columns 6 and 7 give the mean kinetic energies of the electrons (above the cutoff of 189 keV kinetic energy) and contaminant photons in the beams. These and similar data form the basis of an extensive study of electron energies in clinical beams[72]. The final two columns show the percentages of the electrons and photons which are "direct" in the sense that they have not scattered off the scrapers or jaws but have come directly through the beam region. Both the scanned beams have a large proportion of direct electrons by comparison with foil-scattered beams. This is an indication of the effectiveness of their collimation systems at removing electrons completely, rather than scattering them, *e.g.*, the Therac 20, has roughly the same overall transmission as the scattering foil accelerators. The converse of this is that nearly all of the photon contamination in the Therac accelerator comes from the scrapers which stop most of the unused beam, and hence these photons are not "direct". For the scattering-foil accelerators, most of the photon contamination comes from the scattering foils.

A report is available which presents complete spectral and other data for electrons and photons for up to 5 beam energies from each of the electron accelerators discussed

in this paper (see chapter 3). [73]. It must be emphasized that data for one particular accelerator cannot necessarily be applied to another accelerator of the same model.

2.7 Electron Beams from a Clinac 2100C

In sections 2.5 and 2.6 we presented results for a variety of accelerators to demonstrate the ability of the code to model them accurately and as a source of useful information. In this section we present a more detailed analysis of the beams from a Clinac 2100C, both to demonstrate the capabilities of BEAM and to investigate the physics of electron beams.

Fig. 2.12 shows the 18 MeV depth-dose curve from the Clinac 2100C and breaks the curve into various components. Electrons which do not scatter from any part of the accelerator outside the beam are referred to as direct. It can be seen that these produce somewhat more than 70% of the dose and define the fall-off region of the curve. Electrons which scatter from the applicator scrapers or main jaws both contribute roughly 10% to the dose near the surface, but are clearly of lower energy. The electrons from the applicator are the least penetrating since some of them go through the scrapers. The photon tail can be seen to be primarily generated by contaminant photons incident on the phantom as opposed to bremsstrahlung photons generated in the phantom. The figure also shows the dose from incident secondary electrons which are defined as those knocked on by any of the others. These low-energy electrons contribute very little to the dose (2% or less) and play no role past a few mm.

Fig. 2.13 shows the dose components in beam profiles at depths of 1 and 10 cm (which is just past R_p). Near the surface it can be seen that the major contribution to the dose outside the beam comes from electrons passing through the applicator along with photon contamination to a lesser extent. At depth it is seen that on-axis about 60% of the dose comes from the contaminant photons generated in the accelerator

(mostly the scattering foil). The contaminant photons also extend into the wings whereas the dose from the bremsstrahlung generated in the phantom by the direct electrons is relatively flat but mostly within the beam.

Fig. 2.14 shows the components of the energy spectra in the 9 MeV beam from the same Clinac 2100C fitted with either the older Type II applicator or the new Type III Varian applicator design as described by Kassae *et al.*[74] who have recently published a study including Monte Carlo calculations using the cylindrically symmetric code CYLTRAN[75]. Consistent with Kassae *et al.*'s results, our spectra show clearly that the new applicator design, which gives a significantly cleaner beam, avoids generating electrons from the applicator. The low-energy peak in the old design is just over 2 MeV below the main peak and this corresponds to electrons which pass through a 6 mm aluminum scraper which defines the beam about 31 cm from the patient (this was shown in a separate run which separated the spectra from each scraper component). By calculation it was found that the peak could be made to disappear using an additional piece of lead on the top of this part of the applicator. The calculated depth-dose curve with the lead in place was virtually identical to that of the new design applicator. We have been advised that Varian sells an upgrade kit which performs the same function.

2.8 Problems

Despite the abilities of the Monte Carlo technique to do very accurate simulations, there are a variety of practical problems involved.

The most important is getting adequate information about the specific accelerator to be modeled. Manufacturers are reticent to provide the necessary level of detail because of the obvious commercial value. Furthermore, accelerator designs are constantly improving, even within a given model and thus one set of information will not cover *e.g.*, all Clinac 2100Cs. Finally, the machines are often adjusted for individual

purchasers, often to match a previous machine's characteristics.

Another area of potential concern is adequate knowledge of the phase space of the electron beam as it leaves the accelerator vacuum. Our method of varying the energy of a mono-energetic incident beam to match R_{50} allows for accurate calculation of dose distributions, but there may be other, more subtle effects, for which this approximation fails. Thus our calculated energy spectra can be thought of as a smearing function for the accelerator and not necessarily an accurate representation of the final spectra. Nonetheless, as discussed in section 2.5, this uncertainty is not thought to affect dose-distributions significantly.

Another short-coming of the present technique is the large data files involved with phase-space files. This problem should be solved by detailed models of the beam which will allow much smaller data files to represent the beams accurately[30].

2.9 Summary and Conclusions

The code BEAM has been described in some detail and extensive comparisons to measured dose distributions in water phantoms have shown that it can reproduce these well within 3% of D_{max} at all depths. Extensive data concerning the characteristics of electron beams from 4 commercial accelerators of very different designs have been presented. Similar work is underway concerning photon beams from commercial accelerators[76]. The ability of the code to help design and modify accelerator equipment is obvious.

The major strength of the BEAM code is its flexibility and extensibility. It is possible to put together, with little effort, models which will cover very different accelerators. In fact, the system represents a powerful tool for building a variety of complex geometries for simulating many situations which are not accelerators.

The code will be made generally available to the medical physics community as a

research and educational tool. The current version requires a Unix operating system, lots of disk space (> a 1Gbyte), especially if generating phase-space files, and the NRC Unix-based EGS4 system must be in place with the “Big” Mortran pre-processor.

Table 2.1: Timing variations for simulation of the 18 MeV electron beam from a Clinac 2100C. The beam is 10×10 cm² at an SSD of 100 cm but the particles are scored to a radius of 14 cm. Times are for an SGI Indigo with an R4400 cpu which runs at 55 VUPs (*i.e.* 55 times faster than a VAX11/780) for the EGS4 benchmark code (which can be compared to many other machines via the data in ref[27]). Values of ECUT are automatically increased further from the scoring plane to improve range rejection efficiency, except where noted. AE is the threshold for the production of secondary electrons (total energy, kinetic energy=AE-0.511 MeV), a lower value implies more accurate energy-loss straggling. Electron histories are followed down to a total energy of ECUT(MeV). Electrons below energy ESAVE (total) are subject to range rejection, 5 MeV implies more than at 2 MeV, and 0 MeV implies no range rejection is used. The “total to file” is the number of particles written to file per incident particle ($\pm 0.5\%$). The last two columns show the breakdown between electrons and photons written to file($\pm 1\%$, except where noted). Latch values are scored for all cases and data are written to a phase space file. Default PRESTA is used except where noted.

Table 2.1:

Case	AE (MeV)	ECUT (MeV)	ESAVE (MeV)	CPU s per history	total to file/inc	e ⁻ per 100 inc	γ per 100 inc
1	0.700	0.700	5.0	0.0124	0.417	9.15	32.4
2	0.700 ¹	0.700	5.0	0.0127	0.411	9.09	31.9
3	0.700	0.700	2.0	0.0125	0.420	9.07	32.8
4	0.700	0.700	0.0	0.0249	0.421	9.03	33.0
5	0.700	0.900	5.0	0.0118	0.417	9.09	32.5
6	0.700	1.100	5.0	0.0083	0.411	9.13	31.9
7	0.521	0.521	5.0	0.0538	0.414	8.79	32.6
7a	0.521 ^{1,2}	0.521	5.0	0.0735	0.413	8.51	32.7
8	0.521	0.521	2.0	0.0631	0.416	8.83	32.7
9	0.521	0.521	0.0	0.300	0.421	8.94	33.1
10	0.521	0.700	5.0	0.0178	0.416	9.09	32.4
11	0.700 ³	0.700	5.0	0.0172	0.414	9.01	32.3
12	0.700 ⁴	0.700	5.0	0.0120	0.413	9.23	32.0

¹ Automated variation of ECUT away from scoring plane not done.

² Ten times fewer histories run, thus statistical uncertainties are 3 times worse.

³ ESTEPE = 1%.

⁴ ESTEPE = 4%.

Figure 2.3: Four commercial electron accelerators (Varian Clinac 2100C, AECL Therac 20, Philips SL75-20 and Scanditronix MM50) as shown by EGS_windows . with about 100 histories.

Table 2.2: Various data associated with electron beam simulations for 4 accelerators. Most particle fluences (planar) and average kinetic energies refer to the average inside a $10 \times 10 \text{ cm}^2$ field at an SSD of 100 cm. For the Therac 20 the scan pattern was set for a $40 \times 40 \text{ cm}^2$ field but the scored field was $10 \times 10 \text{ cm}^2$ while for the MM50 the scan pattern was set for a circle of 18 cm radius and a field size of $25 \times 25 \text{ cm}^2$ (25 MeV) and $20 \times 20 \text{ cm}^2$ (50 MeV). The EGS4 parameters ECUT, AP, and ESTEPE are 0.700 (total) and 0.010 MeV and 1%, respectively. Direct electrons are those which do not scatter from anything outside the beam. Times are for a SUN Sparcstation 2 which runs at 11 VUPs for the EGS4 benchmark code (which can be compared to many other machines via the data in ref[27]).

Linac	Beam Energy (MeV)	e^- / cm^2 / 10^4 /inc e^-	γ / cm^2 / 10^4 /inc e^-	cpu s/ (e^- / cm^2) on patient	$\bar{E}(\text{MeV})$		% direct e^-	% direct γ
					e^-	γ		
CL2100C	6	2.12	2.08	250	6.11	1.36	78.9	69.6
	18	7.56	12.47	66	17.35	2.81	80.8	63.7
T20	9	5.64	1.24	50	8.25	1.30	95.9	16.3
	20	6.35	4.63	50	18.74	2.17	94.8	6.0
SL75-20	10	4.14	4.07	96	9.34	1.96	66.8	75.6
	20	6.94	13.60	68	17.76	3.35	69.5	78.5
MM50	25	13.6	0.80	21	24.53	3.50	99.6	50.9
	50	12.7	2.63	105	48.90	5.59	99.1	42.0

Figure 2.4: Scanned beam pattern for 10 MeV electrons at 100 cm from the accelerator exit window. In the upper figure the electrons pass through vacuum after being scanned and in the lower figure they are scattered by 100 cm of air. The field size is clipped at 12×12 cm².

Figure 2.5: Comparison of measured and Monte Carlo calculated depth-dose curves on the NRC research linac with the incident energy in the simulations taken as the independently known energy of the electron beam leaving the accelerator vacuum (20.0 MeV) or increased by 1.5% to match the experimental data. The depth-dose curves are measured with diodes and ion-chambers. The ion-chamber readings are corrected with stopping-power ratios from the AAPM TG-21 protocol. The lower panel compares BEAM-calculated and dose profiles measured with diodes at the depths shown by the vertical lines in the upper panel. Energy used in the simulation is 20.3 MeV. The first 9 mm of the phantom are PMMA.

Figure 2.6: Comparison of measured and BEAM/DOSXYZ calculated central-axis % depth-dose curve for 6 (upper) and 18 (lower) MeV electron beams from the Varian Clinac 2100C accelerator at the University of Wisconsin. The field size is 10×10 cm² at an SSD of 100 cm. The measurements are made with a diode detector. This accelerator has dual scattering foils, the monitor chamber is made with gold plated mica and the applicator has 4 sets of scraper bars.

Figure 2.7: Comparison of Monte Carlo calculated and measured dose profile at $d_{\max}$ (≈ 1.9 cm) for the 18 MeV electron beam from the Varian Clinac 2100C accelerator at the University of Wisconsin. The field size is 10×10 cm² at an SSD of 100 cm. The measurements are made with a diode detector.

Figure 2.8: Comparison of measured and BEAM/DOSXYZ calculated central-axis % depth-dose curve for 9 (upper) and 20 (lower) MeV electron beams from an AECL Therac 20 accelerator. The field size is $10 \times 10 \text{ cm}^2$ at an SSD of 100 cm. This accelerator has a scanned electron beam (with a $40 \times 40 \text{ cm}^2$ scan pattern), the monitor chamber is made with aluminized Kapton and the applicator has thick scraper bars. Measured data are from Joanna Cygler of the Ottawa Regional Cancer Centre.

Figure 2.9: Comparison of measured and BEAM/DOSXYZ calculated central-axis % depth-dose curve for 10 (upper) and 20 (lower) MeV electron beams from an SL75-20 accelerator. This accelerator has single scattering foil and the applicator has solid walls. The field size is $10 \times 10 \text{ cm}^2$ at an SSD of 100 cm. Measured data and information on the accelerator are from Udale[31].

2

PM-1 3½"x4" PHOTOGRAPHIC MICROCOPY TARGET
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Figure 2.10: Comparison of measured and BEAM/DOSXYZ calculated central-axis % depth-dose curve for a 25 MeV (upper) and 50 MeV (lower) electron beam from the Racetrack MM50 accelerator. This accelerator has a scanned beam (which would uniformly irradiate a circle of radius 18 cm at the phantom surface) with a very thin scattering foil and no scrapers. The field sizes are $25 \times 25 \text{ cm}^2$ and $20 \times 20 \text{ cm}^2$ at SSDs of 100 and 110 cm for the 25 and 50 MeV beams respectively. The measured dose curves come from measured ionization curves which are converted to dose using stopping power ratios calculated using the same BEAM simulation (see chapter 5). The measured ionization data are from Chen Chui, Memorial Sloan Kettering Cancer Center.

Figure 2.11: Energy spectra from the 4 clinical accelerators studied in detail here. The spectra are for planar fluence averaged over the entire beam (usually 10×10 cm^2). Note that the y-axis is logarithmic and that on a linear plot the scanned beams are virtually monoenergetic and the scattered beams contain a considerable low-energy tail. These spectra are for incident monoenergetic beams and any width in the incident beam from the accelerator vacuum would add to the breadth of the final spectra. All spectra are calculated with $\text{ECUT} = 0.700$ MeV and $\text{AE} = 0.521$ MeV (total energy) to ensure adequate energy-loss straggling.

Figure 2.12: Calculated components of depth-dose curve for Clinac 2100C with beam energy 18 MeV, field size = 10×10 cm² and SSD = 100 cm. The y-axis is offset from zero for clarity. Direct electrons are those which have hit no walls or the applicator. Electrons from the jaws or applicators are those which have interacted in the respective components at some point (and hence include some double counting). The contaminant photons and secondary electrons are those incident on the surface of the phantom. The central-axis scoring region is 2×3 cm². Thresholds of $AE=ECUT = 0.700$ MeV are used.

Figure 2.13: Components of dose for dose profiles for an 18 MeV electron beam at depths of 1 and 10 cm for for a Varian Clinac 2100C, field size = $10 \times 10 \text{ cm}^2$ and SSD $\approx 100 \text{ cm}$. Components have the same meaning as in fig. 2.12. The profile at 10 cm depth is past the range of the incident electrons but the bremsstrahlung associated with a given class of electrons is included in that component. The fluctuations are statistical.

Figure 2.14: Energy spectra from Clinac 2100C with old and new designs of applicators. The spectra for electrons from various components as defined in previous figures are shown. The peak at 7.8 MeV in the lower panel for the older, Type II design applicator can be seen to come from electrons which hit the applicator (and mostly come through the third layer up, as shown in other runs) and the other components change very little.

Chapter 3

**Energy spectra, angular spread
and dose distributions of electron
beams from various accelerators
used in radiotherapy**

3.1 Introduction

The energy and angular distributions of electrons and contaminant photons from medical accelerators are the most important characteristics of a radiotherapy electron beam. A knowledge of clinical beams is essential for dosimetry, treatment planning, quality assurance, and design of an accelerator. In the radiation treatment of cancer the treatment plan can only be made correctly if characteristics of the beam are known. Beam quality has influence on the physical, chemical and radiobiological effects as well as on the response of various detectors. Due to the variations in the design of beam applicators design the properties of scattered electrons are complicated[77, 78] and these scattered electrons may have a large influence on the resultant dose distributions in a patient. Experimentally it is difficult to measure detailed information because of various limitations in the clinical environment and detectors[68]. In addition to this, it is virtually impossible to use experimental methods to distinguish electrons which are scattered by the beam defining system, from those scattered only by air.

The EGS4 Monte Carlo simulation code, BEAM, is a powerful and flexible tool to simulate clinical radiation beams and to obtain a detailed knowledge of the characteristics of therapy beams from accelerators. A previous study[77] described the code and simulation in detail and demonstrated that BEAM can predict dose distributions very accurately. Here we will systematically analyze the simulated beams from a variety of electron accelerators including the NRC research accelerator, the University of Wisconsin Varian Clinac 2100C, a Philips SL75-20, the Ottawa Regional Cancer Center Siemens KD2 and AECL Therac 20 and the Memorial Sloan-Kettering Cancer Center Scanditronix Racetrack MM50. These accelerators have very different designs and some of them produce scanned electron beams instead of scattered beams. The energy range of beams is from 5 to 50 MeV. Figs. 3.1 to 3.8 are the representations of six of the accelerator's treatment head studied in this work as shown by

EGS_windows[57]. The figures show the actual simulation of particles transported inside the treatment head.

For beams produced by various accelerators this report contains detailed information on the energy and angular distribution of: all electrons; scattered electrons from each beam defining component; and contaminant photons. Comparisons between calculated and measured dose distributions in water phantoms are presented along with calculated dose components contributed by contaminant photons and electrons scattered from beam defining components.

It must be emphasized that results apply to the specific accelerators studied and should not be taken as applying to all similar machines of the same model (e.g. the University of Wisconsin Clinac 2100C had special foils to match beams with an earlier Clinac 1800).

3.2 Monte Carlo Calculation

The EGS4 user code BEAM[77] is used to simulate accelerators and obtain detailed information about each particle emerging from an accelerator. The position, energy, angle and a complete history of where a particle has been are stored in a "phase-space" data file. The phase-space data files are used repeatedly for analyzing beam or as input to the EGS4 user code DOSXYZ[14] to calculate the dose distributions in a water phantom.

By using the variable LATCH[77] which records each particle's complete history in the beam simulation we are able to obtain not only the energy spectra of all electrons and contaminant photons but also the spectra of electrons and photons scattered from collimators or applicators of the beam defining system.

The DOSXYZ code has been modified to read a phase-space file produced by the BEAM simulation as an incident beam and to make use of the variable LATCH

to calculate dose components contributed by scattered electrons and contaminant photons in the beam.

The beam field size studied here is $10 \times 10 \text{ cm}^2$ for beams with energies below 22 MeV and $25 \times 25 \text{ cm}^2$ or $20 \times 20 \text{ cm}^2$ for those with energies above 22 MeV. The SSD is 100 cm and 110 cm for energies 5 - 40 MeV and 50 MeV, respectively. The incident monoenergetic electron energies at the vacuum-exit-window are adjusted to match R_{50} , the depth at which the dose falls to 50% of dose maximum on the measured central-axis depth-dose curve in a water phantom.

The spectra presented in the following are for incident monoenergetic beams and any spread in the incident beam energy in front of the vacuum-exit-window of an accelerator would add to the breadth of the final spectra [39].

In the beam simulation the energy cutoffs for particle transport are set to $\text{ECUT} = 0.700 \text{ MeV}$, $\text{AE} = 0.521 \text{ MeV}$ (total energy) and $\text{PCUT} = \text{AP} = 0.010 \text{ MeV}$ to ensure adequate energy-loss straggling. In the dose calculations the energy cutoffs for particle transport are set to $\text{ECUT} = \text{AE} = 0.700 \text{ MeV}$ (total energy) and $\text{PCUT} = \text{AP} = 0.010 \text{ MeV}$. In all cases, the PRESTA electron transport algorithm[53] is employed with $\text{ESTEPE} = 0.01$ or 0.04 .

3.3 Results and Discussion

3.3.1 Overview and statistics of the results

There are about 10^6 electrons inside the beam field in the phase-space file of a simulated beam. Since a phase-space file contains particles outside the beam field as well as contaminant photons, the actual number of particles in a phase-space file varies depending on the characteristics of the beam.

The two computer codes, PHSP_EXY and PHSP_SPEEXY[73], are used to ana-

lyze the phase-space files from BEAM to obtain the mean energy and energy spectra of simulated beams at the surface of the patient. The codes allow the user to select particles according to particle's history by making use of the LATCH feature of BEAM[77].

The spectra and angular distributions of electrons as well as photons presented in the following figures are all normalized to 1 at the peak of electron spectra. The bin size of spectra and angular distributions is 100 keV and 1° for all beams. The angle θ is relative to the z-axis which is the central-axis of the beam. The figure captions each specify how many electrons are in the bins normalized to 1 for energy and angular distributions. One can thereby determine n , the number of electrons in each bin, and hence the fractional statistical uncertainty which is given by $\sqrt{n}/n$ (1 standard deviation).

The dose distributions are all normalized to 100% at the maximum of the total dose which is contributed by all electrons and contaminant photons in the beam. The statistical uncertainties on the depth-dose covers are typically 1% or better.

Table 3.1 summarizes characteristics of simulated clinical electron beams from five accelerators. The average kinetic energies $\bar{E}$ refer to the number averaged inside the irradiated field at the phantom surface. $E_{o,p}$ is the most probable energy inside the field at the surface of the phantom. Direct electrons are those which scatter only in air and do not include electrons scattered from the beam defining system of a treatment head. The difference in surface doses between total and that from direct electrons is indicative of the number and angle of scattered electrons in the beam. R_{50} and R_p are obtained from calculated central-axis depth-dose curves. The % surface doses of total (contributed by all particles in the simulated beam) or direct (contributed only by those electrons that do not hit any jaws, collimators or applicators) is relative to maximum total dose. E_{max} for electrons and photons is the maximum energy of electrons (kinetic energy) and photons in the phase-space file respectively. The difference between the maximum electron and contaminant photon energy reflects the

total thickness of vacuum-exit-window, scattering foils, beam monitor chamber, mirror and air in the beam. The maximum contaminant photon energy is approximately the kinetic energy of the electrons incident on the vacuum-exit-window.

Table 3.2 presents the simulation data of clinical electron beams from five medical accelerators. E_m is the incident electron kinetic energy at the vacuum exit window of an accelerator. It is adjusted to match the calculated R_{50} to that of the measurement. $e^-/100$ inc e^- and $\gamma/100$ inc e^- inside field are the number of electrons and number of photons inside the field per 100 incident electrons at the vacuum-exit-window. These data can be used to estimate the number of incident electrons (at the vacuum exit window) that the simulation has to run to produce the required number of electrons at the phantom surface and inside the beam field. % direct e^- and % direct γ are defined as:

$$\% \text{ direct } e^- = \frac{\text{number of direct electrons inside field}}{\text{number of total electrons inside field}} \quad (3.1)$$

$$\% \text{ direct } \gamma = \frac{\text{number of direct photons inside field}}{\text{number of total photons inside field}} \quad (3.2)$$

where direct electrons or photons are those that do not hit any jaws, collimators or applicators. The CPU per incident history is machine dependent as well as simulation parameter dependent[77]. The data presented in table 3.2 are for an SGI Indigo with an R4400 CPU.

It is interesting to note that there is almost no build-up in the component of the central-axis depth-dose curves contributed by electrons scattered from applicators. This is due to the combination of the broad angular distributions and the low-energy spectra of electrons from applicators.

3.3.2 Electron beams from the NRC research accelerator

The NRC accelerator is a Vickers research accelerator with no head assembly. The geometry of the experimental set up (fig.3.1) includes a titanium vacuum-exit-window

(0.127 mm thick), a tungsten scattering foil (0.092 mm thick), a square collimator (steel, 20.3 cm thick) and a monitoring ion chamber (very thin mylar). The water phantom container is made of PMMA (9 mm thick). The beams irradiate the water phantom horizontally. The field size is 8×8 cm² on the phantom surface and SSD = 96 cm. The energy of the electron beam incident on the vacuum-exit-window is known independently with an accuracy of $\pm 1\%$ [70]. The two beams (10.0 and 20.0 MeV) have an identical experimental set up except for the electron energy incident on the vacuum-exit-window. The dose measurements have been done with a small diode detector, a parallel plate ion chamber and a cylindrical farmer chamber. After correction of the ion chamber results for stopping power ratios, all three measurement techniques gave the same results within 0.5%. The measured data were obtained from Carl Ross of National Research Council of Canada.

The results for these two beams are shown in fig. 3.9 and 3.10 respectively. Because of the geometry of the experimental set up, the electron's energy spread is very small and there are few contaminant photons in the beams. The two beams have narrow angular spread although the electrons from the jaws lead to a small tail extending to 80° . Note that in order to match the experimental values of R_{50} in the central-axis depth-dose curves the energies of electrons incident on the vacuum-exit-window used in the simulation were increased by 1.5% compared to those determined independently (*viz.* 10 and 20 MeV). This slight discrepancy may be caused by the measured energy or by the stopping power used in the calculation (to be resolved in future investigations). The remarkable agreement between calculated and measured dose distributions shown in fig.3.10 marked a milestone in validating the simulation code BEAM[79].

3.3.3 Electron beams from a Varian Clinac 2100C accelerator

The Varian Clinac 2100C accelerator (fig.3.2) has an exit window, scattering foils, a beam monitor ion chamber, a mirror, a "shielding plate", X and Y jaws and an electron applicator. This particular accelerator at the Wisconsin Comprehensive Cancer Center in Madison has thicker scattering foils (same as in an older Clinac 1800) in order to match the depth-dose curves of an earlier machine. It also has the relatively thick-walled gold-plated mica ion chamber and a type II applicator cone with 4 relatively thin scrapers[77]. All measured data were obtained from Carol Wells and Rock Mackie of the University of Wisconsin.

Figs. 3.11, 3.12, 3.13, 3.14 and 3.15 present the spectra and the angular distributions at the phantom surface and comparison between calculated and measured dose distributions along with dose components contributed from each beam defining component from 6, 9, 12, 15 and 18 MeV electron beams respectively.

There is excellent agreement between calculations and measurements. All dose measurements have been done with a small diode detector.

Note that there are two peaks in the electron spectra for all beams. The low energy peak is caused by electrons passing through a 6 mm aluminum applicator scraper and remaining in the beam[77] based on a detailed study which separated the spectra from each scraper component.

Note that the total bremsstrahlung tails in the depth-dose curves are contributed mainly by the contaminant photons in the beams.

In order to explain experimental observations[75] on the influence of an applicator design we also simulated the 9 MeV beam with a newer applicator design (Type III). The energy and angular distributions with this new applicator cone (Type III) from a 9 MeV beam are shown in fig. 3.16 which should be compared to fig. 3.12 for the Type

II applicator. The scattered electrons from the applicator are greatly reduced and the second low-energy peak disappears. The surface dose contributed by scattered electrons from the applicator is reduced from 17% with Type II applicator to 8% with the Type III applicator.

By calculation it was found that the peak could be made to disappear using an additional piece of lead on the top of this part of the applicator. Fig. 3.17 presents the result of a 9 MeV beam with our modified applicator design. The calculated depth-dose curve with the lead in place was virtually identical to that of the new design applicator shown in figs. 3.16 and 3.17 respectively. The calculated dose profiles at $d_{\max}$ for these three different applicators along with dose components are shown in fig. 3.18. After this "discovery", we learned that Varian sells an upgrade kit which has a similar design and performs the same function.

3.3.4 Electron beams from a Philips SL75-20 accelerator

The geometry of the SL75-20 accelerator shown in fig.3.5 can be found in references[31, 32, 33]. It has an exit window, primary collimator, scattering foils (there are three scattering foil positions in a carousel in the treatment head, the appropriate one is set according the beam energy), a beam monitor ion chamber, a steel shielding ring, a mirror, X and Y jaws, an accessory ring and electron applicators which are solid-walled flat tubes made of aluminium. The measured data are from Udale-Smith, reference [31].

Our results for the electron spectrum from the 5 MeV beam are similar to those of earlier study by Udale-Smith[32] although there are some unexplained differences[77].

Figs.3.19, 3.20, 3.21, 3.22 and 3.23 present the energy spectra and angular distributions of electrons and contaminant photons and dose distributions along with dose components contributed from each beam defining component for 5, 10, 14, 17 and 20 MeV electron beams respectively. Good agreement is obtained between cal-

culated and measured central-axis depth-dose curves. Calculated dose profiles at the phantom surface and at $d_{\max}$ (1.1 cm depth) are shown in fig. 3.24 for the 5 MeV beam. It can be seen that even at $d_{\max}$, dose contributions from scattered electrons are significant (33%).

The electron spectra from the SL75-20 have a wide spread partially because of its thicker scattering foils.

At higher beam energies there are two small peaks in the energy spectra. This is due to the geometry of the scattering foils which are stacked disks with increasing radii (see fig. 3.25). Electrons going through the scattering foils near the central-axis lose more energy. A simulation, which uses LATCH to separate components which have passed through various thickness of scattering foils, is shown in fig. 3.25.

Again due to the design of the applicator of the SL75-20 there is a considerable amount of dose contribution from scattered electrons, particularly from the applicator (about 32% to 22% of total surface dose for 5 to 20 MeV beams respectively).

3.3.5 Electron beams from a Siemens KD2 accelerator

The geometry of the Siemens KD2 accelerator shown in fig.3.6 consists of a vacuum-exit-window, primary collimator, scattering foils, a beam monitor ion chamber, a mirror, x and y jaws, and electron applicators which are solid-walled flat tubes made of aluminium with scrapers of brass, steel and aluminium. The experimental data are from Joanna Cygler of Ottawa Regional Cancer Center (6 and 11 MeV) and from Jack Janssen and Henk Huizenga of the Dr Daniel den Hoed Cancer Clinic Netherlands (21 MeV beam). We have also tried to simulate other beams with higher energies from the Ottawa clinic's KD2 but we are unable to reproduce the measured high bremsstrahlung tail. This may be due to our lack of an accurate description of the scattering foils of the accelerator for these higher energies.

Figs. 3.26, 3.27 and 3.28 show the energy spectra and angular distributions of electrons and contaminant photons and dose distributions along with dose components contributed from each beam defining component for 6, 11 and 21 MeV electron beams respectively.

The dose measurements (6 and 11 MeV beam) have been done with a small diode detector. The 21 MeV beam data have been measured with an NACP plane parallel ionization chamber, using the NACP protocol to convert ionization to dose.

Excellent agreement is obtained between calculated and measured central-axis depth-dose curves for these beam energies except for the photon tail of the 21 MeV beam which is underestimated.

Fig. 3.29 presents calculated dose profiles at the phantom surface and at $d_{\max}$ from various components of an 11 MeV beam from the KD2 accelerator.

Due to the applicator design of the KD2 there are more scattered electrons near the field boundary than near the central-axis. The dose profiles contributed by electrons scattered from the applicator become flatter at larger depth because scattered electrons from the applicator have lower energy and relatively large angle.

3.3.6 Electron beams from an AECL Therac 20 accelerator

The Therac 20 accelerator shown in fig.3.7 uses a scanned beam to achieve a flat beam profile. There are no scattering foils in this accelerator. The geometry consists of a vacuum-exit-window, a beam monitor ion chamber, a series of x and y jaws and electron beam applicators. The beam monitor chamber is made with aluminized Kapton and the applicator has thick scraper bars. For the Therac 20 the scan pattern (see chapter 2) was set for a $40 \times 40 \text{ cm}^2$ field but the treatment field was $10 \times 10 \text{ cm}^2$.

The experimental data have been measured with a diode detector by Joanna

Cygler of the Ottawa Regional Cancer Center.

Figs. 3.30, 3.32, 3.33, 3.34 and 3.35 show the energy spectra and angular distributions of electrons and contaminant photons and comparison between calculated and measured central-axis depth-dose curves along with dose components contributed from each beam defining component for 6, 9, 13, 17 and 20 MeV beams respectively.

Due to the combination of no scattering foils and the design of the beam defining system, particularly the design of the applicator in the Therac 20, both the energy and angular spread of the electrons are small.

There are few contaminant photons in the beam. The large angular spread of contaminant photons is due to the fact that most contaminant photons are originating in the beam defining system.

Fig. 3.31 presents a comparison of calculated and measured dose profiles at the depth of $d_{\max}$ and at the phantom surface from various dose components for a 6 MeV beam from the AECL Therac 20. The agreement is excellent.

3.3.7 Electron beams from a Racetrack MM50 accelerator

The geometry of the Scanditronix MM50 (see fig.3.8) at the Memorial Sloan Kettering Cancer Center consists of a vacuum-exit-window, a thin scattering foil, a beam monitor ion chamber, a mirror, a primary collimator and a multileaf collimator which is set to a square for these calculations. The accelerator head is filled with helium gas. The MM50 uses both beam scanning and a very thin scattering foil to give a uniform beam distribution. The beam sizes simulated are larger than for the lower-energy beams from other accelerators to ensure broad-beam conditions. Unlike other accelerators, this machine does not have electron applicators at these beam energies. The beam is shaped by the primary collimator and the multileaf collimator. The scan pattern was set for a circle of 18 cm radius according to the machine specifications.

Figs. 3.36, 3.37 and 3.38 show the energy spectra of 25, 40 and 50 MeV beams incident on the phantom surface and inside the treatment field.

The energy spectra from this accelerator are very narrow and close to monoenergetic because of two factors: the scanned beam and the large air space between last part of the beam defining system and the phantom surface. This air gap means many of the electrons scattered from the beam defining system leave the field.

Note that in the central-axis depth-dose curves the contaminant photons in the beam have a negligible contribution to the bremsstrahlung tails.

Fig. 3.39 presents calculated dose profiles at various depths in a water phantom for the 25 MeV and the 50 MeV beams from the Scanditronix Medical Microtron 50. The beams are very flat across the field at various depths in a water phantom.

3.4 Conclusion

We have presented energy and angular distributions, as well as dose components of clinical electron beams from various medical accelerators with energy ranging from 5 to 50 MeV along with a detailed study of spectra of scattered electrons from the beam defining components. The spectra presented here are based on simulated clinical beams. We have calculated dose distributions using the simulated beams. The calculated and measured dose distributions in water phantoms are in very good agreement in all cases which give us the confidence that the simulations are accurate.

This investigation demonstrates the strength of the full Monte Carlo simulation of accelerator head which allow us to obtain detailed information of clinical electron beams. This information enhances our knowledge on many aspects of clinical electron beams. Extensive beam data from various medical accelerators presented here will be very useful in understanding the characteristics of the clinical electron beams. The information obtained from the simulation is critical to many aspects of optimal

radiotherapy and will help to make more accurate treatment planning and improve electron beam dosimetry. The simulated realistic clinical beams can be used to calculate various dosimetry parameters [39, 73, 80] and to reduce the uncertainty of these parameters which have been calculated using monoenergetic beams because of lack of details of realistic beams.

This report and data sets for all figures are available on the www (word wide web) at <http://www.irs.inms.nrc.ca/inms/irs/papers/PIRS139/pirs139.html>.

Table 3.1: Characteristics of simulated clinical electron beams from 5 clinical electron accelerators. The energies are all number averaged values for particles inside the beam and above 189 keV kinetic energy for e^- and 10 keV for photons. E_{max} is the maximum kinetic energy for particles anywhere in the scoring plane (at the surface of the phantom here).

Linac	beam energy MeV	R_{50} cm	R_p cm	$E_{o,p}$ MeV	% surface dose		$\bar{E}$		E_{max}	
					total $e^- + \gamma$	direct e^-	MeV e^-	MeV γ	MeV e^-	MeV γ
Clinac 2100C	6	2.63	3.26	6.88	82	61	6.11	1.36	7.10	7.94
	9	4.00	5.03	10.39	85	60	9.10	1.85	10.59	11.66
	12	5.18	6.30	13.05	89	70	11.74	2.19	13.23	14.33
	15	6.50	7.92	16.31	92	70	14.59	2.52	16.62	17.52
	18	7.72	9.45	19.49	93	70	17.35	2.81	19.67	20.73
SL75 -20	5	2.08	2.60	5.66	78	47	5.07	1.23	5.90	6.68
	10	4.12	5.20	10.66	88	55	9.34	1.96	10.98	11.98
	14	5.98	7.40	15.34	93	57	13.28	2.52	15.82	16.79
	17	6.96	8.70	17.79	93	61	15.28	2.98	18.67	19.68
	20	8.10	10.0	21.04	94	61	17.76	3.35	21.96	22.96
KD2	6	2.31	2.84	6.05	73	62	5.53	1.30	6.17	6.99
	11	4.21	5.12	10.51	81	70	9.70	1.88	10.63	11.45
	21	8.30	10.1	20.59	89	79	18.74	3.11	20.78	21.87
Therac 20	6	2.18	2.68	5.60	68	65	5.50	1.00	5.62	5.86
	9	3.42	4.12	8.42	74	70	8.25	1.30	8.43	8.65
	13	5.14	6.15	12.41	80	76	12.14	1.66	12.42	12.53
	17	6.85	8.10	16.35	88	82	15.93	1.97	16.35	16.57
	20	8.10	9.63	19.30	90	85	18.74	2.17	19.31	19.47
MM50	25	10.36	12.30	24.77	92	92	24.53	3.50	24.78	24.81
	40	15.47	19.50	39.50	94	94	38.90	4.91	39.50	39.60
	50	18.55	24.00	49.69	95	95	48.90	5.59	49.69	49.73

Table 3.2: Simulation data of clinical electron beams from five medical accelerators. The field size is 10×10 cm² except beams from MM50 where field sizes are 25×25 (25 and 40 MeV beams) and 20×20 (50 MeV beam). The CPU time is for an R4400 CPU in an SGI Indigo.

Machine	beam nominal energy	E_m incident MeV	e^- in field / 10^2 inc e^-	γ in field / 10^2 inc e^-	% direct e^-	% direct γ	CPU /inc e^- ms
Clinac 2100C	6	8.00	2.15	2.10	78.9	69.6	21.1
	9	11.70	2.99	4.15	74.8	67.4	44.8
	12	14.35	3.83	5.83	82.9	70.9	56.3
	15	17.62	5.59	9.08	81.5	67.2	45.7
	18	20.80	7.48	12.7	80.8	63.7	71.2
SL75-20	5	6.70	2.06	1.21	65.2	71.4	9.78
	10	12.00	4.14	4.05	66.8	75.6	13.2
	14	16.80	11.8	13.4	66.4	72.1	29.4
	17	19.70	5.59	10.1	70.0	80.1	18.8
	20	23.00	6.95	13.6	69.5	78.5	54.1
KD2	6	7.00	0.70	1.10	84.5	79.9	9.07
	11	11.55	2.72	3.18	87.2	77.0	11.1
	21	21.88	5.59	11.7	90.0	70.7	18.0
Therac 20	6	5.88	3.68	0.52	96.1	25.1	62.2
	9	8.69	5.60	1.35	95.9	16.3	92.0
	13	12.69	6.09	2.58	95.6	10.5	125.
	17	16.63	6.25	3.93	95.3	7.42	165.
	20	19.59	6.33	5.07	94.8	5.98	197.
MM50	25	24.90	84.9	4.97	99.6	50.9	46.7
	40	39.65	82.1	11.4	99.4	58.6	82.2
	50	49.86	51.0	10.6	99.1	42.0	155.

Figure 3.1: A 20 MeV electron beam from the NRC research accelerator as shown by EGS windows with about 100 histories. See section 3.3.2 for a detailed description.

Figure 3.2: A 9 MeV electron beam from the Varian Clinac 2100C accelerator with type II applicator as shown by EGS₂ windows with about 100 histories. See section 3.3.3 for a detailed description.

Figure 3.3: A 9 MeV electron beam from the Varian Clinac 2100C accelerator with type III applicator as shown by EGS_{windows} with about 100 histories. See section 3.3.3 for a detailed description.

Figure 3.4: A 9 MeV electron beam from Varian Clinac 2100C* accelerator with our modified applicator as shown by EGS_windows with about 100 histories. See section 3.3.3 for a detailed description.

Figure 3.5: A 20 MeV electron beam from the Phillips SL75-20 accelerator as shown by FGS windows with about 100 histories. See section 3.3.4 for a detailed description.

Figure 3.6: An 11 MeV electron beam from the Siemens KD2 accelerator as shown by EGS windows with about 100 histories. See section 3.3.5 for a detailed description.

Figure 3.7: A 20 MeV electron beam from the AECL Therac 20 accelerator as shown by EGS windows with about 100 histories. See section 3.3.6 for a detailed description.

Figure 3.8: A 50 MeV electron beam from the Scanditronix MM50 accelerator as shown by EGS₄ windows with about 100 histories. See section 3.3.7 for a detailed description.

Figure 3.9: The energy spectra (upper left) and angular distributions (upper right) at the phantom surface and inside the $8 \times 8 \text{ cm}^2$ field of a 10 MeV electron beam from NRC' accelerator. The ratio of the number of electrons to that of contaminant photons inside the field is $1 : 0.58$. The normalization values are 4.52×10^5 and 2.39×10^5 for spectra and angular distributions respectively. The lower figure is a comparison between calculated and measured central-axis depth-dose curves.

Figure 3.10: The energy spectra (upper left) and angular distributions (upper right) at the phantom surface and inside the $8 \times 8 \text{ cm}^2$ field of a 20 MeV electron beam from the NRC research accelerator. The energy used in the simulation is 20.3 MeV. Starting 10^6 electron histories incident on the vacuum-exit-window there are 8.6×10^5 electrons and 4.3×10^5 photons inside field at the phantom surface. The ratio of the number of electrons to that of contaminant photons inside the field is 1 : 0.50. The normalization values are 4.11×10^5 and 2.78×10^5 for spectra and angular distributions respectively. The lower figures show the calculated and measured central-axis depth-dose curves (lower left) and dose profiles (lower right) at different depths in the 20 MeV electron beam.

Figure 3.11: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 6 MeV electron beam from the Varian Clinac 2100C at the University of Wisconsin. The normalization values are 1.01×10^5 and 7.96×10^4 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves along with dose contributions from various components.

Figure 3.12: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 9 MeV electron beam from the Varian Clinac 2100C at the University of Wisconsin. The normalization values are 7.49×10^4 and 1.02×10^5 for the spectra and the angular distributions respectively. The lower figures are a comparison between measured and calculated depth-dose curves and the dose profiles at the surface (0.1 cm depth) from various components.

Figure 3.13: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 12 MeV electron beam from the Varian Clinac 2100C at the University of Wisconsin (Wisconsin Comprehensive Cancer Center in Madison). The normalization values are 6.28×10^4 and 9.58×10^4 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves along with dose contributions from various components.

Figure 3.14: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 15 MeV electron beam from the Varian Clinac 2100C at the University of Wisconsin. The normalization values are 6.50×10^4 and 1.28×10^4 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves along with dose contributions from various components.

Figure 3.15: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 18 MeV electron beam from the Varian Clinac 2100C at the University of Wisconsin. The normalization values are 8.06×10^4 and 1.83×10^5 for the spectra and the angular distributions respectively. The lower figures are a comparison between measured and calculated depth-dose curves and dose profiles at d_{max} (1.9 cm depth) from various components.

Figure 3.16. The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 9 MeV electron beam from the Clinac 2100C with a newer applicator (Type III applicator). The normalization values are 7.56×10^4 and 1.01×10^5 for the spectra and the angular distributions respectively. The lower figures are the calculated central-axis depth-dose curves (left) and the dose profiles (right) at the surface (0.1 cm depth) from various components.

Figure 3.17: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 9 MeV electron beam from the Clinac 2100C with our modified applicator. The normalization values are 5.45×10^4 and 1.02×10^5 for the spectra and the angular distributions respectively. The lower figures are the calculated central-axis depth-dose curves (left) and dose profiles (right) at the surface (0.1 cm depth) from various components.

Figure 3.18: The calculated dose profile at $d_{\max}$ (2.3 cm depth) in a 9 MeV electron beam from the Varian Clinac 2100C with a Type III applicator (upper left). The dose profiles at $d_{\max}$ (2.3 cm depth) from various components in a 9 MeV electron beam from the Clinac 2100C with our modified applicator (upper right). The dose profiles at $d_{\max}$ (1.9 cm depth) from various components in a 9 MeV electron beam from the Varian Clinac 2100C with a Type II applicator (lower). It can be seen that our simply modified applicator has virtually identical dose profiles at $d_{\max}$ to that of the new design applicator (Type III), even for dose components.

Figure 3.19: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 5 MeV electron beam from the Philips SL75-20 at Leeds University[31]. The normalization values are 2.47×10^5 and 1.42×10^5 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves along with dose contributions from various beam defining components.

Figure 3.20: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 10 MeV electron beam from the Philips SL75-20 at Leeds University[31]. The normalization values are 1.22×10^5 and 1.94×10^5 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves along with dose contributions from various beam defining components.

Figure 3.21: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 14 MeV electron beam from the Philips SL75-20 at Leeds University[31]. The normalization values are 1.58×10^5 and 3.24×10^5 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured[31] and calculated depth-dose curves along with dose contributions from various beam defining components.

Figure 3.22: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 17 MeV electron beam from the Philips SL75-20 at Leeds University[31]. The normalization values are 3.83×10^4 and 1.88×10^5 for the spectra and the angular distributions respectively. The lower figure shows calculated depth-dose curves along with dose contributions from various beam defining components.

Figure 3.23: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 20 MeV electron beam from the Philips SL75-20 at Leeds University. The normalization values are 3.48×10^4 and 1.48×10^5 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves along with dose contributions from various beam defining components.

Figure 3.24: The calculated dose profiles at the surface (0.1 cm depth) (upper) and at $d_{\max}$ (1.1 cm depth) (lower) of a 5 MeV electron beam from the Philips SL75-20 at Leeds University for various dose components.

SL75-20, geometry of scattering foils for 20 MeV beam

Philips SL75-20, 20 MeV electron beam

Figure 3.25: Geometry of scattering foils (upper) for the 20 MeV beam of the SL75-20, where regions 1 to 5 are thin metal disks. Region 1 has a radius of 0.725 cm and a thickness of 0.0127 cm. The distance between the scattering foils and the vacuum-exit-window is about 15 cm. Electron spectra at the phantom surface are shown in the lower panel. Those electrons which have been in region 1 lose more energy because they have to pass through the extra thickness of region 1.

Figure 3.26: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 6 MeV electron beam from Siemens KD2 at the General Hospital in Ottawa. The normalization values are 5.78×10^4 and 5.37×10^4 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves (lower left) and dose profiles along with various dose components.

Figure 3.27: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of an 11 MeV electron beam from Siemens KD2 at the General Hospital in Ottawa. The normalization values are 7.60×10^4 and 6.88×10^4 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves along with various dose components.

Figure 3.28: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 21 MeV electron beam from a Siemens KD2 in the Netherlands. The normalization values are 5.69×10^4 and 1.17×10^5 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated central axis depth-dose curves from various dose components.

Figure 3.29: Calculated dose profiles at the phantom surface (upper) and at the depth of $d_{\max}$ (lower) from various dose components for the 11 MeV electron beam from the Siemens KD2 at the General Hospital in Ottawa.

Figure 3.30: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 6 MeV electron beam from the AECL Therac 20. The normalization values are 9.63×10^5 and 3.02×10^5 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves along with dose components contributed from each beam defining component.

Figure 3.31: Calculated and measured dose profiles at $d_{\max}$ (upper) and at the phantom surface (lower) from various dose components in a 6 MeV beam from the AECL Therac 20.

Figure 3.32: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 9 MeV electron beam from the AECL Therac 20. The normalization values are 8.95×10^5 and 3.42×10^5 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves along with dose components contributed from each beam defining component.

Figure 3.33: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 13 MeV electron beam from the AECL Therac 20. The normalization values are 4.71×10^5 and 2.61×10^5 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves along with dose components contributed from each beam defining component.

Figure 3.34: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the 10×10 cm² field of a 17 MeV electron beam from the AECL Therac 20. The normalization values are 4.02×10^5 and 1.74×10^5 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth dose curves along with dose components contributed from each beam defining component.

Figure 3.35: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $10 \times 10 \text{ cm}^2$ field of a 20 MeV electron beam from the AECL Therac 20. The normalization values are 2.76×10^5 and 2.27×10^5 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves along with dose components contributed from each beam defining component.

Figure 3.36: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $25 \times 25 \text{ cm}^2$ field of a 25 MeV electron beam from the Scanditronix Medical Microtron 50 at Memorial Sloan-Kettering Cancer Center. The normalization values are 2.26×10^5 and 4.68×10^5 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves along with dose components contributed from each beam defining component.

Figure 3.37: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $25 \times 25 \text{ cm}^2$ field of a 40 MeV electron beam from the Scanditronix Medical Microtron 50 at Memorial Sloan-Kettering Cancer Center. The normalization values are 1.23×10^5 and 4.62×10^5 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves along with dose components contributed from direct electrons and contaminant photons.

Figure 3.38: The energy spectra (upper left) and angular distributions (upper right) from various components at the phantom surface and inside the $20 \times 20 \text{ cm}^2$ field of a 50 MeV electron beam from the Scanditronix Medical Microtron 50 at Memorial Sloan-Kettering Cancer Center. The normalization values are 1.39×10^5 and 7.68×10^4 for the spectra and the angular distributions respectively. The lower figure is a comparison between measured and calculated depth-dose curves along with dose components contributed from direct electrons and contaminant photons.

Figure 3.39: Calculated dose profiles at various depth in a water phantom for a 25 MeV (upper) and a 50 MeV (lower) beam from the Scanditronix Medical Microtron 50 at Memorial Sloan-Kettering Cancer Center.

Chapter 4

**Average energy, energy-range
relationships and depth-scaling
factors of clinical electron beams**

4.1 Introduction

The dose distributions produced by an electron beam depend strongly upon the beam's energy distribution and other aspects of the beam's quality. Clinical electron beams can be characterized by a number of parameters. The mean energy and the most probable energy of electrons at a phantom surface are two important parameters for electron beams as they are used for the choice of energy dependent dosimetric parameters, such as the stopping-power ratios and the electron fluence correction factors.

4.1.1 Most probable and mean energies on the surface

The most probable energy of the electron beam at the surface of the phantom, $E_{o,p}$, is related to the practical range in water, R_p (in cm), by the relationship[81, 11, 82, 13]

$$E_{o,p} = 0.22 + 1.98R_p + 0.0025R_p^2 \quad (\text{MeV}) \quad (4.1)$$

Eqn 4.1 was first obtained by fitting the experimental results for nearly monoenergetic 10 to 50 MeV electron beams from a Racetrack microtron[83]. This is a modification of the linear Markus energy-range formula ($E_{o,p} = 1.95R_p + 0.48$) [84]. From 1 to 50 MeV eqn 4.1 agrees within 1% with practical ranges extracted from Berger and Seltzer's Monte Carlo calculated depth-dose curves between 1 and 60 MeV[82, 85]. It has also been found[81, 82] that eqn 4.1 reproduces the measured data with an accuracy of 2% from 5 to 50 MeV. The Monte Carlo calculations[86] by Rogers and Bielajew have generated similar values except at high energies where EGS4 calculations predict energies which are 4 to 6% less than predicted by the ICRU/ETRAN results[86]. Sorcini and Brahme have used a general analytical energy-range relation to relate R_p to $E_{o,p}$ of incident electron beams in the range 1 to 50 MeV for absorbers of any atomic number[87]. However, for water their results are very similar to those in eqn 4.1.

Several dosimetry protocols[81, 11, 13, 12] recommend the following relationship for the determination of the mean energy of the electron beam at the surface of the phantom:

$$\bar{E}_0 = 2.33R_{50} \quad (\text{MeV}) \quad (1.2)$$

where $\bar{E}_0$ is the mean energy and R_{50} is the depth in cm of either the 50% ionization or dose levels. The value of 2.33 MeV/cm was recommended by the AAPM TG-21 protocol[11] and was obtained from an analysis of depth-dose curves calculated by Berger and Seltzer[88] for plane-parallel, infinitely wide beams of monoenergetic electrons incident upon a semi-infinite water phantom. This value was an average for values in the energy range 5 – 50 MeV and, as pointed out by Wu *et al.*[89], is applicable to the 50% value of the depth-dose curve although TG-21 applied it to depth-ionization curves. The same value of 2.33 MeV/cm was recommended by Brahme and Svensson[90] based on measurements for depth-dose curves corrected to infinite SSD although they also explicitly state that their results are in agreement with the Monte Carlo calculations of Berger and Seltzer. After isolating an error in the earlier Monte Carlo calculations with ETRAN, Rogers and Bielajew[86] used the EGS4 Monte Carlo system to calculate and to tabulate $\bar{E}_0$ as a function of R_{50} for various values of SSD. Their results systematically give higher $\bar{E}_0$ values and are recommended in the AAPM TG-25 Report[12] for more accurate estimates of the mean energy of electron beams.

The above relationships are based on depth-dose curves. The IAEA[13] and the NACP[81] protocols tabulate values which use the half-value depth measured from either absorbed dose or ionization curves at SSD = 100 cm and we have recently presented explicit data to facilitate using either the depth of 50% dose or ionization (see chapter 5)[39].

In the previous studies mentioned above, either parallel, monoenergetic beams were used in Monte Carlo calculations or the “measurements” used approximate methods to estimate the energy of the beam. With the development of a Monte

Carlo code to simulate realistic electron beams (described below), we can investigate the accuracy of these previous approximations. Udale-Smith has done a similar study previously [32] for 3 specific machines and found considerable disagreement with various energy-range relationships. However, as reported previously, there are some unexplained differences between her calculated spectra and ours, with mean energies on the surface differing by up to 300 keV for the same 10 MeV beam (see chapter 2).

4.1.2 Mean electron energy as a function of depth in a phantom

The mean electron energy as a function of depth in the phantom is another parameter which is needed for the choice of the replacement correction for absorbed dose measurements with an ion chamber. Harder's relation is employed in the AAPM protocol[11, 12] to estimate the mean electron energy at a depth, d

$$\bar{E}_d = \bar{E}_o(1 - d/R_p), \quad (4.3)$$

where $\bar{E}_o$ is the average incident electrons energy at the phantom surface, d is the depth of measurement and R_p is the practical range in the same units.

Instead of Harder's relationship, the IAEA recommends using the Monte Carlo results of Andreo and Brahme[91] for the evaluation of $\bar{E}_d$ given $\bar{E}_o$ and d/R_p . These calculations are for monoenergetic broad parallel incident beams. The mean electron energy is calculated for the primary electrons and the secondary electrons set-in-motion in the phantom are not included. Since there are secondary electrons set-in-motion in the phantom, it is interesting to see how this will affect the mean electron energy in phantoms.

As well as changes from considering secondary electrons in the phantom, clinical beams have energy and angular distributions which introduce uncertainty even if the

mean energy of the clinical beams is estimated correctly and used with calculations for monoenergetic beams. Udale-Smith has calculated $\bar{E}_d$ values for all electrons and compared calculations with monoenergetic incident beams to those using simulated clinical beams from SL75 machines for beam energies less than 12 MeV[32]. Her results showed that the value of $\bar{E}_d$ for the monoenergetic beams decreased with depth more quickly than that of the simulated clinical beam.

4.1.3 Depth-scaling for plastic phantoms

It is recommended by major dosimetry protocols[11, 12, 13] that water be used as the standard phantom material for the dosimetry of high-energy electrons. However it is not always possible or practical to perform dosimetry measurements in a water phantom. In this case it is necessary to scale depths measured in nonwater phantoms to water equivalent depths. One form of depth-scaling is range scaling whereby the range in one material is used to deduce the corresponding range in water. Using Monte Carlo methods, Grosswendt and Roos[92] calculated depth-dose curves in different phantom materials for monoenergetic broad electron beams with incident energies between 1 and 10 MeV. From these the corresponding values of R_{50} and R_p were deduced and compared to the predictions for a variety of protocols. Their results showed that deviations in the range scaling from different recommendations and in the energy-range relations for water may accumulate to give errors in assigned energies of several per cent.

As an example of this type of range scaling, the IAEA recommends scaling R_p and R_{50} in cm to water equivalent ranges using[13]:

$$\frac{R_{plastic}}{R_{water}} = \frac{(r_o/\rho)_{plastic}}{(r_o/\rho)_{water}} \quad (4.4)$$

where r_o/ρ is the linear continuous-slowng-down range in cm tabulated as a function of energy by ICRU[82]. The mean energy at the phantom surface is used as an input parameter for the choice of the ratios.

Depth scaling can also be done for other purposes. In the AAPM TG-21 and TG-25 protocols, the objective is to find depths at which the electron spectra are identical so that one can predict absolute dose in water given an ion chamber reading in plastic at the scaled depth. The spectra are assumed identical if the mean energies are the same. The AAPM TG-25 protocol[12] recommends that the water-equivalent depth be determined by scaling the depth in plastic by the ratio of R_{50} values *i.e.*:

$$d_{water} = d_{med} \times \rho_{eff} = d_{med} \left(\frac{R_{50}^{water}}{R_{50}^{med}} \right) \quad (4.5)$$

where ρ_{eff} is the effective density for nonwater phantoms. The recommended values of ρ_{eff} in the AAPM TG-25 protocol[12] are based exclusively on R_{50} measurements of an 18 MeV beam. The IAEA Code of Practice does depth-scaling in the same manner as described above for range-scaling.

4.1.4 The BEAM code

BEAM is a general-purpose Monte Carlo simulation code (see chapter 2) which is based on the EGS4 Monte Carlo system[15]. BEAM allows the simulation of radio-therapy treatment units and produces data on realistic clinical beams. It has been extensively verified against measured dose distributions for a variety of accelerators and excellent agreement was obtained in central-axis percentage depth-dose as well as in dose-profile curves[77, 93]. The excellent agreement indicates that the simulation of clinical beams is accurate. However, the energy of the assumed mono-energetic beam of electrons incident on the accelerator vacuum window is a parameter which is adjusted to match the experimental value of R_{50} . In one case the incident electron energy was known independently with an accuracy of $\pm 1\%$ and the calculated values of R_{50} agreed with the experimental values within 1.5%. This suggests an established overall accuracy on the energies of only 1 to 2%. However, a major thrust of this work is to study the effects of using incident realistic *vs* mono-energetic beams so that even with an overall error of 2%, a difference of 0.2% is likely calculated accurately.

4.1.5 Outline

Using the simulated realistic clinical beams we calculate various beam parameters: $\bar{E}_0$, $E_{0,p}$, R_{50} and R_p . The two empirical formulas (eqn 4.1 and 4.2) commonly used in electron beam dosimetry can be tested against simulated realistic clinical beams, in addition, $\bar{E}_d$ as function of depth can be calculated using realistic clinical beams.

As mentioned above, the purpose of depth-scaling is to scale the depth measured in a nonwater phantom to the water-equivalent depth in water where the electron energy spectra are identical. Because energy-loss straggling and multiple scattering depend upon the effective atomic number of the phantoms, it is difficult to find corresponding depths where the energy spectra are identical. However, there are corresponding depths where the mean energies are identical. The TG-25 protocol assumes the depth-scaling factor defined in eqn 4.5 holds for these depths[12]. By calculating the mean electron energy $\bar{E}_d$ as a function of depth in different phantom materials we are able to find the scaling factor directly and establish whether it is constant with depth.

In the following we first briefly describe the simulated clinical beams and the calculation codes used in this study. The beam parameters of $\bar{E}_0$, $E_{0,p}$, R_{50} and R_p calculated using various simulated clinical beams are presented and their relationships are studied. We assess the accuracy of the energy-range relationships recommended in protocols when they are applied in realistic clinical electron beams.

We will present the number averaged and fluence averaged electron energy at the surface and at depth in a phantom as a function of off-axis distance.

We calculate the depth-scaling factors directly by scaling the depth in nonwater phantom to the water-equivalent depth where the mean energies are the same for various beams and materials and compare our calculated values with those recommended in protocols.

4.2 Method of Calculations

4.2.1 The incident clinical beams and phantom calculations

The simulation of various accelerators and the calculation of dose distributions in a phantom are described in detail in chapter 2. The position, energy and angle of every particle emerging from the accelerator are stored in a "phase-space" data file. The energy cutoffs for particle transport are set to $ECUT = 0.700$ MeV (total energy), $AE = 0.521$ MeV and $PCUT = AP = 0.010$ MeV. In all cases, the PRESTA electron algorithm[53] is employed with $ESTEPE = 0.01$, although this value is conservative and the default value is adequate in the BEAM code (see chapter 2).

The simulated clinical electron beams have been described in detail in chapter 3. The dose distributions are calculated by using the EGS4 user code DOSXYZ. The comparison between calculated and measured dose distributions can be found in references [77, 93].

The calculation of depth-dose curve in plastic phantoms was done with materials as defined in ICRU Report 37 except for white (or high impact) polystyrene which is taken as having a density of 1.48g/cm^3 and a TiO_2 content of 3% by weight[12].

4.2.2 Analysis of phase-space files

Two computer codes, PHSP_ENXY and PHSP_SPEEXY, are used to analyze the phase-space files from BEAM to obtain the mean energy and energy spectra of simulated beams at the surface of the phantom. The codes allow the user to select particles according to the particle's history by making use of the LATCH feature of BEAM (see chapter 2).

PHSP_ENXY calculates both the number averaged mean energy, $\bar{E}_o$, and the fluence averaged mean energy, $\bar{E}_{o,f}$, in square strips on the phantom surface. The

number averaged energy $\bar{E}_o$ and fluence averaged energy $\bar{E}_{o,f}$ in some region A are calculated as:

$$\bar{E}_o = \frac{\sum_{i=1}^n E_i}{n} \quad (1.6)$$

$$\bar{E}_{o,f} = \frac{\sum_{i=1}^n E_i (\cos\theta_i)^{-1}}{\sum_{i=1}^n (\cos\theta_i)^{-1}} \quad (1.7)$$

where E_i and θ_i are the energy and the angle (relative to z-axis) of all particles in region A, and θ is capped at 89° to prevent distortion (see reference [14]). The fluence averaged mean energy, $\bar{E}_{o,f}$, is the quantity which corresponds to the mean energy scored in a phantom (see next section). The number averaged mean energy, $\bar{E}_o$, corresponds to that measured by various instruments.

PHSP_SPECCY calculates the energy spectra in regions specified by the user at the surface of the phantom. The energy bin size of the spectrum is set by the user.

The calculated energy spectra are used to obtain the $E_{o,p}$ of clinical incident beams. The detailed spectra and angular distributions of all beams analyzed in this paper are presented in an internal report[93] which is available on the www (world-wide web) at <http://www.irs.inms.nrc.ca/inms/irs/papers/PIRS439/pirs439.html>.

4.2.3 The EGS4 user-code ENXYZ

The "phase-space" data files are used as input (incident beams) to the EGS4 user-code ENXYZ to calculate the spatial variation of the electron mean energy in the phantom. In using the code ENXYZ, Cartesian coordinates are employed and the phantom is divided into many small rectangular volume elements at the user's choice and the mean energy is calculated in each volume of interest. ENXYZ uses the techniques used in FLURZ[94] to calculate the mean energy in phantoms. The mean energy of an electron spectrum at a position x, y, z is defined by

$$\bar{E}(x, y, z, \Delta) = \frac{\int_{\Delta}^{E_{max}} E(x, y, z) \Phi_1(x, y, z) dE}{\int_{\Delta}^{E_{max}} \Phi_1(x, y, z) dE} \quad (4.8)$$

where $\Phi_1(x, y, z)$ is the fluence spectrum of charged particles in the region of interest (i.e. $\Phi_1(x, y, z)dE$ is the fluence of charged particles with energies between E and $E + dE$), and Δ is the low energy cut-off (=ECUT) below which all electron histories are to be terminated. The value of Δ is set by the user. For each volume element the ENXYZ code calculates the sum of the curved path-length for each step times the mid-step energy of electrons divided by the summation of path-length of electrons in the region of interest[14]. This technique avoids any inaccuracies associated with using a binned representation of the fluence spectrum in eqn 4.8. The code allows the user to exclude the effects of secondary charged particles and consider only the fluence of primary electrons. The electron histories have been calculated down to an energy Δ . The code also allows the user to make use of the LATCH feature of BEAM (see chapter 2) to study the effects of scattered electrons from the beam defining system and contaminant photons when a phase-space file from BEAM is used as an incident beam.

The stopping powers used in the calculations are the ICRU Report 37 values[95] as implemented in EGS4[96]. The PRESTA electron algorithm[53] is employed in all the Monte Carlo calculations. The calculation speed is similar to that of DOSXYZ to calculate dose distributions (a few hours for a typical calculation for a beam with a $10 \times 10 \text{ cm}^2$ field size and with good statistics $\pm 1\%$).

4.3 Results of Monte Carlo calculations

4.3.1 Mean energy at the surface of the phantom and energy-range relationships

The mean energy at the surface of the phantom is not precisely defined since there is a difference between the number averaged mean energy (eqn 4.6) and the fluence averaged mean energy (eqn 4.7). In addition, the mean energy varies with off-axis

positions.

Figs. 4.1, 4.2 and 4.3 show the calculated electron mean energy above 700 keV as function of off-axis position (in square rings) at the phantom surface for an 18 MeV beam from a Clinac 2100C, a 10 MeV beam from an SL75-20 and a 6 MeV beam from a Therac 20 respectively.

Fluence averaged mean energy is about 2% lower than the number averaged mean energy at the surface of the phantom due to the fact that the lower-energy electrons in the incident beam have large angles on average.

Note that in fig. 4.1 the average energy of all electrons near the field boundary is about 0.4 MeV higher than that near the central-axis while for direct electrons (those that do not hit any jaws, collimators or applicators) the average energy near the field boundary is only about 0.1 MeV higher than that near the central-axis.

Unlike the beams from the Clinac 2100C and the SL75-20 of figs. 4.1 and 4.2, the mean energies decrease slowly beyond beam field for a 6 MeV beam from the Therac 20 as shown in fig.4.3. This is due to the design of the applicator of the Therac 20 and the large air gap (≈ 10 cm) between the applicator and the surface of the phantom. The decrease is more rapid for higher energy beams from the Therac 20 since there is less scattering. It should also be noted that the field edge is sharply defined and this slow drop in energy for the 6-MeV beam is for a very small number of electrons.

Table 4.1 lists the characteristics of the simulated beams and calculated $\bar{E}_n$ along with values determined according to protocols. Values of $\bar{E}_n$ presented in table 4.1 are the number averaged mean energy at the surface of the phantom inside the beam field. The calculated $\bar{E}_o$ is the average energy of all electrons inside the treatment field and $\bar{E}'_o$ is the average energy of only direct electrons inside treatment field.

Incident electrons at a phantom surface in a realistic clinical beam contain scattered low-energy electrons from beam defining components and these scattered electrons decrease the average electron energy at the phantom surface but they have very

little influence on the value of R_{50} . Depending on the design of an accelerator, the number of scattered electrons varies considerably (see chapter 3) and it is very difficult to obtain a relationship between R_{50} and $\bar{E}_0$ for various beams from different accelerators.

Fig. 4.4 shows the energy-range relationship for electron beams from different accelerators. The scatter in the relationship between $\bar{E}_0$ and R_{50} improves considerably when the scattered low-energy electrons are excluded from the calculation of average electron energy (filled symbols). The relationship obtained using direct electrons is close to the result of monoenergetic beam calculations by Rogers and Bielajew[86, 12] except at high energies.

Although the electrons from the high-energy scanned beams are almost monoenergetic, there are discrepancies between the results of this work and those of Rogers and Bielajew[86]. We have found that the discrepancies relate to the widely used inverse-square law correction to convert central-axis depth-dose curves for parallel incident beams to the central-axis depth-dose curves for phantoms at a finite SSD from a point source. Fig. 4.5 shows that there is considerable difference between a 50 MeV depth-dose curve at SSD = 110 cm calculated for an idealized point source in vacuum vs the depth-dose curve calculated for an incident parallel beam and corrected using $((SSD + d)/SSD)^2$. It can be seen that the error in using the $1/r^2$ corrected parallel beam results for R_{50} or R_p in point source beams is about 3 mm. The results in the upper right corner of fig. 4.4 show that the difference between the results of Rogers and Bielajew[86] and the current results can be explained by this effect. This effect is only important for high energies since the $1/r^2$ corrections are much less at shallow depths.

Since the most-probable electron energy at the phantom surface is not affected by the electron beam defining system it is not surprising that the relationship is independent of beam quality. Fig. 4.6 shows the relationship between most-probable energy $E_{0,p}$ and practical range R_p obtained from realistic clinical beam simulations

and using various formulae [82, 86, 87]. Despite wide differences in the energy spectra of clinical beams (see chapter 3), there is little variation in $E_{0,p}$ as a function of R_p for beams from various accelerators. Fig. 4.6 shows that various formulas [82, 86, 87] reproduce the simulated beam data with an accuracy of 5% from 5–50 MeV. For the 50 MeV beam from the MM50 accelerator, $E_{0,p}$ is underestimated by 2% and 0.7% using the ICRU formula and the analytical energy-range relationship by Sorcini and Brahme [87] respectively. The overestimation (1%) at this high energy by the relationship of Rogers and Bielajew [86] is again due to the inverse-square law correction approximation as discussed above. Table 4.2 presents comparisons between values of R_p and $E_{0,p}$ obtained from the Monte Carlo simulations (using 10 keV energy bins) and those obtained from experiments and from the ICRU formula (eqn 4.1) using the Monte Carlo calculated values of R_p . R_p is obtained from the depth-dose curves as the depth of intersection of a line drawn tangent to the depth-dose curve at the point of maximum slope and a line extrapolated from the bremsstrahlung tail. The uncertainty in values of R_p is about 1% due to the procedure for obtaining them.

The close agreement between the calculated and measured values of R_p shown in Table 4.2 shows that by adjusting the energy from the accelerator vacuum to match the experimental values of R_{50} , we are also accurately matching R_p values. The agreement between the most probable energies calculated with the standard (ICRU) formula based on the R_p value, is quite satisfactory. Nonetheless, for scattering foil accelerators there is a clear trend for the Monte Carlo value to be typically 200 to 300 keV higher.

4.3.2 Relationship between incident energy E_{in} and $\bar{E}_e$

The relationship between incident electron energy at the vacuum-exit-window of an electron beam, E_{in} , and the average electron energy at the phantom surface, $\bar{E}_e$, is influenced by the thickness of scattering foils and the design of the beam defining

system of an accelerator. Fig. 4.7 shows $E_{in} - \bar{E}_o$ and $E_{in} - \bar{E}'_o$ vs R_{50} for various clinical beams, where $\bar{E}'_o$ is the average energy of direct electrons at the phantom surface. These are not influenced by the design of beam defining system. It can be seen that the difference between E_{in} and $\bar{E}_o$ or $\bar{E}'_o$ is small (<1 MeV) for beams from the Therac 20 (which are scanned beams with no scattering foils) and from the MM50 accelerator (which produces scanned beams and has a very thin scattering foil).

ICRU Report 35[82] gives an approximate formula to relate $\bar{E}_o$ to E_{in} :

$$\bar{E}_o = E_{in} - \Sigma(\Delta E)_{tot,n} \quad (4.9)$$

where $(\Delta E)_{tot,n}$ is the mean total energy loss in layer n of material being traversed, i.e., the sum of collision and radiative energy losses. Since each side in eqn 4.9 can be calculated using the simulated clinical beams we can assess the accuracy of this formula. This is an idealized case because both calculations are known to use identical descriptions of the scattering foils.

Fig. 4.8 shows the discrepancies between $E_{in} - \bar{E}_o$ and calculated $\Sigma(\Delta E)_{tot,n}$ for various beams from different medical accelerators (filled symbols). The calculated energy loss in the intervening material layers is given by $\Sigma(\Delta E)_{tot,n} = \Sigma dt_n (dE/dt)_n$. It can be seen (fig.4.8) that the discrepancies are small (<0.4 MeV) for a variety of clinical beams although they range between 600 keV high and 700 keV low. Fig. 4.8 also shows the discrepancies between $E_{in} - \bar{E}'_o$ and calculated $\Sigma(\Delta E)_{tot,n}$ (open symbols). One might expect that these discrepancies should be smaller, however they are larger for beams from accelerators with scattering foils. This is because the simple model of eqn 4.9 assumes that all electrons reach the patient plane. In reality, and in the Monte Carlo calculations, those electrons which undergo large energy loss events are more likely lost from the beam and hence the electrons left in the beam have a higher average energy. By including the electrons scattered from the beam defining system, the model works better. That this model works at all is because secondary electrons in the beam play little role in the average energy. We have studied the effect of secondaries on the average energy. The results show that for the 6 and 18

MeV electron beams from the Clinac 2100C only 0.8% and 1.7% of the total electrons inside the beam field are secondary electrons with average energies of 1.45 and 3.11 MeV respectively. By excluding these secondaries, the average energy would increase by 0.65% (0.04 MeV) and 1.7% (0.25 MeV) for the 6 and 18 MeV beams from the Clinac 2100C respectively.

4.3.3 Mean energy as a function of depth for monoenergetic incident electron beams

For incident monoenergetic electron beams, fig. 4.9 compares the mean energy of primary electrons as a function of depth calculated with the code ENXYZ to those calculated by Andreo and Brahme and used in the the IAEA Code of Practice [91, 13] and to those predicted by the Harder formula and used in the AAPM TG 21. Our results using monoenergetic incident beams and considering only primary electrons in the medium agree reasonably well with values calculated by Andreo and Brahme[91, 13] despite the differences in the Monte Carlo models used in the calculations(different codes and point source vs parallel beams). The maximum differences in calculated mean energies are 0.15 MeV, 0.3 MeV and 0.5 MeV for 10 MeV, 20 MeV and 30 MeV incident beams respectively. The discrepancies are reduced by a factor of two if parallel incident beams are used in our calculation.

We have calculated mean energies for several further cases with 10, 20 and 30 MeV monoenergetic incident beams as shown in fig. 4.10. In case 1, only primary electrons are considered in the calculation, as in fig. 4.9. In case 2, both primary and secondary electrons are considered but those secondary electrons set in motion by photons are not included. In case 3, all electrons are considered. In case 4, the electron cut off energy, $ECUT = 0.521$ MeV (total electron energy), is used in the calculation instead of $ECUT = 0.700$ MeV (total electron energy) as in all other calculations. It can be seen that:

- there is a significant drop in mean energy when secondary electrons are included in the averaging, especially at shallow depth. The decrease at $d_{\max}$ is 5%, 10% and 13% for 10 MeV, 20 MeV and 30 MeV beams respectively. At the tail there is an increase in the mean energy because only photons set secondary electrons in motion beyond R_p .
- changing ECUT from 0.700 MeV to 0.521 MeV has a small but not negligible effect on the calculated mean energy of all electrons. The further decrease in mean energy at $d_{\max}$ is 4.7%, 3.7% and 3.6% for 10 MeV, 20 MeV and 30 MeV beams respectively.

It is found by calculation that although knock-on electrons set in motion by phantom-generated bremsstrahlung photons play some role past R_p , they play no role in the mean energy over the entire therapeutic depth(not shown). This is because it is a second order effect. It is also found that the calculated mean energy of primary electrons is not sensitive to a change of ECUT from 0.700 MeV to 0.521 MeV. The curve with ECUT=0.521 (not shown) is virtually identical to that with ECUT=0.700 MeV. This is because there are very few primary electrons with energies below 0.700 MeV.

4.3.4 Mean energy as a function of depth for incident clinical electron beams

According to protocols for clinical electron beams $\bar{E}_d$ is determined based on two beam parameters, $\bar{E}_0$ and R_p . In the AAPM TG-21 protocol, $\bar{E}_0$ is determined from the relationship $\bar{E}_0=2.33R_{50}$ while in the IAEA Code of Practice, $\bar{E}_0$ is deduced from a table vs R_{50} . The mean incident electron energies determined according to the AAPM and the IAEA protocols for each beam are shown in table 4.1. In addition to the difference in $\bar{E}_0$ values determined according to the two protocols, the AAPM protocol[12] uses a simple formula, $\bar{E}_d = \bar{E}_0(1-d/R_p)$ while the IAEA Code of Practice[13] gives values tabulated as a function of $\bar{E}_0$ and d/R_p .

Fig. 4.11 shows a comparison between our calculated values and those obtained from protocols for electron mean energy along the central axis for an 18 MeV beam from a 'Clinac 2100C' accelerator. It can be seen that $\bar{E}_0$ determined according to protocols is overestimated by 0.5 MeV (3%) for an 18 MeV beam from a 'Clinac 2100C'. This leads to an overestimation of $\bar{E}_d$ by the protocols over the entire therapeutic depth. At $d_{\max}$, $\bar{E}_d$ is overestimated by 20% or 15% for this beam using the AAPM or IAEA protocols respectively. To study the discrepancy in detail we also calculated three different cases as shown in fig. 4.11. By comparing cases 1 and 2 it is seen that $\bar{E}_d$ is significantly lower when secondary electrons are included in the calculation, especially at shallow depth. By comparing cases 2 and 3, it is clear from fig. 4.11 that contaminant photons in the realistic beam have a negligible influence on $\bar{E}_d$.

Fig. 4.12 shows both dose profiles and the electron mean energy profile at different depths (surface, $d_{\max}$ and R_{50}) in a water phantom for the same beam as in fig 4.11. Mean energies are slightly higher (1 - 3%) near the field edges than near the central axis at all depths due to scattered electrons in water phantoms and in the beam as we discussed in section 4.3.1. The mean energy shows a roughly flat distribution across the field and drops quickly beyond the treatment field.

Fig. 4.13 compares the calculated $\bar{E}_d$ with values determined according to the AAPM and the IAEA protocol for four clinical electron beams with incident energy range from 5 to 50 MeV. It can be seen that the discrepancies vary with beam energies as well as the depths. For a 5 MeV beam from an SL75-20 at $d_{\max}$, $\bar{E}_d$ is overestimated by 15% according to the AAPM protocol and overestimated by 5% according to the IAEA protocol. For a 10 MeV beam from an SL75-20 at $d_{\max}$, $\bar{E}_d$ is overestimated by 26% or 10% according to the AAPM or the IAEA protocol respectively. For a 21 MeV beam from a KD2 at $d_{\max}$, $\bar{E}_d$ is overestimated by 30% or 18% according to the AAPM or the IAEA protocol respectively. For a 50 MeV beam from MM50 at $d_{\max}$, $\bar{E}_d$ is overestimated by 8% according to AAPM or IAEA protocol. These discrepancies are due to our use of more sophisticated models which include secondary

electrons and due to the problems in how the protocols assign the mean energy at the surface.

We also calculated $\bar{E}_d$ using a monoenergetic incident beam with an incident energy which is the same as the calculated $\bar{E}_o$ for the 10 MeV beam from the SL75-20 shown in fig. 4.13. The results show that $\bar{E}_d$ of the monoenergetic beams drops more slowly than that of the clinical incident beam near the phantom surface and then it drops more quickly than that of the clinical incident beam. The lower-energy electrons in the incident clinical beam generally have a large incident angle and are more easily scattered inside the phantom which increase the effect of fluence weighting. Also, at the large depths, $\bar{E}_d$ is dominated by the high-energy electrons in the clinical beam because the lower-energy electrons cannot reach there and thus the average energy is higher at depth in the clinical beam. Our result is consistent with the study by Udale-Smith[32].

Generally speaking, if $\bar{E}_o$ is overestimated according to the protocols, $\bar{E}_d$ will be underestimated over the entire therapeutic depth for any beam. If $\bar{E}_o$ is underestimated according to the protocols the discrepancies between calculated $\bar{E}_d$ and that determined according to protocols vary with the amount of underestimation of $\bar{E}_o$. In the case of the 5 MeV beam from an SL75-20 accelerator, $\bar{E}_o$ is underestimated by 2% according to IAEA protocol. This leads to a 3% underestimation of $\bar{E}_d$ at $d_{\max}$. In the case of 50 MeV beam from a MM50 accelerator, $\bar{E}_o$ is underestimated by 5% according to the IAEA protocol. This leads to an 8% overestimation of $\bar{E}_d$ at $d_{\max}$.

In most cases, $\bar{E}_o$ is underestimated using the relationship $\bar{E}_o=2.33R_{50}$. However, this underestimation in $\bar{E}_o$ actually reduces the discrepancy between our calculated values of $\bar{E}_d$ at $d_{\max}$ and those determined according to protocols.

Figure 4.1: Calculated electron mean energy averaged over square rings as function of off-axis position at a phantom surface for an 18 MeV, $10 \times 10 \text{ cm}^2$ beam from the Clinac 2100C. The average energy of all electrons near the field boundary is about 0.4 MeV higher than that near the central-axis while for direct electrons the average energy near the field boundary is only about 0.1 MeV higher than that near the central-axis.

The range scaling values recommended in the IAEA Code of Practice are based on the ratio of continuous-slowing-down ranges [13] (eqn 4.4) with values from ICRU Report 35, which differ by up to 0.2% from the values based on ICRU Report 37[95] values used everywhere else in the Code) These values generally overestimate our calculated depth-scaling factors by 1 - 2% for clear polystyrene or PMMA phantoms (off scale in fig. 4.16).

To investigate the influence of beam energy and quality on the depth-scaling factors we calculated the factors for a variety of beams from various accelerators with nominal beam energies from 5 to 50 MeV. The calculated depth-scaling factors (or effective densities) are plotted as a function of R_{50}^w in fig. 4.16. It can be seen that the variation of depth-scaling factors is small (<1.5%) for beam energies in the range 5 - 50 MeV and that the calculated depth-scaling factors have little variation with beam quality. We do a least-square-fit of the calculated data points for each phantom material to:

$$\rho_{eff}^p = A_p + B_p (R_{50}^w)^2 + C_p (R_{50}^w)^{\frac{1}{2}} \quad (4.10)$$

where ρ_{eff}^p is the fitted depth-scaling factor for plastic p and A_p , B_p and C_p are the fitting parameters. The values of fitted parameters for the three plastics are given in table 4.4 and the fitted curves are shown in fig. 4.16.

We also calculate the depth-scaling factors for monoenergetic point source incident beams for electron energies of 6, 10, 16 and 20 MeV. The calculated values of the depth-scaling factor as a function of R_{50} for monoenergetic beams are in a close agreement with those for realistic incident beams for the three plastics studied (not shown). This result further confirms that the calculated depth-scaling factors are independent of the beam quality but vary with beam energy. After scaling to common densities, our results of depth-scaling factors agree with the Monte Carlo calculations using monoenergetic beams by Grosswendt and Roos[92] within their stated uncertainties of the calculation, although they did not explicitly report their depth-scaling factors.

To study the influence of density variations for a practical nonwater phantom we change the density of a plastic phantom by 16% and calculate the depth-dose curve and mean energy vs depth curve as shown in fig. 4.17. It can be seen that the influence of density variation has been included if the depth-scaling factor of $R_{50}^{water}/R_{50}^{plastic}$ is applied. This indicates that the influence of density variation is negligible as long as the measured values of R_{50} in water and plastic are used for the depth-scaling.

If the user uses our calculated values shown in fig. 4.16 instead of measuring their R_{50} in water and in plastic, and if the mass density of the plastic used in the measurement is different from the one used in our calculations, the user must correct for the mass density effect by using:

$$\rho_{eff}^{usr} = \rho_{eff}^p \times \frac{\rho_{usr}^{plastic}}{\rho_{nom}^{plastic}} \quad (4.11)$$

where ρ_{eff}^p is the value from eqn 4.10 (fig. 4.16), and $\rho_{usr}^{plastic}$ and $\rho_{nom}^{plastic}$ are the mass densities of the plastics used in user's measurements and in our calculations respectively. For the above extreme case of changing the mass density of the PMMA from 1.19 to 1.00 g/cm³, eqn 4.11 gives the value of depth-scaling factor as $1.132 \times 1.00/1.19 = 0.951$ which is 0.7% less than the value obtained from the ratio of $R_{50}^{water}/R_{50}^{plastic} = 7.72/8.02 = 0.958$. However, in clinical practice the variation in density of a given type of plastic is usually less than 2 to 3%. Thus the discrepancies caused by using eqn 4.11 will be very small (<0.1%).

The above scaling factors, while deduced in terms of mean energies at depth, can also be used as range scaling factors as can be seen by the close agreement of the entire depth-dose curves in fig 4.15.

4.4 Conclusions

We have presented detailed data about the mean energies at the surface of the phantom for the simulated clinical beams from a variety of medical accelerators including

the Varian Clinac 2100C, the Philips SL75-20, the Siemens KD2, the AECL Therac 20 and the Scanditronix Medical Microtron 50. We have presented the mean electron energy at the surface of the phantom as a function of off-axis distance as well as the differences between the number averaged and the fluence averaged mean energy. It is shown that the mean energy near the field boundary is about 2% higher than that near the central-axis and that fluence averaged mean energy is about 2% lower than that of number averaged mean energy at the surface of the phantom. We have studied $\bar{E}_0$ and $E_{0,p}$ and their relationship with R_{50} and R_p for beams in the energy range 5–50 MeV. It is impossible to obtain a universal and highly accurate relationship between R_{50} and $\bar{E}_0$ because the number of scattered low-energy electrons varies with the machine's treatment head design. However, there exists an approximate relationship between R_{50} and $\bar{E}'_0$, the mean energy of direct electrons. This relationship is in close agreement with the energy-range relationships calculated using monoenergetic beams by Rogers and Bielajew [86] except at very high energies. Here the simple inverse square $1/r^2$ technique to calculate depth-dose curves for a point source from depth-dose curves for a parallel beam is shown to break down and explicit calculations with a point source are required.

We have shown that the empirical formula used by the AAPM and IAEA protocols to predict the most probable energy in the electron beam, *viz.* $E_{0,p} = 0.22 + 1.98R_p + 0.0025R_p^2$, is accurate for clinical beams with energies from 5 to 50 MeV. This formula reproduces the calculated data from realistic simulated clinical beams with an accuracy of 3%.

The electron mean energy, $\bar{E}_d$, as a function of depth is calculated in water as well as plastic phantoms. $\bar{E}_d$, calculated using realistic beams, is compared both with Harder's relationship, $\bar{E}_d = \bar{E}_0(1-d/R_p)$, employed in AAPM protocols and with values determined according to the IAEA Code of Practice. The conventional relations generally overestimate $\bar{E}_d$ over the entire therapeutic depth. For example, the AAPM or IAEA overestimates $\bar{E}_d$ by up to 20% or 15% respectively for an 18 MeV beam from

the University of Wisconsin's Clinac 2100C at d_{max} . It is also found that at all depths mean energies are 1 - 3% higher near the field edges than on the central axis. The differences in the mean energies between those calculated using clinical beams and those determined according to protocols vary with the beam quality. Discrepancies in $\bar{E}_d$ decrease when the beam's mean energy $\bar{E}_0$ is underestimated by using eqn 4.2 [11, 12, 13].

The differences in the mean energies between those calculated using clinical beams and those determined according to protocols reflect that:

- AAPM's simple formula, $\bar{E}_d = \bar{E}_0(1-d/R_p)$, is not accurate;
- IAEA's values are calculated using monoenergetic beams and only primary electrons are considered in the calculations;
- scattered low-energy electrons from the beam defining system in a clinical beam tend to lower the incident beam's mean energy for a given value of R_{50} .

At d_{max} the values of $\bar{E}_d$ determined according to the IAEA Code of Practice is up to 1 MeV lower than that determined according to the AAPM protocol, depending on the beam energy as well as beam quality. It should be emphasized that in both protocols, $\bar{E}_d$ is only used to specify which fluence correction factors to use since these factors, which only apply at d_{max} , were originally specified this way. Thus, at the practical level, if more accurate values of $\bar{E}_d$ are used in clinical situations, the original data must also be re-analyzed using the more accurate data. However the IAEA Code of Practice has provided improved values of $\bar{E}_d$ without modifying the original data for fluence correction factors[97]. This inconsistency leads to only a 0.1% error in electron fluence correction factors determined according to the IAEA Code of Practice. This error is small compared to the other uncertainties in the fluence correction factor.

Using the same incident clinical beam we calculate mean energy in different nonwater phantoms. The scaling factors used to convert depth in plastic to water equivalent

depth are obtained. By scaling the depth in the nonwater medium to the depth in water where the mean energies are equal, we are able to obtain the scaling factor directly. The depth-scaling factor is found to be a constant as a function of depth with a small variation ($<1.5\%$) depending on the incident beam energies. The calculated depth-scaling factor is found to be equal to $R_{50}^{water}/R_{50}^{plastic}$ which is consistent with the AAPM definition of effective density. However the discrepancies between our calculated values and those recommended by the AAPM and the IAEA protocols are up to 2%. The values of the depth-scaling factor obtained by using the ratio of the continuous-slowing-down ranges employed in the IAEA Code of Practice are shown to be inaccurate and overestimate our calculated values by up to 1 – 2% in all cases.

Our calculated values of depth-scaling factor for three plastics are shown in fig. 4.16. If the mass density of the user's plastic phantom is different from what we used in the calculation, eqn 4.11 should be used to correct for this density effect. The curves in fig. 4.16 can also be used for range scaling, *e.g.* to determine R_{50} in a water phantom from a measured value in a plastic phantom or vice versa.

Table 4.1: Comparison of values of $\bar{E}_0$, the mean energy of electrons inside the beam on the phantom surface, as calculated for the realistic beams (for $\Delta=189$ keV) and as determined according to the AAPM TG-21[11] and the IAEA [13] protocols and the tabulations of Rogers and Bielajew[86] (RB-86) based on the value of R_{50} . The values of R_{50} and $d_{\max}$ are from the calculations.

Machine	E_{nominal} MeV	R_{50} (cm)	$d_{\max}$ (cm)	calculated		$\bar{E}_0$ (MeV)		
				$\bar{E}_0$	$\bar{E}'_0$	$2.33R_{50}$	IAEA	RB 86
Clinac 2100C	6	2.63	1.4	6.11	6.52	6.13	6.21	6.61
	9	4.00	1.9	9.10	9.75	9.32	9.40	9.68
	12	5.20	2.7	11.74	12.27	12.12	12.23	12.48
	15	6.50	2.1	14.59	15.31	15.15	15.25	15.51
	18	7.72	1.9	17.35	18.33	17.96	17.82	18.43
SL75-20	5	2.08	1.1	5.07	5.41	4.85	4.97	5.27
	10	4.12	2.1	9.34	10.08	9.60	9.64	10.00
	14	5.98	1.7	13.28	14.53	13.93	13.95	14.33
	17	6.96	1.9	15.28	16.67	16.22	16.32	16.64
	20	8.10	1.8	17.76	19.55	18.87	18.75	19.36
KD2	6	2.31	1.3	5.53	5.75	5.38	5.52	5.87
	11	4.21	2.5	9.70	10.06	9.81	9.82	10.23
	21	8.30	3.0	18.74	19.42	19.34	19.25	19.87
Therac 26	6	2.18	1.3	5.50	5.52	5.08	5.20	5.56
	9	3.42	2.1	8.25	8.30	7.97	8.05	8.40
	13	5.14	3.2	12.11	12.26	11.98	12.09	12.39
	17	6.85	4.2	15.93	16.16	15.96	16.10	16.38
	20	8.10	4.2	18.74	19.09	18.87	18.75	19.36
MM50	25	10.36	3.0	24.53	24.56	24.14	24.21	25.06
	40	15.47	2.5	38.90	38.98	36.05	37.55	39.62
	50	18.55	2.8	48.90	49.05	43.22	46.38	49.90

Table 4.2: Comparison of values of R_p from the Monte Carlo calculations and experiments and comparison of values of $E_{o,p}$ obtained using the Monte Carlo simulations and those using the ICRU formula (eqn 4.1) based on calculated R_p for various clinical electron beams.

Machine	$E_{nominal}$ MeV	R_p (cm)		$E_{o,p}$ (MeV)	
		Monte Carlo	$R_p^{exp} - R_p^{MC}$	Monte Carlo	$E_{o,p}^{MC} - E_{o,p}^{ICRU}$
Clinac 2100C	6	3.26	0.00	6.88	0.18
	9	4.99	0.01	10.39	0.23
	12	6.25	0.02	13.05	0.36
	15	7.92	0.05	16.31	0.25
	18	9.44	0.12	19.49	0.36
SL75-20	5	2.60	0.03	5.66	0.28
	10	5.10	0.05	10.66	0.28
	14	7.43	-0.04	15.34	0.27
	17	8.70	—	17.79	0.15
	20	10.10	-0.20	21.04	0.57
KD2	6	2.84	0.01	6.05	0.19
	11	5.12	-0.02	10.51	0.09
	21	10.00	0.10	20.59	0.32
Therac 20	6	2.68	0.01	5.60	0.06
	9	4.10	0.01	8.42	0.04
	13	6.10	0.00	12.41	0.02
	17	8.05	0.03	16.35	0.03
	20	9.59	0.01	19.30	-0.14
MM50	25	12.23	0.10	24.77	-0.04
	40	19.20	0.12	39.50	0.34
	50	23.70	0.09	49.69	1.14

Table 4.3: Densities and effective densities or depth-scaling factors from nonwater materials to water for an 18 MeV electron beam from a Clinac 2100C.

Material	ρ (g/cm ³)		scaling-factor ρ_{eff}		scaling factor IAEA ^e
	ours ^a	AAPM ^b	ours ^c	AAPM ^d	
polystyrene	1.060	1.045	1.001	0.975	1.017
white PST	1.048	1.055	0.999	0.99	
PMMA	1.190	1.18	1.132	1.115	1.144
graphite	2.000	-	1.729		

^aused in the Monte Carlo calculations from ICRU37[95].

^bgiven in the AAPM TG-25 protocol[12].

^ccalculated using calculated values of R_{50} and the definition $\rho_{eff} = R_{50}(\text{in water})/R_{50}(\text{in medium})$ with an 18 MeV beam from Clinac 2100C.

^drecommended in the AAPM TG-25 protocol[12], these are independent of $\bar{E}_0$.

^erecommended in the IAEA Code of Practice where the mass densities are same as those used in our calculation. Since the tabulated values given in the IAEA Code of Practice depend on $\bar{E}_0$ (1.027 - 0.997 (polystyrene) and 1.149 - 1.133 (PMMA) for $\bar{E}_0$ from 5 - 50 MeV respectively) we choose the value corresponding to $\bar{E}_0 = 15$ MeV which is close to the beam we used in the calculation.

Table 4.4: The fitted parameters for the effective density given as $\rho_{eff}^p = A_p + B_p (R_{50}^w)^2 + C_p (R_{50}^w)^{\frac{1}{2}}$ (eqn 4.10) as a function of R_{50}^w for clear polystyrene, white polystyrene and PMMA phantoms.

phantom materials	A_p	B_p (cm^{-2})	C_p ($\text{cm}^{-\frac{1}{2}}$)
clear polystyrene	0.9689	-1.016×10^{-4}	1.356×10^{-2}
white polystyrene	0.9762	-1.232×10^{-4}	1.092×10^{-2}
PMMA	1.1177	-9.008×10^{-5}	7.096×10^{-2}

Figure 4.1: Calculated electron mean energy averaged over square rings as function of off-axis position at a phantom surface for an 18 MeV, $10 \times 10 \text{ cm}^2$ beam from the Clinac 2100C. The average energy of all electrons near the field boundary is about 0.4 MeV higher than that near the central-axis while for direct electrons the average energy near the field boundary is only about 0.1 MeV higher than that near the central-axis.

Figure 4.2: Calculated electron mean energy averaged over square rings as function of off-axis position at a phantom surface for a 10 MeV, $10 \times 10 \text{ cm}^2$ beam from the SL75-20. The average energy of all electrons near the field boundary is about 0.3 MeV higher than that near the central-axis while for direct electrons the average energy near the field boundary is only about 0.1 MeV higher than that near the central-axis.

Figure 4.3: Calculated electron mean energy averaged over square rings as function of off-axis position at the phantom surface for a 6 MeV, $10 \times 10 \text{ cm}^2$ beam from the Therac 20. There are very few electrons contributing to the average outside the field edge at 5 cm.

Figure 4.4: Energy-range relationship for clinical electron beams as calculated for various realistic beams at an SSD of 100 cm. In the symbol description, “all” means “all electrons (inside 10×10 cm² field) including scattered electrons from all beam defining components”, and “direct” means “these electrons (inside 10×10 cm² field) which exclude scattered electrons interacting with beam defining components”. At the upper right corner of the figure, the symbols of + and × show the results from using the inverse square $1/r^2$ corrected 50 MeV monoenergetic parallel beam and 50 MeV monoenergetic point source respectively.

Figure 4.5: Errors introduced by using the inverse-square-law approximation to convert depth-dose curves for parallel beams to depth dose-curves for point sources for a 50 MeV monoenergetic incident beam. The incident beam field is $20 \times 20 \text{ cm}^2$ in both cases and SSD is 110 cm for the point source beam. The R_{50} values are 18.87 and 18.58 cm for the point source incident beam and the parallel incident beam that has been corrected to $\text{SSD} = 110 \text{ cm}$ using the inverse-square-law respectively.

Figure 1.6: The relationship between most-probable energy $E_{o,p}$ and practical range R_p for various clinical electron beams in a water phantom. $E_{o,p}$ and R_p of clinical beams are obtained from the spectra and calculated central-axis depth-dose curves of simulated beams. Values of Rogers and Bielajew[86] are from Monte Carlo calculations by Rogers and Bielajew[86, 12]. Values of SB-94 are from analytical energy-range relationships by Sorcini and Brahme[87].

Figure 4.7: $E_{in} - \bar{E}_0$ and $E_{in} - \bar{E}'_0$ vs R_{50} , where $\bar{E}_0$ is the average energy of all electrons at the phantom surface and $\bar{E}'_0$ is the average energy of direct electrons at the phantom surface. The direct electrons do not include the electrons scattered from the beam defining system.

Figure 4.8: $\Delta E - \Sigma dt(dE/dt)$ vs R_{50} , where filled symbols correspond to $\Delta E = E_{in} - \bar{E}_o$ and empty symbols correspond to $\Delta E = E_{in} - \bar{E}'_o$ where $\bar{E}'_o$ is the average energy of direct electrons at the phantom surface. The direct electrons do not include the electrons scattered from the beam defining system.

Figure 4.9: Comparison of the mean energy of primary electrons calculated with the code ENXYZ to those calculated by Andreo and Brahme[91] and recommended in the IAEA Code of Practice and those calculated using Harder's relation employed in the AAPM protocols.

3

PM-1 3½"x4" PHOTOGRAPHIC MICROCOPY TARGET
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monoenergetic incident beams

Figure 4.10: Comparison of the calculated mean energy of primary electrons and that of both primary and secondary electrons.

Figure 4.11: Comparison between calculated mean energy and that obtained from protocols for an 18 MeV beam from a Clinac 2100C accelerator. In case 1, only electrons are included in the realistic incident beam and only primary electrons are considered in the water phantom. In case 2, only electrons are included in the realistic incident beam and both primary and secondary electrons are considered in the water phantom. In case 3, both electrons and contaminant photons are included in the realistic incident beam and both primary and secondary electrons are considered in the water phantom.

Figure 4.12: a) Calculated dose profile and b) mean-energy profile at different depths in a water phantom for an 18 MeV beam from the Clinac 2100C. In the lower panel, solid lines are the mean energy of primaries in phantom and only including electrons in the incident beams. Dotted lines are the mean energy of both primary and secondary electrons and only including electrons in the incident beams. Dashed line are the mean energy of both primary and secondary electrons and including electrons and contaminant photons (complete realistic beam).

Figure 4.13: Comparison between calculated $\bar{E}_d$ for full beams with $\Delta=189$ keV and that from protocols for 5, 10, 21 and 50 MeV clinical electron beams. In the right upper panel the electron incident energy of the monoenergetic point source is 9.34 MeV which is the same as the calculated mean energy of the 10 MeV beam from the SL75-20.

Figure 4.14: Calculated central-axis depth dose curves and mean energies $\bar{E}_d$ as a function of depth in different phantom materials for an 18 MeV beam from the Clinac 2100C. The values of the mass density used in the calculation are: $\rho_{\text{polystyrene}}=1.060$ g/cm³, $\rho_{\text{white polystyrene}}=1.048$ g/cm³, $\rho_{\text{PMMA}}=1.190$ g/cm³ and $\rho_{\text{graphite}}=2.00$ g/cm³. The ECUT used in the calculation is 0.700 MeV.

Figure 1.15: Calculated depth-dose curves and $\bar{E}_d$ values for different media for an 18 MeV beam from the Clinac 2100C after scaling to the water equivalent depth using the scaling factors, effective density $\rho_{\text{eff}} = R_{50}(\text{in water})/R_{50}(\text{in medium})$.

Figure 4.16: Calculated depth-scaling factors as a function of R_{50}^w for clear polystyrene, white polystyrene and PMMA phantoms using clinical incident electron beams.

Figure 4.17: Calculated depth-dose curves and $\bar{E}_d$ using PMMA with a different density and applying the scaling factor $\rho_{\text{eff}} = R_{50}(\text{density}=1.19)/R_{50}(\text{density}=1.00)$ for the 18 MeV beam from a Clinac 2100C.

Chapter 5

Calculation of stopping-power ratios using realistic clinical electron beams

5.1 Introduction

Spectrum-averaged electron restricted mass collision stopping-power ratios, denoted $\left(\frac{\bar{L}}{\rho}\right)_{\text{air}}^{\text{med}}$ in AAPM notation and $s_{\text{med,air}}$ in ICRU notation, play a central role in radiation dosimetry. They appear in a fundamental equation in radiation dosimetry which relates an ionization chamber reading of the charge measurement to the dose to a medium when the medium is irradiated by a photon or electron beam, viz:

$$D_{\text{med}} = MN_{\text{gas}} \left(\frac{\bar{L}}{\rho}\right)_{\text{air}}^{\text{med}} P_{\text{ion}} P_{\text{repl}} P_{\text{wall}} \quad (5.1)$$

where M is the charge from the ion chamber corrected to standard atmospheric conditions, N_{gas} is the cavity gas calibration factor and the three P factors are corrections which appear in the AAPM or other dosimetry protocols [11]. The ratio of water to air stopping powers is about 15% greater for electrons with an energy of 1 MeV compared to those at 20 MeV because of the difference in density effects for the dense and gas phase media. Thus the stopping-power ratios depend strongly on the incident beam energy and increase with depth as the electrons slow down. The traditional method of determining stopping-power ratios is to assign a mean energy to the electron beam at the phantom surface, and use tabulated values of stopping-power ratios *vs* depth calculated for mono-energetic incident beams.

ICRU Report 35 contains a full discussion of the calculation of stopping-power ratios [82]. Several authors have reported extensive series of calculations for use in radiation dosimetry. For electron beams, Berger and Seltzer of NIST were the pioneers in the field [82, 98] and their calculated stopping-power ratios for electron beams are used in the AAPM and IAEA protocols [11, 13, 12].

The energy and angular distributions of electrons and contaminant photons at the phantom surface are the most important characteristics of clinical electron beams. Andreo and co-workers have shown that insufficient information about these distributions may introduce an uncertainty in dosimetry [18, 69].

If a low-energy component is added to an electron spectrum, the stopping-power ratio near the surface increases because the stopping-power ratio for low energy electrons is higher than that for high-energy electrons. If the stopping-power ratio is calculated as if the beam were mono-energetic at the correct mean energy, then at large depths the realistic spectrum will have a lower stopping-power ratio than that of the mono-energetic beam because the high-energy electrons in the realistic spectrum will penetrate further and these higher energy electrons have a reduced stopping-power ratio. Based on these simple arguments, one expects the stopping power ratio vs depth curves for a realistic beam to cross that for the corresponding mono-energetic beam. The detailed calculations of Andreo *et al.*[18, 69] have confirmed this expectation.

In one paper, Andreo *et al.*[18] used the Monte Carlo method to estimate the influence of energy and angular spread on the water/air stopping-power ratios in electron beams by using simple models of the angular and energy spread of the incident electrons. For electron beams around 10 MeV they showed that beams with broad energy and angular distributions had stopping-power ratios at $d_{\max}$ which can be up to 1.0 or even 1.5% *lower* than that for mono-energetic parallel beams with the same most-probable energy. Although Andreo *et al.* could derive a somewhat hard-to-use expression for relating these stopping-power ratios, they quite rightly emphasized the more important case of showing the difference between the stopping-power ratio which would be determined by using protocols to determine $\bar{E}_0$ to establish the stopping-power ratio from the mono-energetic data, and the stopping-power ratio determined from the more realistic beam. For their simplified "realistic" beams near 10 MeV they showed *increases* of up to 0.5% might be expected at $d_{\max}$ for typical clinical accelerator beams.

Andreo and Fransson[69] extended the previous study by utilizing the actual electron and photon energy spectra and angular distributions calculated by Udall[33] for the 10 MeV beam from a Philips SL75-20 scattering-foil accelerator with a tubular applicator. They experienced difficulties since their calculated depth-dose curve did

not agree either with that calculated by Udale herself with the same data nor with experiment. They convincingly showed that sampling independently from energy and angular distributions is incorrect in principle because they are highly correlated, but sampling independently was the best they, or anyone else had been able to do. They showed that the angular distribution of the electrons, while important to their calculated depth-dose curves, had very little effect on the calculated stopping-power ratios. They also showed that the photon contamination had a "small influence" on the stopping-power ratios for this beam. Their major result was consistent with the earlier result[18], and they estimated that the effect of the angular and energy distributions would be well less than 0.5% at $d_{\max}$.

Klevenhagen has recently pointed out the importance of photon contamination of the incident beam when determining dose in electron beams[99]. Using several approximations and measured data for 2 specific accelerators, he estimated that the effect on the stopping-power ratios was to increase them by 0.3 or 0.4% at 10 MeV and up to 1.2% at 20 MeV. Aware of the lack of good data, he suggested the need for Monte Carlo calculations.

We have recently developed a very general EGS4 Monte Carlo simulation code, called BEAM, for simulating the beams from radiotherapy machines[77]. It has been extensively validated against measured depth-dose curves for electron beams from 7 different accelerators. By treating the energy of the mono-energetic beams from the accelerator vacuum as a parameter which was varied to match the calculated and measured values of R_{50} , it was found that depth-dose curves could be reproduced within 3% at all depths. In order to overcome the problems concerning the correlation between energy and angle[33, 69], we have stored the phase space coordinates (energy, direction, position, charge) of each particle leaving the accelerator and entering the patient plane. Such realistic beams contain not only scattered lower-energy electrons but also a large number of contaminant photons as seen in fig 5.1.

Since the beam quality varies with accelerators, particularly between scattered

and scanned beam accelerators, it is important to investigate the differences in the calculated stopping-power ratios for a variety of machines. In this investigation we calculate electron stopping-power ratios in electron beams from five different clinical accelerators including the Varian Clinac 2100C, Philips SL75-20, Siemens KD2, AECU Therac 20 and Scanditronix Racetrack Microtron MM50 with electron energy range from 5 to 50 MeV. Our results confirm and extend those of Andreo *et al.*[18, 69] concerning the minor role of angular distributions of incident particles on stopping-power ratios. However, we will show that Andreo and Fransson [69] studied an accelerator for which photon contamination happens to play a small role. In general, although not quite as large as estimated by Klevenhagen[99], photon contamination is important, especially at higher energies where the radiative yield increases and hence there is more photon contamination which is also higher in energy (see fig 5.1).

Following the procedure of Andreo *et al.*[18, 69] our calculated results are compared with values determined according to the various dosimetry protocols. Correction factors have been obtained to correct the stopping-power ratios determined according to the dosimetry protocols for clinical beams at d_{max} or reference depth. Protocols provide water/air stopping-power ratios as a function of mean energy at the phantom surface and at the depth of measurement. The mean energy at the phantom surface, $\bar{E}_0$, is determined from the depth at which the absorbed dose or ionization falls to 50% of its maximum level. Wu *et al.*[89] pointed out that the TG 21 protocol suggested using ionization-depth curves whereas the procedure to be applied was based on data for depth-dose curves. Although the NACP and IAEA protocols take this effect into account, in section 4, we quantitatively examine the differences in the stopping-power ratios determined by using the 50% ionization level I_{50} instead of using d_{50} , the depth at which the absorbed dose level falls to 50% of D_{max} . In the meantime, throughout this paper, we apply the AAPM procedure to the actual absorbed-dose *vs* depth curves.

In the AAPM (1983) protocol[11] the mean energy of the electron beam at the

surface of the phantom, $\bar{E}_0$, is given by $\bar{E}_0 = 2.33d_{50}$, while in the IAEA protocol[13] the relationship between $\bar{E}_0$ at the phantom surface and R_{50} at $SSD = 100\text{ cm}$ is tabulated. Using the EGS4 code, Rogers and Bielajew(RB-86)[86] calculated $\bar{E}_0/R_{50}$ as a function of R_{50} for various values of SSD. Their results are recommended for more accurate estimates of the mean energy of electron beams up to 25 MeV in the AAPM TG-25 Report[12]. The relationships between $\bar{E}_0/R_{50}$ and R_{50} obtained from the AAPM (1983)[11], IAEA (1987)[13] and Rogers and Bielajew(RB-86)[86] are shown in fig. 5.2. The difference between IAEA (1987) and AAPM (1983) procedures is very small in the energy range from 6 to 20 MeV. The mean energy determined according to Rogers and Bielajew[86] is about 400 keV higher than that determined according to the $\bar{E}_0 = 2.33 R_{50}$ relationship for beam energies from 5 to 20 MeV. However, $\bar{E}_0$ determined according to Rogers and Bielajew(RB-86)[86] and $\bar{E}_0 = 2.33 R_{50}$ can differ by 6.4 MeV (i.e. a 13% difference) for a 50 MeV beam. The values of water/air stopping-power ratios selected using these two different mean energies can differ by up to 1% and 3% for 5 MeV and 50 MeV electron beams respectively as shown in fig. 5.3. Throughout this work, we have used interpolations of the tabulated data in the paper by Rogers and Bielajew(RB-86)[86] to determine $\bar{E}_0$ from R_{50} , not the fitted formula since it does not cover the entire energy range of interest.

5.2 Method of Calculations

Throughout this work, stopping-power ratios determined using mono-energetic data from the AAPM TG21 (or equivalently IAEA) protocol are determined by linear interpolation of the tabulated values in energy.

The calculation of the stopping-power ratios for realistic clinical beams in this study is carried out in two steps, *viz.* the generation of the clinical incident beams followed by their use as incident beams to compute the water/air stopping-power ratios.

5.2.1 The incident clinical beams

To obtain realistic clinical electron beams, a series of Monte Carlo simulations have been undertaken using the EGS4 [15] user-code BEAM[77]. The position, energy, charge and direction of every particle emerging from the accelerator are stored in a "phase-space" data file. The energy cutoffs for particle transport are set to ECUT = AE = 0.700 MeV (total electron energy) and PCUT = AP = 0.010 MeV. In all cases, the PRESTA electron algorithm[53] is employed with ESTEPE = 0.04.

The clinical electron beam field sizes are $10 \times 10 \text{ cm}^2$, $25 \times 25 \text{ cm}^2$ and $20 \times 20 \text{ cm}^2$ for beam energies 5 – 20 MeV, 25 – 40 MeV and 50 MeV, respectively. The SSD is 100 cm and 110 cm for energies 5 – 40 MeV and 50 MeV, respectively. In the beam simulation, the energy of electrons at the exit vacuum window of an accelerator is adjusted to match the R_{50} between the measured and Monte Carlo calculated depth-dose curves. The depth-dose curves are calculated using the EGS4 user-code DOSXYZ[14, 27] which has been extended to accept the phase-space data files as inputs.

For the accelerator beams utilized here, descriptions of the specific accelerators, relative numbers and average energies of photons and electrons and many other data, including comparisons to measured depth-dose data are given in the BEAM paper[77]. The agreement with experiment is within 3% everywhere, and better in most cases. In particular the agreement with experiment for the 10 MeV beam from the SL75-20 accelerator is much better than obtained by Andreo and Fransson[69]. At least part of this improvement is because we could adjust the incident energy of the electron beam from vacuum to match R_{50} and include full correlation between the electron energy and angle.

5.2.2 The EGS4 user-code SPRRZ

The “phase-space” file produced by BEAM is used as input to the EGS4/PRESTA NRC user-code SPRRZ (Rogers and Bielajew, unpublished) to calculate Spencer-Attix water/air stopping-power ratios. The code SPRRZ has been used extensively[100, 48] and has been shown to give the same results as previous calculations using EGS4[94], which are, in turn, the same as previous calculations used in protocols[11, 13]. The stopping powers used in the calculations are the ICRU Report 37 values [95] as implemented in EGS4[96]. The Monte Carlo code tracks the electron histories down to an energy Δ which has been taken to be 10 keV as done in major protocols [11, 13]. The code SPRRZ assumes cylindrical symmetry and the stopping-power ratios can be calculated along the central-axis for columns with different radii. We find that the calculated stopping-power ratios are not sensitive to the radius as long as it is much smaller than the beam field-size. For a beam with field size of 10×10 cm², the calculated stopping-power ratios are the same for regions with radii of 1 or 2 cm. In our calculations, a radius of 1 cm is used. The calculations are done using default PRESTA parameters[53] after tests using ESTEPE=0.01 showed that the default values were accurate.

5.3 Results of Monte Carlo calculations

5.3.1 Comparison to previous results using mono-energetic electron beams

For 9 MeV and 18 MeV electron beams, fig 5.4 presents Spencer-Attix water to air stopping-power ratios. Two sets are calculated with the code SPRRZ and the other calculated by Berger and used in the AAPM protocol [11, 98] and the IAEA Code of Practice[13]. Away from the surface our results calculated using plane-parallel

or point source beams agree with AAPM values very well, typically within 0.1%, despite the differences in the incident beams: our point source or parallel beams have finite size ($10 \times 10 \text{ cm}^2$) while Berger's are broad parallel beams. Although the calculated central-axis depth-dose curves are different for the parallel and the point source beams, the energy spectra at a given depth are almost the same along the central axis. The change in the shape of the depth-dose curves is mainly caused by the change in the total electron fluence due to beam divergence as opposed to changes in the electron spectra. Since the calculated stopping-power ratios only depend on the spectrum, the same stopping-power ratios are obtained using the parallel and point source incident beams. The higher values at the surface in the AAPM's stopping-power ratios have been discussed elsewhere[94] and reflect the assumption of charged particle equilibrium for low-energy electrons used in the calculations for the AAPM protocols.

5.3.2 Results

Fig. 5.5(a) shows the calculated central-axis percent depth-dose (%DD) curve for a 20 MeV electron beam from the Philips SL75-20 linear accelerator which contains many scattered lower-energy electrons since it has fairly thick scattering foils and a solid-walled applicator. The figure also shows the dose component due to the contaminant photons incident on the water phantom. Fig. 5.5(b) compares our calculated stopping-power ratios and those determined by using the depth, R_{50} , to establish the mean energy at the surface using $\bar{E}_0 = 2.33R_{50}$ or the prescription in the paper by Rogers and Bielajew(RB-86)[86]. $\bar{E}_0$ determined according to $\bar{E}_0 = 2.33R_{50}$ and RB-86[86] are 18.87 MeV and 19.36 MeV respectively. The average kinetic energy of electrons with energy above 189 keV at the phantom surface simulated using BEAM is 17.81 MeV while the most probable energy occurs at 21.2 MeV. It can be seen that stopping-power ratios determined using the RB-86[86] prescription under-estimate at shallow depth and over-estimate at large depth by up to 1.5% and 1.8% respec-

tively. In order to understand fully the causes of the deviation in the stopping-power ratios between the clinical and the mono-energetic beams we isolated each component and studied contributions from each. Table 5.1 delineates the various types of calculations done for each beam. There is no difference in the stopping-power ratios calculated using the beam with only the electrons from the clinical beam and those calculated using a point source of electrons with the same spectrum as in the clinical beam, except for a small difference near the phantom surface. This is because the low-energy electrons in the clinical beam are often scattered from the applicator at large angles and thus affect the stopping-power ratios near the surface. The close agreement at other depths between the calculations which include all angular effects vs those coming directly from a point source at an SSD of 100 cm demonstrates that angular effects play virtually no role.

For the 20 MeV electron beam of the SL75-20 accelerator the increase in the stopping-power ratios for the clinical beam at the depth of dose maximum (≈ 1.5 cm) is mainly caused by the electron energy spread ($\approx 0.9\%$, as shown by the difference between the curves for the mono-energetic calculations and those for the “electron only” or “electron spectrum” curves) and caused by the contaminant photons ($\approx 0.5\%$, as seen by the difference between the “electron only” curve and the “electron and photon” curve for the realistic beam). Although the contaminant photons in the realistic clinical beam contribute only about 5% of the maximum dose as shown in fig. 5.5(a), the influence on the stopping-power ratios is significant because the stopping-power ratios calculated from these incident photons are much higher than those from electrons (as pointed out by Klevenhagen[99]).

Similar results for a 10 MeV beam from the Philips SL75-20 linear accelerator are obtained as shown in fig. 5.6. This is the same accelerator as studied by Andreo and Fransson[69]. Consistent with their results, the effects of angular distributions and photon contamination are very small for the beam. In fact, even the effect of the electron energy distribution is particularly small near $d_{\max}$ compared to values

predicted using the IAEA protocol.

Fig. 5.7 and 5.8 present the results for 18 and 9 MeV electron beams respectively from the Varian Clinac 2100C linear accelerator at the University of Wisconsin. This machine has a different beam-defining system, in particular an open applicator. The electron beam's energy spectrum is somewhat less broad and the electron's angular spread is relatively smaller. Since the change in the stopping-power ratios due to the angular distributions is small except near the surface, similar changes in the stopping-power ratios are obtained as for the beams from the SL75-20 machine.

Fig. 5.9 and 5.10 present the results for 21 and 11 MeV electron beams respectively from the Siemens KD2 linear accelerator. Due to its beam defining system, the beam has very little angular spread. There is no difference in the stopping-power ratios calculated using the beam with only the electrons from the clinical beam and those calculated using a point source of electrons with the same spectrum as in the clinical beam, even near the phantom surface. The change in the stopping-power ratios due to the contaminant photons is similar to the change for the beams from the SL75-20 machine.

Fig. 5.11 and 5.12 are the results from 20 and 9 MeV electron beams respectively from the AECL Therac 20 linear accelerator. This accelerator uses a magnetically scanned beam[77] to achieve a uniform beam profile. Since there are no scattering foils, only a monitor ion chamber with a total thickness of only 0.02 g/cm^2 in the beam, there is little electron energy spread and little photon contamination. The figures show that the contribution to the bremsstrahlung tail from the contaminant photons is negligible. Since the realistic beam is close to mono-energetic, there is only a small difference in the stopping-power ratios calculated using the realistic beam or by using mono-energetic stopping-power ratio data with energies determined using the RB-86[86] prescription to determine the incident mean energy.

Fig. 5.13 and 5.14 give the results for 50 and 25 MeV electron beams from the Racetrack MM50 accelerator. The MM50 uses magnetic scanning to give a uniform

beam. Unlike other accelerators, this machine does not have electron applicators at these beam energies. The beam field size is defined by collimators and multileaf blocks only. The beams are therefore nearly mono-energetic and have few electrons with large angle deflections. Here we see the biggest difference in selected stopping-power ratios due to the determination of $\bar{E}_o$ using RB-86[86] or the relationship $\bar{E}_o=2.33R_{50}$. It is clear that $\bar{E}_o=1.33R_{50}$ under-estimates $\bar{E}_o$.

5.4 Discussion

5.4.1 Effects of Photon Contamination

The effect of contaminant photons in electron beams is to increase the stopping-power ratios in the therapeutic range as seen from the difference between the “electron only” and the “electron and photon” curves in figures 5.5 to 5.14. The percentage increases due to the contaminant photons for a variety of beams (including calculations at many beam energies not shown above) are shown in fig. 5.15 as a function of D_X , the percentage dose in the bremsstrahlung tail, 2 cm past the projected or practical range, R_p . As seen in the earlier figures, D_X is mostly determined by the contaminant photons except in the case of scanned beams where the bremsstrahlung tail generated within the phantom is dominant. It can be seen from fig. 5.15 that the increase in the stopping-power ratios due to the contaminant photons is, not surprisingly, proportional to D_X (except for the scanned beams) and thus to the amount of photon contamination in the beam. The increases are up to 0.6% near 20 MeV. This is about 1/2 of the increase estimated by Klevenhagen[99] for similar energy beams. For beams near 10 MeV our calculations predict a 0.1% increase due to photon contamination. This is consistent with Andreo and Fransson’s result[69] but less than Klevenhagen’s estimate of 0.3 to 0.4%.

5.4.2 Correction factors for the stopping-power ratios at reference depth

The most critical application of stopping-power ratios is for absolute dosimetry done at some reference depth, traditionally $d_{\max}$ or close to it. In this section we present the different "correction" factors to be applied to stopping-power ratios determined using the mono-energetic stopping-power ratios in the AAPM TG21 protocol and IAEA code of practice and 3 different prescriptions for obtaining the appropriate mono-energetic incident beam energy, $\bar{E}_0$. The use of different prescriptions for $\bar{E}_0$ implies the corrections are different, but the stopping-power ratio determined at a given reference depth in a given clinical beam is the same no matter which prescription is used, *i.e.* the correction factors presented also correct for any errors in the choice of $\bar{E}_0$ (and in fact the choice is irrelevant in this context).

Fig. 5.16 presents the correction required for stopping-power ratios at $d_{\max}$ determined using the mono-energetic stopping-power ratio data of the AAPM TG21 (or equivalently, the IAEA Code of Practice) and $\bar{E}_0$ determined following the prescription of RB-86[86]. By plotting the correction factors as a function of the bremsstrahlung tail, D_X , we find that all values for beams with scattering foils fall close to a straight line. The scanned beam case will be discussed later.

For beams with scattering foils, the corrected water/air stopping-power ratios at $d_{\max}$ can be derived using:

$$\left(\frac{\bar{L}}{\rho}\right)_{\text{air}}^{\text{water}} = (1 + A_s D_X) \left[\left(\frac{\bar{L}}{\rho}\right)_{\text{air}}^{\text{water}} \right]_{\text{RB-86}} \quad (5.2)$$

where: the uncorrected stopping-power ratio on the right is determined using the RB-86 data for assigning the mean energy of the incident beam and the values in the TG-21 or IAEA protocols are used; $A_s = 0.0027$ is a numerical constant obtained from the slope of the straight line on fig. 5.16; and D_X is the percentage dose in the bremsstrahlung tail, 2 cm past R_p . For consistency, the calculated values of D_X were

used to determine this equation but to apply the equation the measured values would be used. The measured and calculated values of D_X generally were found to be in excellent agreement[77].

The bremsstrahlung tail is used to specify the corrections for the stopping-power ratios since it is convenient in clinical practice. The dose contributions from contaminant photons in the clinical beam dominate the bremsstrahlung tails in the depth-dose curve except for scanned-beam accelerators which have fewer contaminant photons. The majority of photon contamination arises in the scattering foil which is also largely responsible for broadening the energy distribution of the primary electrons. Since the correction at $d_{\max}$ is mainly due to the electron energy spread and contaminant photons, the bremsstrahlung tail turns out (somewhat fortuitously) to be a good indicator of the correction needed.

Fig. 5.17 presents the corrections to the stopping-power ratios at $d_{\max}$ determined as above but using $\bar{E}_0 = 2.33R_{50}$. As above, for beams with scattering foils, an equation can be obtained for the corrections:

$$\left(\frac{\bar{L}}{\rho}\right)_{\text{air}}^{\text{water}} = (0.9932 + A_s D_X) \left[\left(\frac{\bar{L}}{\rho}\right)_{\text{air}}^{\text{water}} \right]_{\bar{E}_0 = 2.33R_{50}} \quad (5.3)$$

where $A_s = 0.0038$ is a numerical constant obtained from the slope of the straight line on fig. 5.17.

The corrections to the stopping-power ratios determined according to the IAEA protocol at reference depth are shown in fig. 5.18. The reference depth is defined[13] at $d_{\max}$ for $\bar{E}_0 < 5$ MeV, $d_{\max}$ or 1 cm for $5 \text{ MeV} < \bar{E}_0 < 10$ MeV, $d_{\max}$ or 2 cm for $10 \text{ MeV} < \bar{E}_0 < 20$ MeV and $d_{\max}$ or 3 cm for $20 \text{ MeV} < \bar{E}_0 < 50$ MeV, and the larger depth should always be chosen. Because the relationship between $\bar{E}_0$ and R_{50} given by the IAEA shown in fig. 5.2 is similar to the AAPM (1983) relationship $\bar{E}_0 = 2.33R_{50}$ in the energy range from 6 to 20 MeV, similar corrections are obtained

for beams with scattering foils:

$$\left(\frac{\bar{L}}{\rho}\right)_{\text{air}}^{\text{water}} = (0.9952 + A_s D_X) \left[\left(\frac{\bar{L}}{\rho}\right)_{\text{air}}^{\text{water}}\right]_{\text{IAEA}} \quad (5.4)$$

where $A_s = 0.0032$ is a numerical constant obtained from the slope of the straight line on fig. 5.18.

For the three different approaches, the slope of the correction curve, or alternatively, the maximum range of corrections, is lowest for the RB-86 method of assigning $\bar{E}_0$ and highest for the AAPM approach using $\bar{E}_0 = 2.33R_{50}$. This is because the RB-86 approach increases the assigned value of $\bar{E}_0$, which lowers the calculated stopping-power ratios using mono-energetic beam data. This makes the cross-over point with the stopping-power ratios calculated with the realistic beams closer to the reference depth and thus leaves a smaller range of necessary corrections.

For scanned beams, corrections to the stopping-power ratios at d_{max} determined using RB-86[86] show a slight increase as the beam energy increases and corrections are less than 0.3% for all beams up to 50 MeV. The corrections to the stopping-power ratios determined according to the AAPM (1983) protocol at d_{max} or the IAEA (1987) code of practice at the reference depth show variations with energy from -0.5% to 0.1%. The largest corrections are needed at high and low energies where the methods of assigning $\bar{E}_0$ are the most inaccurate (see fig 5.2).

For scanned beam accelerators, the beam has very little energy spread and is close to a mono-energetic electron beam with very few contaminant photons[77]. The bremsstrahlung tail is caused mainly by the bremsstrahlung photons generated inside the water phantom as seen from the depth-dose figures (figs. 5.11 to 5.14). Since the RB-86 prescription for determining the mean energy of the electrons at the surface is based on EGS4 Monte Carlo calculations for mono-energetic electron beams, it is not surprising that there is very little correction required in these cases.

5.4.3 Using depth-ionization instead of depth-dose curves

In the AAPM (1983) protocol, I_{50} , the depth at which ionization falls to 50% of its maximum is treated the same as R_{50} , the depth at which the absorbed dose falls to 50% of the dose maximum. Wu *et al.*[89] showed that using the depth-ionization curve is, in principle, incorrect. A relation between I_{50} and R_{50} is given in the IAEA protocol [13, 81] for broad beams at SSD = 100 cm.

To study the differences in I_{50} vs R_{50} , we use our realistic beam simulations, calculated depth-dose curves and calculated stopping-power ratios to calculate depth-ionization curves on the assumption that the ion chamber is a perfect cavity. We also assume that the measured I_{50} values include point of measurement effects (*i.e.* the measured ionization corresponds to that somewhat upstream of the center of a cylindrical chamber, or at the front face of a parallel plate ion chamber). For clinical beams, we find that I_{50} and d_{50} are the same only for lower-energy electron beams. A comparison between our calculated results for realistic clinical beams and values from the IAEA Code of Practice[13] is shown in fig. 5.19. It can be seen that our results of $R_{50} - I_{50}$ vs I_{50} agree with the IAEA's data except when I_{50} is larger than 14 cm. The difference between I_{50} and R_{50} generally increases with the beam energy and can be estimated to within 0.4 mm by the following equation which is extracted from fig. 5.19 for clinical beams with energies from 5 to 25 MeV ($2 < I_{50} < 10$ cm):

$$R_{50} = 1.029I_{50} - 0.063 \quad [\text{cm}] \quad (5.5)$$

$$I_{50} = 0.972R_{50} + 0.061 \quad [\text{cm}] \quad (5.6)$$

where R_{50} and I_{50} are in cm. This analysis assumes point of measurement effects have been included but ignores any change of this or other correction factors with depth, but to first order they do not affect the result.

For electron beams the increase in $\bar{E}_o$ caused by considering R_{50} instead of I_{50} is negligible at 5 MeV and up to 0.6 MeV at 22 MeV when using $\bar{E}_o=2.33R_{50}$ or using RB-86[86]. The increase can amount to 1.7 MeV for a 50 MeV beam. The

range of percent errors in stopping-power ratios, $\delta(SPR)/SPR$, introduced by using I_{50} instead of R_{50} in determining electron beam energy is shown in fig. 5.20. Only a range can be given because the difference varies with depth. The maximum difference in the stopping-power ratios is less than 0.8%. The minimum difference occurs at the surface while the maximum difference occurs at large depths. At d_{max} the difference is less than 0.2% for any clinical beam of energy from 5 to 22 MeV. If one is interested only in the central-axis %DD these differences in the stopping-power ratios have a negligible effect on the shape of the %DD curve although there is a significant difference between the %DD and depth-ionization curves.

5.4.4 Influence of electron energy spread at exit vacuum window

The electron energy at the exit window is mono-energetic in our simulations of clinical beams, but it may have some energy spread in real accelerators. Hence it is important to see whether the energy spread has an effect on our calculated stopping-power ratios. To estimate the effect we use a Gaussian function with a 4.7 MeV full width at half maximum (FWHM) instead of a mono-energetic beam at the exit vacuum window in the simulation of a 20 MeV beam from the Philips SL75-20 accelerator. Such a large energy spread at the exit vacuum window is the worst case that can be expected in a real accelerator[68]. Fig. 5.21 shows the electron energy spectrum at the exit vacuum window and at the phantom surface for a mono-energetic and Gaussian spread beam at the vacuum window respectively. The energy spread at the exit vacuum window leads to a similar electron energy spread at the phantom surface. Although there is a dramatic change in the spectrum of the incident electrons, the changes in the calculated depth-dose curves and the stopping-power ratios are small, as shown in fig. 5.22. There is only a small effect on the depth-dose curve because it is dominated by the electron fluence at a given depth, and this only starts to change at large depths where the low-energy electrons can not reach. Thus

we only start to observe changes there. The explanation is similar for the calculated stopping-power ratios except in this case it is also important that stopping-power ratios vary linearly with energy and thus the average of the stopping-power ratios is the stopping-power ratio of the average energy. There is less than 0.2% change in the calculated stopping-power ratios between these two beams from the phantom surface to the depth where the dose falls to 80% of its maximum. The largest decrease ($\approx 0.8\%$) in the calculated stopping-power ratios occurs where the dose falls to 10% of its maximum. Fig. 5.22 shows that the depth-dose curve calculated for the mono-energetic electron beam at the exit vacuum window is closer to the measured data than that calculated with the broad energy spectrum incident. If we use a Gaussian function with a 5% half width (FWHM) instead of a 20% half width (FWHM) for the energy spread at the exit vacuum window, there is negligible change in both the calculated depth-dose curve and the stopping-power ratios. These results concerning the lack of effect on the depth-dose curve are consistent with the recent results of Deasy *et al.*[101].

5.4.5 Accounting for differences in electron and photon responses of ionization chambers

As done in all previous calculations, all of our calculations ignore the difference in response of ion chambers to electrons and photons, *i.e.* the calculated stopping-power ratios used in eqn 5.1 are based on the assumption that it is a pure electron beam. This ignores the fact that P_{wall} and P_{repl} for the photon and electron components of the beam are different[11]. Specifically, P_{wall} is usually taken as unity in electron beams but is up to 1% different from unity in photon beams and P_{repl} can be up to 6% less than unity in low-energy electron beams but is usually within 1% of unity for photon beams.

To estimate the effect of ignoring these differences on the final dose which is

determined, we write:

$$\left[\left(\frac{\bar{L}}{\rho} \right)_{\text{air}}^{\text{med}} \right]^{\text{tot}} = (1 - \gamma) \left[\left(\frac{\bar{L}}{\rho} \right)_{\text{air}}^{\text{med}} \right]^{\text{el}} + \gamma \left[\left(\frac{\bar{L}}{\rho} \right)_{\text{air}}^{\text{med}} \right]^{\text{ph}} \quad (5.7)$$

where γ is the fraction of ionization in the chamber due to contaminant photons. Under these circumstances the relative error in the dose from ignoring these differences is given by:

$$\frac{\Delta D_{\text{med}}}{D_{\text{med}}} = \gamma \left(1 - \frac{P_{\text{repl}}^{\text{ph}} P_{\text{wall}}^{\text{ph}}}{P_{\text{repl}}^{\text{el}} P_{\text{wall}}^{\text{el}}} \right) \frac{\left[\left(\frac{\bar{L}}{\rho} \right)_{\text{air}}^{\text{med}} \right]^{\text{ph}}}{\left[\left(\frac{\bar{L}}{\rho} \right)_{\text{air}}^{\text{med}} \right]^{\text{tot}}} \quad (5.8)$$

In the worst case and using the estimates above for P_{wall} and P_{repl} , one gets $\Delta D_{\text{med}}/D_{\text{med}} \approx 0.05 \times 0.07 \times 1.2 = 0.4\%$. However, in realistic situations, if $P_{\text{repl}}^{\text{el}}$ is far from unity (*i.e.* lower-energy case) so that the middle term in right hand side of eqn 5.8 is large, then there is little photon contamination, $\gamma \leq 0.01$, and the overall error is small. Conversely, whenever there is significant photon contamination, (high-energy case) so that the γ term is large, then $P_{\text{repl}}^{\text{el}}$ is near unity and thus the middle term makes the overall expression small. Thus the error introduced is usually 0.1% or less.

5.5 Conclusions

We have presented detailed data on stopping-power ratios *vs* depth calculated using realistic clinical beams from a variety of clinical accelerators including the Varian Clinac 2100C, the Philips SL75-20, the AECL Therac 20, the Siemens KD2 and the Scanditronix Medical Microtron 50. Our results are compared with the stopping-power ratios determined using various approximations to determine the incident energy of the "equivalent" mono-energetic beam. The most self-consistent approach is to use the tabulated data from RB-86[86] since it is based on Monte Carlo calculations using the same code. This procedure takes beam divergence effects into account to obtain a more accurate mean energy than the relation $\bar{E}_0 = 2.33R_{50}$ used in the AAPM

TG-21 protocol. The IAEA Code of Practice[13] gives the energy-range relationship in a table, however the values are close to 2.33 in the energy range from 6 to 20 MeV. The differences in the stopping-power ratios between our results using the realistic clinical beams and those using equivalent mono-energetic beams vary with the beam quality. Generally speaking, clinical electron beams can be classified on the basis of overall beam quality into two types: scanned beams and beams with scattering foils.

The scanned-beam accelerators, which use a magnetic field to sweep electrons to achieve a uniform distribution at the surface, produce much cleaner beams. Therefore the correction to the stopping-power ratios is rather small, less than 0.3% at $d_{\max}$ using the RB-86 procedure to assign $\bar{E}_0$ or less than -0.5% at $d_{\max}$ or at the reference depth using either the AAPM TG-21 or IAEA procedures to assign $\bar{E}_0$.

For clinical beams from those accelerators which use scattering foils to achieve a flat profile at the surface, if one uses the RB-86 prescription to assign $\bar{E}_0$, the stopping-power ratios are under-estimated by up to 1.5% in regions near $d_{\max}$ and overestimated by up to 1.8% at large depths, depending on the beam energy. Our calculated results are in reasonable agreement with the calculations by Andreo *et al.*[18, 69] for less sophisticated models of beams near 10 MeV although in their particular case they were dealing with a beam in which the photon contamination plays little role and the effects of electron energy spread are negligible at the reference depth. A detailed analysis shows that the changes in the stopping-power ratios between the realistic clinical beams and those determined using calculations for mono-energetic beams and various methods for assigning the "equivalent" value of $\bar{E}_0$, are mainly caused by the electron energy spread and contaminant photons. The increase in the stopping-power ratios due to the contaminant photons in the electron beams is about 0.1% for scanned beams of all energies and is up to 0.6% at 20 MeV for beams with scattering foils. These values are not as large as those estimated by Klevenhagen [99], but they confirm his major conclusion that these effects should be taken into account. As indicated at 10 MeV by Andreo *et al.*'s results, for all the beams studied here it

has been shown that the contribution to the changes in the stopping-power ratios due to the electron angular distributions is small except at the surface.

We find that even a very substantial Gaussian spread (in fact any symmetric distribution with the same mean energy) in the energy spectrum of the electrons leaving the accelerator vacuum has very little effect on the calculated depth-dose or stopping-power ratio *vs* depth curves, despite directly affecting the energy spectrum of the beam on the patient plane. It is the asymmetric low-energy side of the main peak in the realistic electron beam's spectrum which has the most significant effect on the depth-dose and stopping-power ratio *vs* depth curves.

To make this study useful in clinical practice for reference dosimetry, we developed simple equations (equations 5.2,5.3,5.4) which can be used to correct the stopping-power ratios (at d_{max} or the IAEA reference depth) which were determined using any of three procedures for assigning the mean energy at the surface, $\bar{E}_0$. The equations apply to accelerators with scattering foils and are all parameterized in terms of D_r , the percentage dose in the bremsstrahlung tail. The corrections needed for scanned beams are smaller than for scatter-foil beams, especially if the RB-86 procedure is used to assign $\bar{E}_0$.

It is known that the mean energies determined according to Rogers and Bielajew [86] and $\bar{E}_0 = 2.33R_{50}$ differ by about 0.4 MeV for electron beams of energy from 5 to 20 MeV. The difference increases for higher energy beams and it can be up to 6.4 MeV (13%) for a 50 MeV beam. The energy difference causes 1% and 3% changes in selected stopping-power ratios for electron beams of energy 5 and 50 MeV respectively. We suggest that the mean energy should be determined using the procedure of RB-86[86] since it is inherently more accurate. A related study on the relationship between $\bar{E}_0$ and R_{50} using realistic beams confirms this[73]. However, it must be emphasized that using the corrections presented here, the final stopping-power ratio determined is independent of which prescription for $\bar{E}_0$ is used in the clinic.

We have studied the difference between using the depth of 50% dose level, R_{50} ,

and the depth of 50% ionization level, I_{50} , for specifying the incident energy of clinical electron beams. The difference, $R_{50} - I_{50}$, increases with the beam energy and can be estimated by a linear equation. Correcting the depth I_{50} to R_{50} has a very small effect on $\bar{E}_0$ for low-energy beams but increases the beam energy $\bar{E}_0$ by up to 3% for a 50 MeV clinical beam. The maximum change in selected stopping-power ratios introduced by using the depth of I_{50} instead of the depth R_{50} is negligible for 5 MeV beams and up to 0.8% for 50 MeV beams. However, at $d_{\max}$ for clinical beams between 5 and 22 MeV the difference is less than 0.2%.

Table 5.1: Definitions of different stopping-power ratio calculations presented in figures 5.5 to 5.14

Symbol	Caption	type of stopping-power ratio	energy dist'n	angular dist'n
—	$\bar{E}_0 = 10 \text{ MeV}$ (RB-86)	based on mono-energetic spr data(TG21) & $\bar{E}_0$ from RB-86	no	no
---	e^- and photon	complete, realistic beam incident	yes	yes
•	e^- only	electrons only from realistic beam incident	yes	yes
.....	photon only	photons only from realistic beam incident	yes	yes
▲	e^- spectrum	correct electron spectrum but from a point source	yes	no
--- --	$\bar{E}_0 = 10 \text{ MeV}$ (2.33R ₅₀)	based on mono-energetic spr data(TG21) & $\bar{E}_0 = 2.33R_{50}$	no	no

Figure 5.1: The calculated spectra of electrons and photons in 9 and 18 MeV electron beams inside a 10×10 cm² field of a Clinac 2100C accelerator. The ratios of total contaminant photon fluence to total electron fluence inside the beam field are 1.4 and 1.7 for 9 and 18 MeV beams respectively. Spectra calculated with BEAM[77].

Figure 5.2: A comparison of $\bar{E}_0/R_{50}$ versus R_{50} obtained from the AAPM (1983), IAEA (1987) protocol and RB-86[86].

Figure 5.3: The differences in mono-energetic water/air stopping-power ratios determined for energies obtained using $\bar{E}_o = 2.33R_{50}$ (AAPM) or the RB-86 prescription for three R_{50} values which are typical of clinical electron beams near 5, 12 and 50 MeV. The stopping-power ratios are from the linear interpolation of values given in the AAPM protocol.

Figure 5.4: For 9 and 18 MeV electron beams, comparison of the water/air stopping-power ratios calculated using code SPRRZ with a mono-energetic parallel or a point source beam and those from the AAPM protocol for broad parallel beams. The incident beams used in our calculations have finite size $10 \times 10 \text{ cm}^2$. The calculated stopping-power ratios are for the beam near the central-axis.

Figure 5.5: (a) Calculated relative depth-dose curves for a 20 MeV electron beam from the Philips SL75-20. (b) Comparison of the water/air stopping-power ratios calculated in different ways as described in table 5.1.

Figure 5.6: (a) Calculated relative depth-dose curves for a 10 MeV electron beam from the Philips SL75-20; (b) Comparison of the water/air stopping-power ratios calculated in different ways as described in table 5.1.

Figure 5.7: (a) Calculated relative depth-dose curves for an 18 MeV electron beam from the Varian Clinac 2100C at University of Wisconsin; (b) Comparison of the water/air stopping-power ratios calculated in different ways as described in table 5.1.

Figure 5.8: (a) Calculated relative depth-dose curves for a 9 MeV electron beam from the Varian Clinac 2100C at University of Wisconsin; (b) Comparison of the water/air stopping-power ratios calculated in different ways as described in table 5.1.

Figure 5.9: (a) Calculated relative depth-dose curves for a 21 MeV electron beam from the Siemens KD2; (b) Comparison of the water/air stopping-power ratios calculated in different ways as described in table 5.1.

Figure 5.10: (a) Calculated relative depth-dose curves for an 11 MeV electron beam from the Siemens KD2 at the General Hospital in Ottawa; (b) Comparison of the water/air stopping-power ratios calculated in different ways as described in table 5.1.

Figure 5.11: (a) Calculated relative depth-dose curves for a 20 MeV electron beam from the Therac 20 at the Ottawa Civic Hospital; (b) Comparison of the water/air stopping-power ratios calculated in different ways as described in table 5.1.

Figure 5.12: (a) Calculated relative depth-dose curves for a 9 MeV electron beam from the Therac 20 at the Ottawa Civic Hospital; (b) Comparison of the water/air stopping-power ratios calculated in different ways as described in table 5.1.

Figure 5.13: (a) Calculated relative depth-dose curves for a 50 MeV electron beam from the Racetrack MM50 at Memorial Sloan-Kettering Cancer Center; (b) Comparison of the water/air stopping-power ratios calculated in different ways as described in table 5.1.

Figure 5.14: (a) Calculated relative depth-dose curves for a 25 MeV electron beam from the Racetrack MM50 at Memorial Sloan-Kettering Cancer Center; (b) Comparison of the water/air stopping-power ratios calculated in different ways as described in table 5.1.

Figure 5.15: The percentage increase in stopping-power ratios due to the contaminant photons in clinical electron beams versus bremsstrahlung tail D_x . The electron beam energies shown above are 6, 9, 12, 15 and 18 MeV for Clinac 2100C, 5, 10, 14, 17, 20 MeV for Philips SL75-20, 11 and 21 MeV for Siemens KD2, 9 and 20 MeV for Therac 20 and 25, 40 and 50 MeV for Racetrack MM50.

Figure 5.16: The percent corrections at $d_{\max}$ to the Spencer-Attix water to air stopping-power ratios determined using the mono-energetic stopping-power ratio data in the AAPM TG21 protocol and assigning the incident mean energy using the prescription of RB-86[86]. The correction is given as a function of the bremsstrahlung tail, D_x , for the same clinical electron beams as in the previous figure.

Figure 5.17: As in fig. 5.16 but with the incident beam energy assigned using $\bar{E}_0 = 2.33R_{50}$. These are the corrections needed if the TG-21 protocol of the AAPM were used.

Figure 5.18: As in fig. 5.16 but with the incident beam energy assigned using the tabulated relationship between $\bar{E}_0$ and R_{50} found in the IAEA Code of Practice and for the stopping-power ratio at the IAEA's reference depths rather than d_{max} (see the text). Those beams for which d_{max} and d_{ref} are different are shown with a circle around the symbol.

Figure 5.19: The differences between the depth R_{50} , at which the depth-dose curve falls to 50% of dose maximum and the depth I_{50} , at which the depth-ionization curve falls to 50% of its maximum for clinical electron beams of energy from 5 to 50 MeV. For comparison the values provided in the IAEA Code of Practice are shown (for $I_{50}=18.2$ cm the corresponding value is 1.5).

Figure 5.20: The range of percent change, $\delta(\text{SPR})/\text{SPR}$, in the stopping-power ratios introduced by using the depth of 50% ionization level, I_{50} , instead of the depth of 50% dose level, R_{50} , for clinical electron beams of energy from 5 to 50 MeV. The percent change, $\delta(\text{SPR})/\text{SPR}$, is defined as $[(\text{SPR})_{I_{50}} - (\text{SPR})_{R_{50}}]/(\text{SPR})_{R_{50}}$.

Figure 5.21: (a) Electron energy spectra at the exit vacuum window. The incident energy of mono-energetic electrons is 23 MeV. The source with a 20% energy spread has a 4.7 MeV full width at half maximum(FWHM). (b) Electron energy spectra of the SL75-20 accelerator at the phantom surface. The relative number of electrons (normalized to a peak of 1) is counted within the $10 \times 10 \text{ cm}^2$ field. $AE = 0.521$ and $ECUT = 0.700 \text{ MeV}$.

Figure 5.22: (a) Comparison of the % depth-dose curve calculated using the beam simulated with the mono-energetic source and that simulated with the source of 20% energy spread at the exit vacuum widow of the accelerator SL75-20. The measured data are from Udale[31]. (b) Comparison of the water/air stopping-power ratios calculated using the two clinical beams simulated with the two energy sources at the exit vacuum windows.

Chapter 6

**Electron fluence correction factors
used in conversion of dose
measured in plastic to dose in
water**

In radiation dosimetry protocols[11, 12, 13, 81] plastic is recommended besides water as a phantom material for the determination of absorbed dose to water in high energy electron beams. According to protocols when the measurements are performed using an ion chamber, the dose measured in a plastic phantom has to be converted to the dose to water in a water phantom for reference dosimetry.

To make the dose conversion from a plastic to a water phantom, two considerations are involved. The first is to find corresponding depths in each phantom such that the electron spectra are the same so that various conversions hold (discussed below). As shown in chapter 4, there is a simple scaling of depths which causes the mean energies, and by assumption the entire spectra, to be the same at corresponding depths (also discussed below). However, because of the differences in scattering and stopping powers of the different materials, and because the scaled depths are at different distances from the virtual source (so there are inverse square effects), there is still an overall difference at the scaled depths in the total electron fluence, as opposed to fluence spectra. This difference in fluence is corrected by ϕ_p^w , the electron fluence correction factor, which is the ratio of the total electron fluence in the water to that at the scaled depths in a plastic phantom.

In the AAPM TG-25 protocol when plastic phantoms are used in dose measurements for electron beams, the dose to water in a water phantom, D_w^w , is related to the dose to plastic in a plastic phantom, D_p^p , by

$$D_w^w(d^w) = D_p^p(d^p) \left(\frac{\bar{S}}{\rho} \right)_p^w \phi_p^w \quad (6.1)$$

where $\left(\frac{\bar{S}}{\rho} \right)_p^w$ is the ratio of the mean unrestricted mass collision stopping power in water to that in the solid, ϕ_p^w is the electron fluence correction factor and the depth in water, d^w , is related to the depth in plastic, d^p , by a depth-scaling factor called the effective density ρ_{eff} :

$$d^w = d^p \rho_{eff} = d^p \frac{R_{50}^w}{R_{50}^p} \quad (6.2)$$

where R_{50}^w and R_{50}^p are depths at which the absorbed dose level falls to 50% of its

maximum in water and plastic respectively. In this paper, D_B^A refers to the dose to medium B in a phantom of medium A . For example D_a^w means the dose to a small air cavity in a water phantom. The AAPM TG-25 protocol provides recommended values of the effective densities and $\left(\frac{\bar{S}}{\rho}\right)_{\text{polystyrene}}^w = 1.03$. The AAPM also recommends that the mass density of each solid phantom be measured and that the listed data be corrected for each phantom accordingly.

According to Spencer-Attix cavity theory, the dose to water in a water phantom and the dose to plastic in a plastic phantom are given by.

$$D_w^w(d^w) = M^w N_{gas} \left(\frac{\bar{L}}{\rho}\right)_a^w (P_{ion} P_{repl} P_{wall})^w \quad (6.3)$$

$$D_p^p(d^p) = M^p N_{gas} \left(\frac{\bar{L}}{\rho}\right)_a^p (P_{ion} P_{repl} P_{wall})^p \quad (6.4)$$

where M^w and M^p are ion chamber meter readings measured in the water and plastic phantom respectively, d^w and d^p are the depths of the measurements in water and in plastics respectively, $\left(\frac{\bar{L}}{\rho}\right)_a^{\text{med}}$ is the ratio of the mean restricted mass collision stopping power in med to that in air and the P corrections are defined by the AAPM TG-21 protocol and are in principle dependent on the medium of the phantom.

If the spectra (or the mean energy) are the same at a depth of d^w in water and at a corresponding depth of d^p in plastic but electron fluences are different then:

$$D_w^w(d^w) = D_p^p(d^p) \left(\frac{\bar{L}}{\rho}\right)_p^w \phi_p^w \quad (6.5)$$

On the right hand side, the product $D_p^p(d^p) \left(\frac{\bar{L}}{\rho}\right)_p^w$ (where the stopping-power ratio is averaged over the electron spectrum in the plastic phantom) gives, according to Spencer-Attix cavity theory, D_w^w , the dose in the plastic phantom to a small volume of water. Since these are at scaled depths, $D_w^w \phi_p^w$ just gives the dose in a water phantom to water, i.e. the left hand side of the equation because the dose to water is just proportional to the total fluence in each phantom and ϕ_p^w gives the ratio of these

fluences. In order to meet the requirement that the mean energies are identical at corresponding depths, d^w is related to d^p by eqn 6.2 [12, 73]. Eqn 6.5 defines how the fluence correction factors are calculated in this study. Note that ϕ_p^w is independent of the characteristics of any measuring device.

In practice we need the dose to water in a water phantom from an ion chamber reading measured in a plastic phantom. Combine eqns 6.3 to 6.5 and assuming

$$\left(\frac{\bar{L}}{\rho}\right)_p^w \times \left(\frac{\bar{L}}{\rho}\right)_a^p = \left(\frac{\bar{L}}{\rho}\right)_a^w \quad (6.6)$$

since at the scaled depths we have assumed the spectra are identical in the water and plastic phantoms, gives :

$$D_w^w(d^w) = M^p(d^p) N_{gas} \left(\frac{\bar{L}}{\rho}\right)_a^w (P_{ion} P_{repl} P_{wall})^p \phi_p^w \quad (6.7)$$

where $M^p(d^p)$ is the ion chamber reading measured in a plastic phantom at depth d^p corresponding to the scaled depth in water at which the dose is desired and D_w^w is the dose to water in a water phantom. Eqn 6.7 can be used to obtain the dose to water in a water phantom directly from the measurement performed in plastics phantoms. This result is very similar to a proposal of Andreo *et al.*[102] although we have not introduced a Bragg-Gray stopping-power ratio into the derivation and eqn 6.7 explicitly recognizes the need for P_{repl} and P_{wall} in the plastic phantom. The distinct advantage of the approach is that only stopping-power ratio for water to air are required and values for plastics to air are not necessary.

The IAEA Code of Practice[13] has a slightly different definition of electron fluence correction factors from the AAPM TG-25 protocol. The IAEA Code of Practice converts the electrometer reading at the ionization maximum in a plastic phantom to the equivalent reading at the ionization maximum in a water phantom using:

$$M_u^w = M_u^p h_m \quad (6.8)$$

where M_u^w and M_u^p are the electrometer reading monitor unit in water and plastic corrected for P_{ion} respectively.

From its definition, in principle h_m is a chamber dependent factor although to minimize this chamber dependence, the IAEA defined h_m in terms of measurements with a plane parallel chamber since it was believed $P_{\text{repl}} = P_{\text{wall}} = 1.00$ for these chambers although this is not necessarily so[103]. Furthermore many measurements have been done with Farmer chambers.

Conceptually the IAEA procedure is different from the AAPM TG-21 and TG-25 procedures because it relates measurements at ionization maxima, and these values do not necessarily apply at the reference depth. Fig. 6.1 shows a comparison of calculated central-axis depth-dose curves and central-axis depth-ionization curves for a 12 MeV electron beam from a Clinac 2100C where the depths in plastic have been scaled to the water equivalent depth using the depth-scaling factors from chapter 4. The depth-dose curves are calculated using a simulated realistic incident beam and the depth-ionization curves are obtained by converting the depth-dose curves using the calculated Spencer-Attix water/air and plastic/air stopping-power ratios of the same incident beam. The depth of dose maximum in water is the same as the scaled depth of dose maximum in plastic, while the depth of maximum ionization in water is very different from the scaled depth of maximum ionization in plastic. Furthermore, the depths of ionization and dose maximum are at different locations.

The IAEA recommends that the depths in plastic and in water are related by:

$$d^p = d^w \frac{(r_o/\rho)_{\text{plastic}}}{(r_o/\rho)_{\text{water}}} \quad (6.9)$$

where r_o/ρ is the linear continuous-slowng-down range tabulated as a function of energy by the ICRU[82]. However, we have shown that the values of depth-scaling factors obtained by using the ratio of continuous-slowng-down ranges are inaccurate and overestimate our calculated values by 1 – 2% in all cases (see chapter 4).

The absorbed dose to water in a water phantom, D_w^w , at the effective point of measurement using an ion chamber is given by following equation according to the

IAEA[13]:

$$D_w^w(P_{eff}) = M_u^w N_{gas} \left(\frac{\bar{L}}{\rho} \right)_{air}^w (P_{wall} P_{fl})^w \quad (6.10)$$

where $\left(\frac{\bar{L}}{\rho} \right)_{air}^w$ is written $(s_{w,air})_u$ in the IAEA notation, N_{gas} is N_D in the IAEA and $P_{fl} P_{wall}$ are denoted p_u in the IAEA notation. This equation differs from equation 6.3 since P_{eff} is an effective point of measurement that takes into account the spatial extent of the air cavity by locating the point of measurement P_{eff} , in front of the chamber center to correct for the gradient of fluence within the chamber cavity and thus $P_{repl} = P_{gr} P_{fl}$ in equation 6.3 but is only P_{fl} in this case, i.e. the correction for perturbation of the electron fluence by the cavity.

Using eqns 6.10 and 6.8 gives:

$$h_m = \frac{D_w^w(z_{max}^w) / \left(\frac{\bar{L}}{\rho} \right)_{air}^w (P_{fl} P_{wall})^w}{D_p^p(z_{max}^p) / \left(\frac{\bar{L}}{\rho} \right)_{air}^p (P_{fl} P_{wall})^p} \quad (6.11)$$

where z_{max}^w and z_{max}^p are the depths of maximum ionization in water and plastic phantoms respectively.

It can be seen that the definition of the fluence correction factor is different between the AAPM definition, eqn 6.5, and the IAEA Code of Practice (eqn 6.8) unless:

$$\left(\frac{\bar{L}}{\rho} \right)_p^w = \frac{\left(\frac{\bar{L}}{\rho} \right)_{air}^w (P_{fl} P_{wall})^w}{\left(\frac{\bar{L}}{\rho} \right)_{air}^p (P_{fl} P_{wall})^p} \quad (6.12)$$

$$z_{max}^w = d_{max}^w \quad (6.13)$$

$$z_{max}^p = d_{max}^w / \rho_{eff} \quad (6.14)$$

where d_{max}^w is the dose maximum in the water phantom and the ρ_{eff} is depth-scaling factor from plastic to water for the incident beam. These conditions require that the ion chamber corrections in the depth of water phantom be the same as those in plastic phantom and that the depth of maximum dose equal the depth of maximum ionization in a water phantom and that the depth scaling factors to convert the depth in plastic to the equivalent depth in water be the same for the AAPM and the IAEA.

These conditions cannot be satisfied in principle and for some beams as shown in fig. 6.1, the difference between the depth of maximum dose and the depth of maximum ionization may be quite large.

The IAEA[13] recommended values of h_m are 1.016 ± 0.006 , 1.006 ± 0.004 and 1.000 ± 0.003 for clear polystyrene, white polystyrene and PMMA phantom materials respectively. The value of h_m is independent of $\bar{E}_0$ between 2 and 10 MeV. These values are based on experimental data.

There are considerable discrepancies in the recommended values of the fluence correction factors between the two protocols when polystyrene is used as the phantom material. Furthermore, there exist significant variations in values of fluence correction factors measured by various authors[104, 105, 106, 107, 108, 109]. The factor appears to vary with the type of polystyrene, with the type of ion chamber and with the energy of incident electron beams.

Similarly, Monte Carlo calculations with mono-energetic incident beams have been reported to be in disagreement with experiment[110] and we have confirmed this for a variety of energies and materials.

With the simulation of realistic clinical beams now possible (see chapter 2) [77], we are able to perform the investigation under "ideal" experimental conditions. We can calculate the electron fluence correction factors using the Monte Carlo method for different plastic phantom materials and resolve these discrepancies. In this investigation, monoenergetic incident electron beams are used for dose calculations as well as the realistic clinical beams with an energy range of 5 – 50 MeV for a variety of different clinical accelerators including the Varian Clinac 2100C, the Philips SL75-20, the Siemens KD2, the Therac 20 and the Scanditronix Medical Microtron 50.

6.1 Methods of calculations

The mono-energetic beams used in this study are point source incident on a phantom with a $10 \times 10 \text{ cm}^2$ field and $\text{SSD} = 100 \text{ cm}$, while the realistic clinical electron beams used in this study are generated by a full simulation of the accelerator treatment head (see chapter 2). The incident beams are used as an input to the code DOSXYZ to calculate the depth-curves in different phantom materials. The central-axis depth-dose curves in different phantoms are calculated separately but the incident beams are identical.

6.1.1 The incident clinical beams

The simulations to obtain realistic clinical beams have been described in detail in chapters 2 and 3. In the beam simulation the Monte Carlo parameters are set to $\text{ECUT} = 0.700 \text{ MeV}$, $\text{AE} = 0.521 \text{ MeV}$ and $\text{PCUT} = \text{AP} = 0.010 \text{ MeV}$. In all cases, the PRESTA electron algorithm[53] is employed with $\text{ESTEPE} = 0.01$ (although the default values of ESTEPE would be adequate for this application).

The clinical electron beam field size is $10 \times 10 \text{ cm}^2$, $25 \times 25 \text{ cm}^2$ and $20 \times 20 \text{ cm}^2$ for beam energies 5 – 21 MeV, 25 – 40 MeV and 50 MeV respectively. The SSD is 100 cm and 110 cm for energies 5 – 40 MeV and 50 MeV respectively. In the beam simulation, the electron's incident energy at the exit vacuum window of an accelerator is adjusted to match the R_{50} between the measured and Monte Carlo calculated depth-dose curves.

6.1.2 Conversion of dose from plastics to water

Throughout this work, we first calculate central-axis dose curves in different phantom materials with an identical incident beam as shown for two beams in fig. 6.2. Then the

fluence correction factor ϕ_p^w is calculated by using eqn.6.5 where $\left(\frac{\bar{L}}{\rho}\right)_p^w$ is calculated by using the realistic incident beams. To calculate the stopping-power ratios needed here, we have used the code SPRRZ as discussed in chapter 5. We have used calculated depth-scaling factors according to eqn 6.2. Our values of depth-scaling factors are taken from chapter 4.

As discussed in the introduction, to derive equation 6.7, we had to assume equation 6.6 holds (viz $\left(\frac{\bar{L}}{\rho}\right)_p^w \times \left(\frac{\bar{L}}{\rho}\right)_a^p = \left(\frac{\bar{L}}{\rho}\right)_a^w$). To check the validity of eqn 6.6 we have calculated stopping-power ratios using a realistic 18 MeV electron beam from a Clinac 2100C incident on plastic phantoms of clear or white polystyrene and PMMA to calculate $\left(\frac{\bar{L}}{\rho}\right)_a^p$ at various water equivalent depths determined according to eqn 6.2 and values of depth-scaling factors from chapter 4 and the same beam incident on a water phantom to calculate $\left(\frac{\bar{L}}{\rho}\right)_p^w$ and $\left(\frac{\bar{L}}{\rho}\right)_a^w$. Fig. 6.3 compares $\left(\frac{\bar{L}}{\rho}\right)_p^w$ to $\left(\frac{\bar{L}}{\rho}\right)_a^w / \left(\frac{\bar{L}}{\rho}\right)_a^p$ and shows that they agree within 0.2% up to the depth of R_{50} , indicating that the depth scaling procedure is giving equivalent depths at which the electron spectra and hence the stopping-power ratios are close to identical.

6.2 Results of calculations

6.2.1 Comparison with values recommended by the AAPM

The AAPM TG-25 protocol provides tabulated values for electron fluence correction factors as a function of depth and energy[12]. The recommended values are based on a theoretical analysis of the Fermi-Eyges theory by Hogstrom and Almond (1982)[12]. Since their theoretical calculation employs assumptions that overestimate angulation in the descending portion of the depth-dose curve, it is recommended in AAPM TG-25 that the fluence ratio at R_{80} be used on the remaining, descending portion of the curve. The AAPM recommends that the most probable incident energy be used for selecting the proper data from the tables. Throughout this work, the values of the

AAPM TG-25 protocol are obtained by interpolating the tabulated data according to the most probable incident energy $E_{0,p}$ which are obtained from the energy spectra (see chapter 3) and which are known to be in good accord with the values obtained by applying the standard energy-range relationship from the ICRU report 35 [82, 73].

Fig. 6.4 presents electron fluence correction factors calculated according to equ 6.5 for a 10 MeV beam from an SL75-20 accelerator. The characteristics of incident beams are all listed in table 6.1. The fluence correction factors are not unity at the surface for clear and white polystyrene materials. This is caused by the lack of back scattering from these two plastics relative to water which is consistent with the experimental results discussed by Klevenhagen[111]. We have studied fluence correction factors with a beam incident on very thin slab phantom and fluence correction factors are found to be unity in this case. The calculated correction factors in the descending region of the central-axis depth-dose curve are very sensitive to the depth-scaling factors (see the lower panels of fig.6.4 and 6.2). The dip in the calculated correction factors in this depth interval depends strongly on the shape of the descending portion of calculated central-axis depth-dose curves in phantoms. In the region of the bremsstrahlung tail the calculated fluence correction factor is close to the value of $\left(\frac{\bar{\mu}_{en}}{\rho}\right)_p^w / \left(\frac{\bar{L}}{\rho}\right)_p^w$ (where $\left(\frac{\bar{\mu}_{en}}{\rho}\right)_p^w$ is the ratio of the average mass energy-absorption coefficient for water to that for plastic) if the spectral distribution and fluence of photons at the depth of interest in a water phantom are the same as at the water-equivalent depth in a plastic phantom. This value is close to unity and hence the calculated value of ϕ_p^w drops at depths in the bremsstrahlung tail as seen in the lower panel of fig.6.4.

Fig. 6.5 and 6.6 show calculated electron fluence correction factors ϕ_p^w as a function of depth for 6 and 20 MeV monoenergetic incident beams and the values determined according to the AAPM TG-25 protocol. Even in this case, where the initial assumption of a monoenergetic beam is common, the calculations show differences of more than 2% in some cases although they show similar trends. The more accurate Monte

Carlo results tend to be larger in the build up region. Some, but certainly not all of the difference comes from the inclusion of differences in backscatter in the Monte Carlo calculations. At the surface of the phantom the correction factors, ϕ_p^w , vary with the incident energy and the type of plastics. The variation of ϕ_p^w with incident beam energy at the surface of the phantom is more than 1% even for the same type of plastic. This reflects that the backscatter is clearly energy dependent since the incident beam energies are monoenergetic. It also indicates that fluence contributed by the backscatter at the phantom surface not only changes in magnitude but also in sign with beam energies.

Fig. 6.7 to 6.9 show calculated electron fluence correction factors ϕ_p^w as a function of depth for incident electron beams of 6 to 18 MeV from a Clinac 2100C accelerator and those determined according to the AAPM TG-25 protocol.

Fig. 6.10 shows calculated electron fluence correction factors ϕ_p^w as a function of depth for 5 and 20 MeV beams from an SL75-20 accelerator and those determined according to the AAPM protocol. $E_{0,p}$ is 5.66 MeV for the 5 MeV beam from an SL75-20. Since the fluence correction factors are not listed below 6 MeV in the protocol, the curves for 6 MeV from the AAPM TG-25 protocol are plotted for reference only.

Fig. 6.11 shows calculated electron fluence correction factors ϕ_p^w as a function of depth for 6 and 11 MeV beams from a KD2 accelerator and those determined according to the AAPM protocol.

Fig. 6.12 shows calculated electron fluence correction factors ϕ_p^w as a function of depth for 6 and 20 MeV beams from a Therac 20 accelerator and those determined according to AAPM protocol.

Fig. 6.13 shows calculated electron fluence correction factors ϕ_p^w as a function of depth for 25 and 50 MeV beams from an MM50 accelerator.

It can be seen (fig. 6.7 to 6.13) that the calculated electron fluence correction factors ϕ_p^w as a function of depth for clinical incident beams are similar if incident

nominal beam energies are similar. There are more electron fluence (about 0.8%) at the surface due to the backscatter in water phantom than those at the surface of clear or white polystyrene phantoms for almost all incident clinical incident beams. This may be caused by the common property that all clinical beams have energy spectra which contain low-energy electrons with large incident angles.

6.2.2 Comparison with experimental data

The published experimental data show considerable variation and a single set of values cannot be given. The electron fluence correction factor depends on various measurement conditions. As indicated in eqn 6.11, h_m depends on the type of plastics, type of ion chamber, and quality of electron beams. Therefore the experimental variation is expected. Table 6.2 and 6.3 summarize experimental results from various authors.

When comparing our calculated correction factors with measured values one should bear in mind that the measured data are mainly for the IAEA's h_m while our calculated values are for ϕ_p^w based on eqn 6.5. As discussed in the introduction, h_m is different from ϕ_p^w . In addition, h_m is measured at the depth of maximum ionization in each material whereas ϕ_p^w is calculated at scaled depths which are related to depth-dose curves. Usually the depth of maximum dose is larger than the depth of maximum ionization and depths of maximum ionization may not be at the corresponding scaled depths. We have calculated h_m for a 12 MeV beam from a Clinac 2100C by converting the depth-dose curves in plastic and water phantoms into the ionization curves using calculated stopping-power ratios for the realistic incident beam as shown in fig.6.1. The results indicate that for this beam the value of h_m is 0.5% smaller than ϕ_p^w at $d_{\max}$ for a clear polystyrene phantom. Furthermore if we chose the ionization ratio of water to plastic at the depth of $I_{\max}^p$, the water equivalent depth of ionization maximum in polystyrene, shown in fig.6.1 the value of h_m becomes 1.019 which is 1.5%

smaller than ϕ_p^w at $d_{\max}$ for a clear polystyrene phantom. This actually explains why measured values of h_m are smaller than our calculated ϕ_p^w .

Fig. 6.14 to 6.16 present calculated fluence correction factors, ϕ_p^w , at $d_{\max}$ for various incident clinical electron beams along with existing experimental data, mostly for h_m . The calculated fluence correction factors at the depth of $0.6R_{50} - 0.1$ are also shown for three plastic phantoms. The reason we show values of correction factors at the depth of $0.6R_{50} - 0.1$ is because the calculated water/air stopping-power ratios are independent of beam quality at this depth, hence this depth may be used as a reference depth.

It can be seen from fig. 6.14 that our calculated values of ϕ_p^w are in accord with those experimental values of ϕ_p^w (Kase *et al.*[106]) but are larger than those experimental values of h_m (Thwaites [108]).

Similarly, it is not surprising that our calculated values of ϕ_p^w are systematically larger than the measured values for the white polystyrene phantom shown in fig. 6.15 because all measured values are h_m [105, 109, 104].

We have used a linear equation to fit the calculated electron fluence correction factors ϕ_p^w as a function of R_{50} in water at $d_{\max}$ as well as at the depth of $0.6R_{50} - 0.1$ for any clinical beam with an energy between 5 and 25 MeV. The fitted simple equation is:

$$\phi_p^w = a_p - b_p R_{50}^{water} \quad (6.15)$$

where a_p and b_p are constants for plastic p and the values are given in table 6.4 for various phantom materials.

6.3 Discussion

We calculate electron fluence correction factors according to eqn 6.5 instead of eqn 6.1 because it is fundamentally more accurate, it is based on a correction factor ϕ_p^w which

is conceptually simple and because the stopping power ratios are electron energy dependent as well as depth dependent rather than constant as recommended by the AAPM for use with 6.1. Figs. 6.17 and 6.18 show the calculated Spencer-Attix water to plastic stopping-power ratios for various incident electron beams for 3 plastics. There are up to 1% variation depending on the incident beam energies or depths.

In this work we do not calculate h_m . The reason is simple: measured values of h_m are ion chamber dependent. The proposed approach which is based on eqn 6.5 and leads to 6.7 for the dose to water seems much less ambiguous. Eqn 6.11 also explains why measured values of h_m vary with the type of chamber used in the experiment even with same incident beams.

This study shows that the beam quality has little influence on the shape of the fluence correction vs depth curves, for example, a calculated fluence correction curve for a 6 MeV beam from the Therac 20 is similar to that for a 6 MeV from the Clinac 2100C or the KD2. However at higher beam energies the "cleaner" beams (such as the 20 MeV beam from the Therac 20) have larger values of d_{max} and therefore their corrections are larger at d_{max} . Hence the values of electron fluence correction factors are indirectly influenced by the beam quality. Previous independent Monte Carlo calculations have shown that at the depth of the absorbed dose maximum fluence correction factors for polystyrene are higher than experimental values. This is mainly due to the fact that d_{max} of a monoenergetic beam is generally larger than d_{max} of a clinical beam of similar energy.

fig. ??.

6.4 Conclusions

We have presented detailed data on electron fluence correction factors vs depth, calculated using realistic clinical beams from a variety of clinical accelerators including

the Varian Clinac 2100C, the Philips SL75-20, the AECL Therac 20, the Siemens KD2 and the Scanditronix Medical Microtron 50 incident on clear polystyrene, white polystyrene, and PMMA phantoms. Our results are compared with the values determined according to the AAPM TG-25 protocol. The differences in ϕ_p^w between our values and those recommended by the AAPM vary with the type of phantom materials as well as the depth in the phantom. Generally speaking, for beams incident on polystyrene phantoms values determined according to the AAPM underestimate ϕ_p^w by up to 3% at $d_{\max}$ depending on the incident beam energy. For beams incident on PMMA phantoms, values determined according to the AAPM only underestimate ϕ_p^w by up to 1% at $d_{\max}$ depending on the incident beam energy. At the surface of the phantom the calculated electron fluence correction factors are not unity due to the lack of backscatter for polystyrene materials compared to water. The theoretical analysis of the Fermi-Eyges theory used in the AAPM protocol cannot account for this effect. In all cases, values determined according to the AAPM overestimate ϕ_p^w near or beyond R_{80} , which reflects the limitation of the application of the Fermi-Eyges theory.

Our results are compared with the existing measured data available. Despite the large variation in the measured values our calculated values of electron fluence correction factor ϕ_p^w generally predict the trend of the measurements and explain the measured values. Since most measured values are for h_m which are ion chamber dependent by definition, the large variations in the measured values are to be expected. We have also demonstrated why the measured values of h_m are smaller than our calculated values of ϕ_p^w .

To make this study useful in clinical practice for reference dosimetry we have developed a simple equation of ϕ_p^w as a function of R_{50} (eqn 6.15). It can be used to obtain the electron fluence correction factors at $d_{\max}$ or at a depth of $0.6R_{50} - 0.1$ cm. Eqn 6.7 can then be used to determine dose-to-water in a water phantom directly from the chamber readings measured in a plastic phantom for any clinical electron

beam with energy of 5 – 25 MeV incident on clear polystyrene, white polystyrene and PMMA phantoms.

Table 6.1: Characteristics of various incident electron beams shown in figures.

Beam	$E_{nominal}$ MeV	$E_{o,p}$ MeV	d_{max} cm	R_{80} cm	R_{50} cm	$0.6R_{50}-0.1$ cm
Monoenergetic point source	6	6.00	1.5	2.1	2.4	1.3
	10	10.00	2.7	3.6	4.2	2.4
	16	16.00	4.3	6.0	6.8	4.0
	20	20.00	4.8	7.4	8.4	4.9
‘Clinac 2100C’	6	6.88	1.4	2.2	2.6	1.5
	9	10.39	2.0	3.3	4.0	2.3
	12	13.05	2.7	4.4	5.2	3.0
	15	16.31	2.1	5.4	6.5	3.8
	18	19.49	1.9	6.3	7.7	4.5
SL75-20	5	5.66	1.1	1.7	2.1	1.2
	10	10.66	2.1	3.4	4.1	2.4
	14	15.34	1.9	4.9	6.0	3.5
	20	21.04	1.7	6.3	8.1	4.8
KD2	6	6.05	1.3	1.9	2.3	1.3
	11	10.51	2.5	3.6	4.2	2.4
	21	20.59	3.0	6.9	8.3	4.9
Therac 20	6	5.60	1.3	1.9	2.2	1.2
	13	12.41	3.2	4.5	5.2	3.0
	20	19.30	4.6	7.0	8.1	4.9
MM50	25	24.77	3.0	8.9	10.4	6.1
	40	39.50	2.5	11.9	15.5	9.2
	50	49.69	2.8	13.6	18.5	11.0

Table 6.2: Experimental data of water to clear polystyrene ionization ratio h_m at depth of maximum ionization.

Machine	$E_{nominal}$	R_{50} (cm)	h_m	chamber type	Source
Dynaray 10	5	1.97	1.025	Farmer	ref.[108]
Mevatron 80	7	-	1.035	Farmer nylon wall	ref.[108, 107]
Dynaray 10	7.5	3.00	1.023	Farmer	ref.[108]
Mevatron XII	7.8	-	1.03	Farmer A.E. wall	ref.[106]
Mevatron XII	7.8	-	1.03	Extradin A.E. wall	ref.[106]
Mevatron XII	7.8	-	1.04	Extradin T.E. wall	ref.[106]
Mevatron XII	7.8	-	1.04	PTW intracavitary	ref.[106]
Dynaray 10	10	4.03	1.022	Farmer	ref.[108]
SL75-20	10	-	1.026	Farmer	ref.[108]
SL75-20	10	-	1.024	Markus	ref.[108]
SL75-20	10	-	1.025	PTW flat	ref.[108]
Mevatron 80	10	-	1.035	Farmer nylon wall	ref.[108, 107]
Mevatron XII	10.2	-	1.03	Farmer A.E. wall	ref.[106]
Mevatron XII	10.2	-	1.03	Extradin A.E. wall	ref.[106]
Mevatron XII	10.2	-	1.04	Extradin T.E. wall	ref.[106]
Mevatron XII	10.2	-	1.05	PTW intracavitary	ref.[106]
Mevatron 80	10.2	-	1.035	Farmer nylon wall	ref.[108, 107]
Mevatron 80	12	-	1.035	Farmer nylon wall	ref.[108, 107]
SL75-20	14	-	1.028	Farmer	ref.[108]
SL75-20	14	-	1.016	Markus	ref.[108]
SL75-20	14	-	1.02	PTW flat	ref.[108]
Mevatron 80	15	-	1.025	Farmer nylon wall	ref.[108, 107]
Mevatron 80	18	-	1.008	Farmer nylon wall	ref.[108, 107]

Table 6.3: Experimental data of water to white polystyrene h_m at d_{max} .

Machine	$E_{nominal}$	R_{50} (cm)	h_m	chamber type	Source
	5	–	1.008	NACP Pl. Paral.	ref.[105]
	6	–	1.009	NACP Pl. Paral.	ref.[105]
SL75-20	6	2.6	1.021	Farmer	ref.[109]
SL75-14	6	–	1.021	Farmer	ref.[109]
SL75-10	6	–	1.020	Farmer	ref.[109]
Sagittaire	7	3.0	1.019	Farmer	ref.[109]
	8	–	1.009	NACP Pl. Paral.	ref.[105]
SL75-20	8	3.4	1.023	Farmer	ref.[109]
	12	–	1.008	NACP Pl. Paral.	ref.[105]
SL75-20	10	4.0	1.021	Farmer	ref.[109]
SL75-14	10	–	1.021	Farmer	ref.[109]
SL75-10	10	–	1.020	Farmer	ref.[109]
	12	–	1.007	NACP Pl. Paral.	ref.[105]
SL75-20	12	4.9	1.017	Farmer	ref.[109]
Sagittaire	13	5.4	1.020	Farmer	ref.[109]
BBC betatron	13	–	1.016	Farmer tufnol	ref.[104]
	14	–	1.006	NACP Pl. Paral.	ref.[105]
SL75-20	14	5.6	1.012	Farmer	ref.[109]
SL75-14	14	–	1.014	Farmer	ref.[109]
	17	–	1.005	NACP Pl. Paral.	ref.[105]
SL75-20	17	6.6	1.006	Farmer	ref.[109]
Sagittaire	19	8.0	1.006	Farmer	ref.[109]
	20	–	1.004	NACP Pl. Paral.	ref.[105]
SL75-20	20	7.5	1.003	Farmer	ref.[109]
	28	–	1.002	NACP Pl. Paral.	ref.[105]
	30	–	1.001	NACP Pl. Paral.	ref.[105]
BBC betatron	31	–	1.016	Farmer tufnol	ref.[104]

Table 6.4: Parameters obtained by linear fitting the calculated electron fluence corrections at d_{max} and at depth of $0.6R_{50} - 0.1$ for three plastics. $\phi_p^w = a_p - b_p R_{50}^{water}$.

depth (cm)	parameter	clear polystyrene	white polystyrene	PMMA
at	a_p	1.054	1.048	1.020
d_{max}	b_p	0.0043	0.0040	0.0025
at	a_p	1.052	1.044	1.021
$0.6R_{50} - 0.1$	b_p	0.0035	0.0028	0.0029

Figure 6.1: Comparison of scaled depth-ionization curves and depth-dose curves for a 12 MeV electron beam from a Clinac 2100C. The depths of dose maximum are the same (2.7 cm) in water or at water equivalent depth in clear polystyrene. The depth of ionization maximum in water is 2.3 cm while the depth of ionization maximum in clear polystyrene is only 1.5 cm.

Figure 6.2: Calculated central-axis dose curves of a 10 MeV monoenergetic electron beam and a 10 MeV clinical electron beam from an SL75-20 accelerator in different phantom materials.

18 MeV beam from Clinac 2100C

Figure 6.3: A comparison between calculated Spencer-Attix stopping-power ratios for an 18 MeV beam from a Clinac 2100C. The empty symbols are the calculated $\left(\frac{\bar{L}}{\rho}\right)_p^w$ in water and the filled symbols are the calculated $\left(\frac{\bar{L}}{\rho}\right)_s^w$ in water divided by the calculated $\left(\frac{\bar{L}}{\rho}\right)_s^p$ in plastic where the depths in plastic are scaled to the water equivalent depth according to eqn 6.2 and values of the depth-scaling factor from chapter 4. This shows that $\left(\frac{\bar{L}}{\rho}\right)_p^w$ and $\left(\frac{\bar{L}}{\rho}\right)_s^w / \left(\frac{\bar{L}}{\rho}\right)_s^p$ agree within 0.2%, indicating that the depth scaling procedure is giving equivalent depths at which the electron spectra are close to identical.

Figure 6.4: Comparison between our calculated fluence correction factors ϕ_p^w as a function of water equivalent depth for a 10 MeV beam from an SL75-20 accelerator and those obtained according to the AAPM TG-25 protocol[12]. The only differences between the upper and lower figure are in the scales.

Figure 6.5: Comparison between Monte Carlo calculated fluence correction factors ϕ_p^w as a function of water equivalent depth for 6 and 10 MeV monoenergetic point source incident beams and those determined according to the AAPM TG-25 protocol.

Figure 6.6: Comparison between calculated fluence correction factors with 16 and 20 MeV monoenergetic point source incident beams and those determined according to the AAPM TG-25 protocol.

Figure 6.7: Comparison between calculated fluence correction factors ϕ_p^w as a function of water equivalent depth for a 6 MeV beam from a Clinac 2100C accelerator and those obtained according to the AAPM TG-25 protocol.

Figure 6.8: Comparison between calculated fluence correction factors ϕ_p^w as a function of water equivalent depth for 9 and 12 MeV beams from a Clinac 2100C accelerator and those obtained according to the AAPM TG-25 protocol.

Figure 6.9: Comparison between calculated fluence correction factors ϕ_p^w as a function of water equivalent depth for 15 and 18 MeV beams from a Clinac 2100C accelerator and those obtained according to the AAPM TG-25 protocol.

Figure 6.10: Comparison between calculated fluence correction factors ϕ_p^w as a function of water equivalent depth for 5 and 20 MeV beams from an SL75-20 accelerator and those obtained according to the AAPM TG-25 protocol. Values of $E_{0,p}$ determined from the energy spectrum are 5.66 MeV and 21.1 MeV respectively. The values from the AAPM are a reference only since no data are listed below 6 MeV and above 20 MeV in the TG-25 protocol.

Figure 6.11: Comparison between calculated fluence correction factors ϕ_p^w as a function of water equivalent depth for 6 and 11 MeV beams from a KD2 accelerator and those obtained according to the AAPM TG-25 protocol.

Figure 6.12: Comparison between calculated electron fluence correction factors ϕ_y'' as a function of water equivalent depth for 6 and 20 MeV beams from a Therac 20 accelerator and those obtained according to the AAPM TG-25 protocol. Values of $E_{o,p}$ determined from the energy spectrum for the 6 MeV beam is 5.55 MeV. The values for a 6 MeV beam from the AAPM are a reference only since no data are listed below 6 MeV in the TG-25 protocol.

Figure 6.13: Calculated electron fluence correction factors ϕ_p^w as a function of water equivalent depth for 25 and 50 MeV beams from an MM50 accelerator.

Figure 6.14: Calculated electron fluence correction factors at d_{max} and at a depth of $0.6R_{50} - 0.1$ cm as a function of R_{50} in water for various clinical beams incident on clear polystyrene phantoms. The symbols which are characters are measured values: a—from ref.[108](NACP chamber), b—from ref.[107](Farmer chamber) and c from ref.[106].

Figure 6.15: Calculated electron fluence correction factors at $d_{\max}$ and at a depth of $0.6R_{50} - 0.1 \text{ cm}$ as a function of R_{50} in water for various clinical beams incident on white polystyrene phantoms. The symbols which are characters are measured values: a – from ref.[105], b – from ref.[109] and c – from ref.[104].

Figure 6.16: Calculated electron fluence correction factors at d_{max} and at a depth of $0.6R_{50} - 0.1$ cm as a function of R_{50} in water for various clinical beams incident on PMMA phantoms.

Figure 6.17: Calculated Spencer-Attix water to plastic stopping-power ratios for electron beams from the Clinac 2100C as a function of beam energy and depth for 3 plastics.

Electron beams from MM50

Figure 6.18: Calculated Spencer-Attix water to plastic stopping-power ratios for three electron beams from the MM50 accelerator as a function of beam energy and depth for 3 plastics.

Chapter 7

**Influence of the design of a
treatment head on output factors
due to the backscatter into the
beam monitor chamber**

7.1 Introduction

The dose per monitor unit (output) is a function of beam energy and field size. The output factor $OF(F)$ for clinical radiotherapy beams is defined as the ratio of dose per monitor unit U at $d_{\max}$ for a given field size F to that for the reference field size at its own $d_{\max,o}$ [12]. The dose per monitor unit, $D/U(F, d_{\max})$, is given by the following relationship[12]:

$$D/U(F, d_{\max}) = OF(F) \times D/U(F_o, d_{\max,o}) \quad (7.1)$$

where F_o is the reference field size.

The output factor depends upon two main factors:

- the dose deposited at $d_{\max}$ which is contributed by electrons that consists of: direct electrons (undergoing only scattering in the scattering foils and in the intervening air and phantom), scattered electrons from the jaws and electron applicators and side scatter in the phantom;
- the electron charge collected in the beam monitor chamber (BMC) which is influenced by the backscatter from the beam defining system of a machine treatment head.

Both factors depend upon the field size and beam energy. Due to the variations in the beam defining system of different accelerators, the contributions to the dose at $d_{\max}$ by scattered electrons from jaws and applicators can be very significant (up to 40%) for some systems according to Monte Carlo calculations in chapter 3 and measurements[113]. It is important to understand the sources which influence the output factors in order to make accurate treatment planing when irregular field or conformal therapy is used.

The effect on field-size dependent output due to backscattered radiation arising from jaws of an accelerator and reaching a beam monitor chamber has been studied by several authors[114, 115, 116, 117, 118, 119] for photon beams.

Mills *et al.*[120] studied the effect on field-size dependent output due to the side scatter for electron beams.

It is very difficult to separate these two factors experimentally since the details of origin of backscatter are unknown. It is even more difficult to study the influence of backscatter into the BMC using experimental method due to the complicated geometry of an accelerator head. This leads us to the Monte Carlo methods.

In this paper we focus on the backscatter and study its influences on the beam output factors.

The investigation provides details of backscatter about its origin and quantities for various beam energies from three accelerators. It also predicts energy dependence of the backscatter and variations of dose to water at d_{max} per monitor unit with beam energies as well as beam types (scanned vs scattered beams).

7.2 Method of Calculations

The Monte Carlo simulation starts electrons from the exit vacuum window of an accelerator and follows the transport of particles down to a water phantom. The dose to air in a BMC as shown in fig. 7.1 can be scored. By using the variable LATCH[77] which records each particle's complete history in the simulation we are able to separate the dose contributions to air in the BMC from particles according to their history. Therefore, the detailed backscatter information about where and how it originates can be obtained. By changing the size of jaw setting we can study the sensitivity of backscatter on field sizes.

The three accelerators (Clinac 2100C, SL75-20, Therac 20) simulated in this study have been described in chapter 2 and 3.

In the simulation the energy cutoffs for particle transport are set to $ECUT = AE = 0.521$ MeV (total energy) and $PCUT = AP = 0.010$ MeV to ensure that the effect

of low-energy electrons is considered adequately.

7.3 Results of Monte Carlo calculations

Tables 7.1 to 7.4 present summaries of calculated backscatter (BSC) into the beam monitor chamber from various beam defining components for electron beams from the Clinac 2100C accelerator with a 10×10 cm² field and with a jaw setting of 2×2 cm² field. Due to the materials between the BMC and jaws which block the backscatter from jaws the backscatter is insensitive to the jaws setting although the quantities of backscatter are not negligible (2 – 7%). Note that the ratio of dose-to-water at $d_{\max}$ in the water phantom to dose-to-air in the beam monitor chamber increases by a factor of 3 as the beam energy goes from the lowest energy to highest energy (table 7.2) for a 10×10 cm² field. It is also found that there is no contribution to the backscatter from the water phantom.

Tables 7.5 to 7.8 give summaries of calculated backscatter into the beam monitor chamber from various beam defining components for electron beams from the SL75-20 accelerator with a 10×10 cm² field and with a jaw setting of 2×2 cm² field. The backscatter is small (1 – 3%) for this accelerator because the geometry of the head design.

Tables 7.9 to 7.12 present summaries of calculated backscatter into the beam monitor chamber from various beam defining components for electron beams from the Therac 20 accelerator with a 10×10 cm² field and with a jaw setting of 2×2 cm² field. It is obvious that the backscatter is very sensitive to the jaw setting for the design of this accelerator. The backscatter increases from 7 to 12% when the jaw setting goes from 10×10 cm² field to 2×2 cm² field for a 6 MeV beam.

Tables 7.13 to 7.18 provide the calculation details of backscatter in each collecting window for all beam studied here. The total dose to air in the monitor chamber is

normalized to per incident electron history at the vacuum-exit-window of the accelerator. The results in tables 7.1 to 7.12 are obtained from the averaging of all collecting windows in the beam monitor chamber. The amount of backscatter increases when the collecting window is close to the bottom of the chamber because the backscatter comes from the downstream of the chamber.

Fig. 7.2 shows the backscatter as function of beam energy and the field sizes for three accelerators. It can be seen that the backscatter is very sensitive to the change of field size for the system of Therac 20. This is because there is no shielding material between the jaws and the beam monitor chamber in Therac 20 head. For the SL75-20 accelerator the backscatter shows some dependence on the field size although the quantities are small. This is due to a shielding ring between jaws and chamber which partially blocks backscatter from jaws.

7.4 Discussion

The amount of backscatter depends on the material downstream of the chamber and the space between these materials and the chamber. If the distance between jaws and the chamber is small (3 - 4 cm) and if there is no shielding material between them as in the case of the Therac 20 accelerator, the amount of backscatter is significant and is sensitive to the jaw setting. This influence will complicate output factors. However the backscatter dependence on field sizes can be removed by adding shielding material between chamber and jaws. Although this may increase the total quantity of backscatter, the total amount of backscatter will be independent of jaw setting since the backscatter into the chamber is from these shielding materials which are stationary and independent of field size.

We have also studied a 6 MeV beam from the Clinac 2100C using a different ECUT in the simulation ($AE = 0.521$ MeV and $ECUT = 0.700$ MeV). The results show a negligible decrease in the calculated backscatter comparing with the results

by using $ECUT = 0.521$ MeV. This indicates that our results are not sensitive to the simulation parameter ECUT.

7.5 Conclusions

We have presented details about backscatter into the beam monitor chamber for three accelerators. The calculated results show that the influence due to backscatter on output factors can be significant depending on the design of an accelerator treatment head.

The backscatter varies with beam energies and field size depending on an accelerator design. For beams from the Therac 20 the amount of backscatter increases from 7% to 12% and 4% to 5% as jaw setting goes from 20×20 to 2×2 cm² field for 6 and 20 MeV respectively. For beams from the Clinac 2100C, the backscatter is almost independent of jaw setting due to the materials between the jaws and the chamber although the quantities of the backscatter are not negligible. The amount of backscatter decreases from 7% to 2% as the beam energy goes from 6 to 18 MeV for the Clinac 2100C. For an SL75-20 accelerator, the backscatter is only increased from 2% to 3% (for 5 MeV beam) and from 0.6% to 0.9% (for 20 MeV beam) as jaw setting goes from 20×20 to 2×2 cm² field.

For a 10×10 cm² field, assuming a constant charge collecting efficiency the calculation predicts that the dose to water per monitor unit at d_{max} in the water phantom increases by a factor of 3 as the beam energy goes from the lowest energy to highest energy for the Clinac 2100C accelerator. This is because the lower-energy electrons have a larger scattering angle and as a result fewer electrons reach the phantom. For scanned beams the calculation predicts that the dose to water per monitor unit at d_{max} in the water phantom slightly decreases (variation is only 20%) as the beam energy increases from 6 to 20 MeV. The predicted energy dependence of the backscatter and dose to water at d_{max} per monitor unit may be verified experimentally.

4

OF/DE

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PM-1 3½"x4" PHOTOGRAPHIC MICROCOPY TARGET
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PRECISIONSM RESOLUTION TARGETS

Table 7.1: The backscatter into the beam monitor chamber from various beam defining components for electron beams from the Clinac 2100C accelerator with a 10×10 cm² field.

E_{nominal} (MeV)	total dose 10^{-12} $\pm 0.3\%$	total BSC 10^{-14} $\pm 3\%$	plate 10^{-14} $\pm 3\%$	jaws 10^{-15} $\pm 15\%$	applicator 10^{-17} $\pm 99\%$	dose at d_{max} 10^{-14} $\pm 1\%$
6	5.07	33.8	33.6	4.05	7	9.22
9	5.03	25.4	25.4	4.19	0.4	11.6
12	4.96	18.9	18.4	8.14	0.3	13.8
15	4.91	12.7	12.3	8.82	0.3	19.9
18	4.91	8.98	8.55	8.19	5	26.6

Table 7.2: The backscatter into the beam monitor chamber from various beam defining components for electron beams from the Clinac 2100C accelerator with a 10×10 cm² field.

E_{nominal} (MeV)	total backscatter % $\pm 2\%$	from plate % $\pm 3\%$	from jaws % $\pm 15\%$	applicator % $\pm 99\%$	dose ratio $d_{\text{max}}/\text{chamber}$ %
6	6.7	6.6	0.08	0.001	1.82
9	5.0	5.0	0.08	0.00008	2.31
12	3.8	3.7	0.16	0.00006	2.78
15	2.6	2.5	0.18	0.00006	4.05
18	1.8	1.7	0.17	0.001	5.42

Table 7.3: The backscatter into the beam monitor chamber from various beam defining components for electron beams from the Clinac 2100C accelerator with $2 \times 2 \text{ cm}^2$ jaw setting.

E_{nominal} (MeV)	total dose 10^{-12} $\pm 0.2\%$	total BSC 10^{-14} $\pm 2\%$	from plate 10^{-14} $\pm 3\%$	from jaws 10^{-14} $\pm 8\%$
6	5.10	35.8	34.7	2.3
12	4.97	20.1	19.3	1.9
18	4.91	10.2	9.22	2.0

Table 7.4: The % backscatter into the beam monitor chamber from various beam defining components for electron beams from the Clinac 2100C accelerator with $2 \times 2 \text{ cm}^2$ jaw setting.

E_{nominal} (MeV)	total backscatter % $\pm 2\%$	from plate % $\pm 3\%$	from jaws % $\pm 8\%$
6	7.0	6.8	0.4
12	4.0	3.9	0.4
18	2.1	1.9	0.4

Table 7.5: The backscatter into the beam monitor chamber from various beam defining components for electron beams from the SL75-20 accelerator with a 10×10 cm² field.

$E_{nominal}$ (MeV)	total dose 10^{-12} $\pm 0.2\%$	total BSC 10^{-14} $\pm 2\%$	ring 10^{-14} $\pm 3\%$	jaws 10^{-14} $\pm 3\%$	applicator 10^{-17} $\pm 50\%$	dose at d_{max} 10^{-14}
5	3.12	6.32	2.66	3.86	3	9.25
10	4.21	4.69	1.71	3.24	2	15.5
14	4.33	2.79	0.69	2.26	2	26.9
17	4.56	3.26	1.12	2.33	1	20.9
20	4.57	2.57	0.81	1.93	0.3	24.8

Table 7.6: The % backscatter into the beam monitor chamber from various beam defining components for electron beams from the SL75-20 accelerator with a 10×10 cm² field.

$E_{nominal}$ (MeV)	total backscatter % $\pm 2\%$	from ring % $\pm 3\%$	from jaws % $\pm 3\%$	applicator % $\pm 50\%$	dose ratio $d_{max}/chamber$ %
5	2.0	0.85	1.2	0.001	2.96
10	1.1	0.41	0.80	0.0005	3.68
14	0.64	0.16	0.52	0.0005	6.21
17	0.71	0.16	0.51	0.0002	4.58
20	0.56	0.18	0.42	0.00007	5.43

Table 7.7: The backscatter into the beam monitor chamber from various beam defining components for electron beams from the SL75-20 accelerator with a jaw setting of $2 \times 2 \text{ cm}^2$ field.

E_{nominal}	total dose to air 10^{-12} $\pm 0.2\%$	total from backscatter 10^{-14} $\pm 6\%$	from ring 10^{-14} $\pm 7\%$	from jaws 10^{-14} $\pm 6\%$
5	3.14	8.84	2.82	6.42
10	4.24	6.24	1.77	5.24
14	4.34	4.74	0.87	4.22
17	4.57	4.55	1.22	3.66
20	4.59	3.94	0.99	3.30

Table 7.8: The % backscatter into the beam monitor chamber from various beam defining components for electron beams from the SL75-20 accelerator with a jaw setting of $2 \times 2 \text{ cm}^2$ field.

E_{nominal} (MeV)	total backscatter % $\pm 2\%$	from ring % $\pm 7\%$	from jaws % $\pm 6\%$
5	2.8	0.90	2.0
10	1.5	0.42	1.2
14	1.1	0.20	1.0
17	1.0	0.27	0.80
20	0.86	0.22	0.72

Table 7.9: The backscatter into the beam monitor chamber from various beam defining components for electron beams from the Therac 20 accelerator with a 10×10 cm² field.

$E_{nominal}$ (MeV)	total dose 10^{-12} $\pm 0.4\%$	total BSC 10^{-13} $\pm 1\%$	from jaws 10^{-13} $\pm 2\%$	applicator 10^{-17} $\pm 60\%$	dose at d_{max} 10^{-13} $\pm 2\%$
6	1.96	1.13	1.42	9	1.55
9	2.13	1.35	1.33	6	1.60
13	2.17	1.10	1.08	9	1.52
17	2.23	0.91	0.89	3	1.46
20	2.22	0.81	0.79	3	1.39

Table 7.10: The % backscatter into the beam monitor chamber from various beam defining components for electron beams from the Therac 20 accelerator with a 10×10 cm² field.

$E_{nominal}$ (MeV)	total from backscatter % $\pm 1\%$	from jaws % $\pm 2\%$	from applicator % $\pm 60\%$	dose ratio $d_{max}/chamber$ %
6	7.3	7.2	0.005	7.9
9	6.3	6.2	0.003	7.5
13	5.1	5.0	0.004	7.0
17	4.1	4.0	0.001	6.5
20	3.6	3.6	0.001	6.3

Table 7.11: The backscatter into the beam monitor chamber from various beam defining components for electron beams from the Therac 20 accelerator with jaw setting of $2 \times 2 \text{ cm}^2$ field.

E_{nominal} (MeV)	total dose to chamber 10^{-12} $\pm 0.4\%$	total from backscatter 10^{-13} $\pm 2\%$	from jaws 10^{-13} $\pm 2\%$
6	2.05	2.44	2.42
9	2.20	2.36	2.18
13	2.23	1.67	1.65
17	2.25	1.33	1.30
20	2.26	1.16	1.14

Table 7.12: The % backscatter into the beam monitor chamber from various beam defining components for electron beams from the Therac 20 accelerator with jaw setting of $2 \times 2 \text{ cm}^2$ field.

E_{nominal} (MeV)	total from backscatter % $\pm 2\%$	from jaws % $\pm 2\%$
6	11.9	11.8
9	10.7	9.9
13	7.5	7.4
17	5.9	5.8
20	5.2	5.1

Table 7.13: The details of backscatter in each collecting window for electron beams from the Clinac 2100C' accelerator with a 10×10 cm² field.

$E_{nominal}$ (MeV)	collecting window #	total dose to air 10^{-12} $\pm 0.3\%$	total BSC' 10^{-14} $\pm 3\%$	plate 10^{-14} $\pm 3\%$	jaws 10^{-15} $\pm 15\%$	applicator 10^{-17} $\pm 90\%$
6	1	4.86	27.2	27.0	3.21	1
	2	5.01	31.3	31.1	3.94	5
	3	5.15	36.1	35.8	4.44	9
	4	5.25	40.6	40.3	4.61	13
9	1	4.91	21.1	20.9	3.39	0.0
	2	5.00	23.6	23.5	3.68	0.0
	3	5.10	27.0	26.8	4.48	1
	4	5.11	30.5	30.3	5.20	0.4
12	1	4.87	15.1	14.7	6.47	0.3
	2	4.92	17.4	17.1	7.58	0.3
	3	4.99	19.4	19.0	7.62	0.3
	4	5.05	23.1	22.7	10.9	0.3
15	1	4.82	9.93	9.65	5.75	0.4
	2	4.89	11.6	11.2	8.66	0.3
	3	4.95	13.8	13.2	10.9	0.3
	4	4.99	15.5	15.1	9.97	0.3
18	1	4.85	6.98	6.64	6.51	0.0
	2	4.90	8.14	7.61	9.01	0.0
	3	4.94	9.78	9.34	8.81	0.0
	4	4.95	11.0	10.6	8.44	20

Table 7.14: The details of backscatter in each collecting window for electron beams from the Clinac 2100C accelerator with 2×2 cm² jaw setting.

$E_{nominal}$ (MeV)	collecting window #	total dose to air 10^{-12} $\pm 0.3\%$	total BSC 10^{-14} $\pm 3\%$	plate 10^{-14} $\pm 3\%$	jaws 10^{-15} $\pm 8\%$
6	1	4.89	28.5	27.5	18.1
	2	5.06	33.1	32.0	22.1
	3	5.17	38.4	37.2	24.5
	4	5.28	43.1	42.0	25.5
12	1	4.87	16.3	15.5	16.1
	2	4.95	18.9	18.1	18.9
	3	5.00	21.1	20.2	20.6
	4	5.04	24.0	23.2	20.2
18	1	4.84	8.11	7.27	15.9
	2	4.91	9.30	8.36	19.6
	3	4.91	10.6	9.58	21.8
	4	4.98	12.7	11.7	22.8

Table 7.15: The details of backscatter from the SL75-20 accelerator with a 10×10 cm² field.

$E_{nominal}$ (MeV)	collecting window #	total dose 10^{-12} $\pm 0.2\%$	total BSC 10^{-14} $\pm 3\%$	ring 10^{-14} $\pm 5\%$	jaws 10^{-14} $\pm 3\%$	applicator 10^{-17} $\pm 70\%$
5	1	3.02	4.99	1.75	3.35	2
	2	3.05	5.51	2.13	3.60	2
	3	3.15	6.27	2.44	1.03	4
	4	3.17	7.02	3.09	4.13	2
	5	3.19	7.83	3.89	4.21	3
10	1	4.17	4.05	1.16	3.05	3
	2	4.16	4.19	1.46	2.96	2
	3	4.27	4.32	1.58	2.96	1
	4	4.23	5.04	1.93	3.12	1
	5	4.23	5.85	2.41	3.82	1
14	1	4.32	2.36	0.46	2.04	1
	2	4.30	2.45	0.49	2.10	1
	3	4.37	2.94	0.85	2.36	1
	4	4.34	2.91	0.73	2.32	4
	5	4.34	3.28	0.90	2.50	4
17	1	4.56	2.53	0.72	1.96	1
	2	4.53	2.80	0.80	2.19	1
	3	4.60	3.27	1.10	2.36	2
	4	4.58	3.94	1.40	2.70	1
	5	4.52	3.76	1.56	2.43	1
20	1	4.59	2.11	0.53	1.72	0.0
	2	4.54	2.31	0.58	1.89	0.0
	3	4.62	2.46	0.76	1.85	0.0
	4	4.57	2.87	0.97	2.13	1
	5	4.55	3.09	1.21	2.06	0.4

Table 7.16: Backscatter details from the SL75-20 with a 2×2 cm² field jaws setting.

$E_{nominal}$ (MeV)	collecting window #	total dose to air 10^{-12} $\pm 0.3\%$	total BSC 10^{-14} $\pm 6\%$	from ring 10^{-14} $\pm 7\%$	from jaws 10^{-14} $\pm 6\%$
5	1	3.04	7.10	1.84	5.53
	2	3.08	8.10	2.19	6.25
	3	3.18	8.85	2.70	6.56
	4	3.19	9.64	3.37	6.77
	5	3.23	10.5	4.02	6.98
10	1	4.21	5.78	1.20	4.87
	2	4.21	5.98	1.39	4.88
	3	4.28	6.98	1.77	5.66
	4	4.27	6.93	2.00	5.32
	5	4.23	7.51	2.48	5.47
14	1	4.31	4.01	0.51	3.77
	2	4.31	4.38	0.76	3.96
	3	4.40	4.88	0.88	4.39
	4	4.37	5.12	1.08	4.45
	5	4.32	5.30	1.13	4.52
17	1	4.57	3.75	0.76	3.24
	2	4.55	4.04	0.99	3.32
	3	4.62	4.80	1.26	3.91
	4	4.58	4.98	1.45	3.89
	5	4.54	5.19	1.63	3.96
20	1	4.59	3.41	0.67	2.99
	2	4.56	3.66	0.83	3.21
	3	4.63	3.84	0.90	3.28
	4	4.60	4.25	1.16	3.47
	5	4.55	4.56	1.39	3.54

Table 7.17: The details of backscatter in each collecting window from a Therac 20 accelerator with a 10×10 cm² field.

$E_{nominal}$ (MeV)	collecting window #	total dose to air 10^{-12} $\pm 0.4\%$	total from BSC 10^{-13} $\pm 2\%$	from jaws 10^{-13} $\pm 2\%$	applicator 10^{-17} $\pm 60\%$
6	1	1.91	1.26	1.26	6
	2	1.96	1.38	1.37	18
	3	1.99	1.48	1.46	6
	4	1.96	1.61	1.57	6
9	1	2.09	1.19	1.19	5
	2	2.16	1.25	1.24	5
	3	2.14	1.43	1.41	10
	4	2.14	1.52	1.46	5
13	1	2.19	0.96	0.96	8
	2	2.16	1.03	1.02	9
	3	2.17	1.15	1.13	5
	4	2.16	1.26	1.21	13
17	1	2.24	0.80	0.80	3
	2	2.22	0.84	0.83	3
	3	2.20	0.96	0.94	3
	4	2.27	1.04	0.98	3
20	1	2.26	0.68	0.68	3
	2	2.24	0.75	0.75	3
	3	2.22	0.86	0.84	3
	4	2.19	0.95	0.90	3

Table 7.18: The details of backscatter in each collecting window for electron beams from the Therac 20 accelerator with a 2×2 cm² field.

$E_{nominal}$ (MeV)	collecting window No.	total dose to air 10^{-12} $\pm 0.4\%$	total from BSC 10^{-13} $\pm 2\%$	from jaws 10^{-13} $\pm 2\%$
6	1	2.00	2.15	2.14
	2	2.04	2.33	2.32
	3	2.09	2.54	2.52
	4	2.08	2.73	2.68
9	1	2.16	1.94	1.93
	2	2.19	2.09	2.08
	3	2.22	2.93	2.28
	4	2.23	2.47	2.41
13	1	2.23	1.46	1.45
	2	2.24	1.58	1.57
	3	2.23	1.74	1.72
	4	2.22	1.89	1.85
17	1	2.26	1.13	1.13
	2	2.26	1.24	1.23
	3	2.24	1.39	1.37
	4	2.22	1.54	1.48
20	1	2.29	0.97	0.97
	2	2.28	1.07	1.06
	3	2.25	1.24	1.22
	4	2.23	1.34	1.29

Figure 7.1: A schematic of beam monitor ion chamber with three collecting windows. A chamber is usually located between scattering foils and jaws and consists of several collecting air cells (collecting windows) separated by thin plastics coated with metal.

Figure 7.2: The backscatter as a function of nominal beam energy for various beams from three accelerators with $10 \times 10 \text{ cm}^2$ field (filled symbols) and with jaw setting of $2 \times 2 \text{ cm}^2$ field at 100 cm SSD (empty symbols).

Chapter 8

Summary

The EGS4 Monte Carlo code, BEAM, has been described and benchmarked against experiments by comparing Monte Carlo predicted dose distributions with measured dose distributions in water phantoms. Using the BEAM code, a variety of electron beams used in radiotherapy has been simulated and systematically studied including the NRC research accelerator, the NPL calibration accelerator, and five commercial medical electron accelerators: a Varian Clinac 2100C, a Philips SL75-20, a Siemens KD2, an AECL Therac 20 and a Scanditronix MM50. The calculated and measured dose distributions in water phantoms are in good agreement in all cases which indicate that the simulated beams are accurate. The detailed information and characteristics of beams produced from these accelerators have been presented. These beams with an energy range of 5 – 50 MeV are commonly used for radiotherapy in cancer clinics.

The only adjustable parameter in the simulation is the incident electron energy, E_{in} , at the vacuum-exit-window of an accelerator. The energy E_{in} for simulations of clinical electron beams is selected by matching the value of the calculated and measured values of R_{50} , the depth at which the dose falls to 50% of the dose maximum. Assuming the simulation is accurate, we can also determine the incident beam

energy[121] at the vacuum-exit-window of an accelerator by matching the value of the calculated and measured values of R_{50} .

The realistic radiotherapy electron beams obtained from the simulation have been used as inputs to calculate various dosimetry parameters. In the past these parameters in dosimetry protocols [11, 13] were calculated using simple monoenergetic incident beams because of lack of details of realistic beams. The results of the calculations using the realistic incident beams improve the accuracy of these parameters.

This study has found that it is impossible to have a common relationship between the mean electron energy at the phantom surface, $\bar{E}_o$, and the half value depth, R_{50} , for various electron beams from different accelerators due to the variation of the number of scattered electrons from the beam defining system. However a common relationship may exist by considering only direct electrons (excluding those that hit jaws or applicators) which is in close agreement with the energy-range relationship calculated using monoenergetic beams by Rogers and Bielajew(1986)[86]. It is also shown that the empirical formula $E_{o,p} = 0.22 + 1.98R_p + 0.0025R_p^2$ describes accurately the relationship between R_p and $E_{o,p}$ for clinical beams of energies from 5 to 50 MeV with an accuracy of 3%. The values of the calculated mean electron energy as a function of depth $\bar{E}_d$ from this study indicate that there are up to 20% uncertainties determined according to the Harder's relationship employed in the AAPM protocol or determined according to the IAEA code or practice at the depth of dose maximum for some beams. It has been shown that the depth-scaling factors obtained by using the ratio of continuous-slowng-down ranges recommended in the IAEA Code of Practice[13] are inaccurate and overestimate the calculated values in this study by 1-2% in all cases.

As a result of this study, the Spencer-Attix restricted water/air collision stopping-power ratios are improved by up to 1.5% for some beams at the depth of dose maximum.

Also, the large variations of the measured electron fluence correction factors

have been understood. More accurate data of fluence correction factors for plastics are provided. The changes comparing to the values recommended in dosimetry protocols[12, 13] are up to 2% for some beams.

The full simulation of an accelerator head provides detailed information about the backscatter into the beam monitor chamber and predicts the influence on the beam output factor. The results show that the amount of backscatter into the beam monitor chamber depends on the material downstream of the chamber and the distance between them. The amount of backscatter may be significant and may be sensitive to the jaw settings depending on the design of an accelerator head.

This work has been devoted to developing a general purpose code for the simulation of the beams from clinical accelerators and to utilize the simulated clinical beams to improve the electron beam dosimetry. The information obtained from the simulated beams enhances our knowledge of many aspects of clinical electron beams. Extensive beam data from various accelerators are not only useful in understanding the radiotherapy electron beams but also are very important in predicting accurate dose distribution during treatment planning.

Up to now we have mainly used experimental results to verify the Monte Carlo simulations. Often, when we fail to get agreement between the experiments and simulations we have found that we were given incorrect information about a given accelerator because not all accelerators are standard, even for the same nominal models. This study demonstrates that Monte Carlo simulation is accurate and is very sensitive to changes of simulated geometry. The potential of using Monte Carlo as a tool to improve the design of accelerators and to predict dose distributions accurately is clear. The investigation indicates that we have arrived at the stage that the Monte Carlo technique may be used as an efficient tool to improve the design of medical accelerators and to predict experiment results accurately.

Bibliography

- [1] A.S. Shiu, "Three-dimensional electron beam dose calculations," PhD Thesis, University of Texas, Houston (1988).
- [2] E. Mah, J. Antolak, J.W. Scrimger and J.J. Battista, "Experimental evaluation of a 2D and a 3D electron pencil beam algorithm," *Phys. Med. Biol.* **34** 1179 – 1194 (1989).
- [3] C.X. Yu, W.S. Ge and J.W. Wong, "A multiray model for calculating electron pencil beam distributions," *Med. Phys.* **15** 662 – 671 (1988).
- [4] D. Jette, "Electron dose calculation using multiple-scattering theory: Localized inhomogeneities - A new theory," *Med. Phys.* **18** 123 – 132 (1991).
- [5] A.S. Shiu and K.R. Hogstrom, "Pencil-beam redefinition algorithm for electron dose distributions," *Med. Phys.* **18** 7 – 18 (1991).
- [6] J. McLellan, G.A. Sandison, L. Papiez and W. Huda, "A restricted angular scattering model for electron penetration in dense media," *Med. Phys.* **18** 1 – 6 (1991).
- [7] B.L. Werner, F.M. Khan and F.C. Deibel, "A model for calculating electron beam scattering in treatment planning," *Med. Phys.* **9** 180 – 187 (1982).
- [8] I. Lax and A. Brahme, "Electron beam dose planning using gaussian beams," *Acta Radiol. Oncol.* **24** 75 – 85 (1985).

- [9] D. Jette, "Electron dose calculation using multiple-scattering theory: A. Gaussian multiple-scattering theory," *Med. Phys.* **15** 123 - 137 (1988).
- [10] I.A.D. Bruinvis, W.A.F. Mathol and P. Andreo, "Inclusion of electron range straggling in the Fermi-Eyges multiple-scattering theory," *Phys. Med. Biol.* **34** 491 - 507 (1989).
- [11] AAPM TG-21, "A protocol for the determination of absorbed dose from high-energy photon and electron beams," *Med. Phys.* **10** 741 - 771 (1983).
- [12] F.M. Khan, K.P. Doppke, K.R. Hogstrom, G.J. Kutcher, R. Nath, S.C. Prasad, J.C. Purdy, M. Rozenfeld and B.L. Werner, "Clinical electron-beam dosimetry: Report of AAPM Radiation Therapy Committee Task Group 25," *Med. Phys.* **18** 73 - 109 (1991).
- [13] IAEA, International Atomic Energy Agency, "Absorbed Dose Determination in Photon and Electron Beams; An International Code of Practice," Technical Report Series No. 277, IAEA, Vienna (1987).
- [14] D.W.O. Rogers and A.F. Bielajew, "Monte Carlo techniques of electron and photon transport for radiation dosimetry," in "The Dosimetry of Ionizing Radiation", Vol III, eds. K.R. Kase, B.E. Bjarngard, and F.H. Attix, (Academic Press) 427 - 539 (1990).
- [15] W.R. Nelson, H. Hirayama and D.W.O. Rogers, "The EGS4 Code System," Stanford Linear Accelerator Center Report SLAC-265 (Stanford Calif) (1985).
- [16] A.F. Bielajew, H. Hirayama, W.R. Nelson and D.W.O. Rogers, "History, overview and recent improvements of EGS4," National Research Council of Canada Report PIRS-0436 (1994).
- [17] M.J. Berger and R. Wang, "Multiple-Scattering Angular Deflections and Energy-Loss Straggling," in "Monte Carlo Transport of Electrons and Pho-

- tons Below 50 MeV", eds. T.M. Jenkins, W.R. Nelson, A. Rindi, A.E. Nahum and D.W.O. Rogers, (Plenum Press) 21 – 56 (1989).
- [18] P. Andreo, A. Brahme, A.E. Nahum and O. Mattsson, "Influence of energy and angular spread on stopping-power ratios for electron beams," *Phys. Med. Biol.* **34** 751 – 768 (1989).
- [19] M.S. Huq and R. Nath, "Comparison of IAEA 1987 and AAPM 1983 protocols for dosimetry calibration of radiotherapy beams," *Med. Phys.* **18** 26 – 35 (1991).
- [20] D.W.O. Rogers, A.F. Bielajew, T.R. Mackie and S.S. Kubsad, "The OMEGA Project: treatment planning for electron-beam radiotherapy using Monte Carlo techniques," *Phys. Med. Biol.* **35** (abs) 285 (1990).
- [21] T.R. Mackie, S.S. Kubsad, D.W.O. Rogers and A.F. Bielajew, "The OMEGA project: Electron dose planning using Monte Carlo simulation," *Med. Phys.* **17** (abs) 730 (1990).
- [22] J. Cygler, J.J. Battista, J.W. Scrimger, E. Mah and J. Antolak, "Electron dose distributions in experimental phantoms: a comparison with 2D pencil beam calculations," *Phys. Med. Biol.* **32** 1073 – 1086 (1987).
- [23] M. Morawska-Kaczynska and H. Huizenga, "Numerical calculation of energy deposition by broad high-energy electron beams: II. Multi-layered geometry," *Phys. Med. Biol.* **37** 2103 – 2116 (1992).
- [24] D. Jette and S. Walker, "Electron dose calculation using multiple-scattering theory: Evaluation of a new model for inhomogeneities," *Med. Phys.* **19** 1241 – 1254 (1992).
- [25] K.R. Shortt, C.K. Ross, A.F. Bielajew and D.W.O. Rogers, "Electron Beam Dose Distributions Near Standard Inhomogeneities," *Phys. Med. Biol.* **31** 235 – 249 (1986).

- [26] T.R. Mackie, "Applications of the Monte Carlo Method in Radiotherapy," in Vol. III of "Dosimetry of Ionizing Radiation" eds. K. Kase, B. Bjarngard and F.H. Attix (Academic Press, New York) 541 - 620 (1990).
- [27] A.F. Bielajew and D.W.O. Rogers, "A standard timing benchmark for EGS4 Monte Carlo Calculations," *Med. Phys.* **19** 303 - 304 (1992).
- [28] M.A. Holmes, T.R. Mackie, W. Sohn, P.J. Reckwerdt, T.J. Kinsella, A.F. Bielajew and D.W.O. Rogers, "The application of correlated sampling to the computation of electron beam dose distributions in heterogeneous phantoms using the Monte Carlo method," *Phys. Med. Biol.* **38** 675 - 688 (1993).
- [29] H. Neuenschwander, T.R. Mackie, P.J. Reckwerdt, and J. Yang, "Improved Macro Monte Carlo method for electron beam dose calculations," submitted to *Phys. Med. Biol.* (1994).
- [30] C.M. Ma, D.W.O. Rogers, B.A. Faddegon, G.X. Ding, J.S. Wei, A.F. Bielajew and T.R. Mackie, "Simplified models of electron beams from a Clinac 2100C accelerator," *Med. Phys.* **20** 1295 (abstract) (1993).
- [31] M. Udale-Smith, "A Monte Carlo Investigation of High Energy Electron Beams Used in Radiotherapy," PhD Thesis, Leeds University (1990).
- [32] M. Udale, "Monte Carlo calculations of electron beam parameters for three Philips linear accelerators," *Phys. Med. Biol.* **37** 85 - 105 (1992).
- [33] M. Udale, "A Monte Carlo investigation of surface doses for broad electron beams," *Phys. Med. Biol.* **33** 939 - 954 (1988).
- [34] C. Manfredotti, U. Nastasi, R. Ragona and S. Anglesio, "Comparison of three-dimensional Monte Carlo simulation and the pencil beam algorithm for an electron beam from a linear accelerator," *Nucl. Instr. Meth.* **A255** 355 - 359 (1987).

- [35] C. Manfredotti, U. Nastasi, R. Marchisio, C. Ongaro, G. Gervino, R. Ragona, S. Anglesio and G. Sannazzari, "Monte Carlo simulation of dose distribution in electron beam radiotherapy treatment planning," *Nucl. Instr. Meth.* **A291** 646 - 654 (1990).
- [36] A.A. Al-Beteri and D.E. Raeside, "Optimal electron-beam treatment planning for retinoblastoma using a new three-dimensional Monte Carlo-based treatment planning system," *Med. Phys.* **19** 125 - 135 (1992).
- [37] T.R. Mackie, J.W. Scrimger and J.J. Battista, "A convolution method of calculating dose for 15 MV x-rays," *Med. Phys.* **12** 188 - 196 (1985).
- [38] R. Mohan, C. Chui and L. Lidofsky, "Differential pencil beam dose computation model for photons," *Med. Phys.* **13** 64 - 73 (1986).
- [39] G.X. Ding, D.W.O. Rogers and T.R. Mackie, "Calculation of stopping-power ratios using realistic clinical electron beams," *Med. Phys.* **22**(5) (1995).
- [40] P.L. Petti, M.S. Goodman, T.A. Gabriel and R. Mohan, "Investigation of buildup dose from electron contamination of clinical photon beams," *Med. Phys.* **10** 18 - 24 (1983).
- [41] P.L. Petti, M.S. Goodman, J.M. Sisterson, P.J. Biggs, T.A. Gabriel and R. Mohan, "Sources of electron contamination for the Clinac-35 25-MV photon beam," *Med. Phys.* **10** 856 - 861 (1983).
- [42] ICRU, "Specification of high-activity gamma-ray sources," International Commission on Radiation Units and Measurements Report 18 (ICRU Bethesda U.S.A.) (1971).
- [43] D.W.O. Rogers, G.M. Ewart, A.F. Bielajew and G. van Dyk, "Calculation of Electron Contamination in a ^{60}Co Therapy Beam," In "Proceedings of the IAEA International Symposium on Dosimetry in Radiotherapy" (IAEA, Vienna), Vol 1 303 - 312 (1988).

- [44] D.W.O. Rogers, G.M Ewart, A.F. Bielajew and G. van Dyk, "Calculation of Contamination of the ^{60}Co Beam from an AECL Therapy Source," NRC Report PXNR-2710 (1985).
- [45] K. Han, D. Ballon, C. Chui and R. Mohan, "Monte Carlo simulation of a cobalt-60 beam," *Med. Phys.* **14** 414 - 419 (1987).
- [46] R. Mohan, C. Chui and L Lidofsky, "Energy and angular distributions of photons from medical linear accelerators," *Med. Phys.* **12** 592 - 597 (1985).
- [47] S.S. Kubsad, T.R. Mackie, M.A. Gehring, D.J. Misisco, B.R. Paliwal, M.P. Mehta and T.J. Kinsella, "Monte Carlo and convolution dosimetry for stereotactic radiosurgery," *Int. J. Radiation Oncology Biol. Phys.* **19** 1027 - 1035 (1990).
- [48] A. Kosunen and D.W.O. Rogers, "Beam Quality Specification for Photon Beam Dosimetry," *Med. Phys.* **20** 1181 - 1188 (1993).
- [49] E.L. Chaney, T.J. Cullip and T.A. Gabriel, "A Monte Carlo study of accelerator head scatter," *Med. Phys.* **21** 1383 - 1390 (1994).
- [50] A.F. Bielajew, R. Mohan and C.S. Chui, "Improved bremsstrahlung photon angular sampling in the EGS4 code system," National Research Council of Canada Report PIRS-0203 (1989).
- [51] B.A. Faddegon, C.K. Ross and D.W.O. Rogers, "Angular distribution of bremsstrahlung from 15 MeV electrons incident on thick targets of Be, Al and Pb," *Med. Phys.* **18** 727 - 739 (1991).
- [52] B.A. Faddegon and D.W.O. Rogers, "Comparisons of thick-target bremsstrahlung calculations by EGS4 and ITS," *Nuc. Inst. Meth. A* **327** 556 - 565 (1993).

- [53] A.F. Bielajew and D.W.O. Rogers, "PRESTA: The Parameter Reduced Electron-Step Transport Algorithm for Electron Monte Carlo Transport," *Nuclear Instruments and Methods* **B18** 165 – 181 (1987).
- [54] W.R. Nelson and D.W.O. Rogers, "Structure and Operation of the EGS4 code system," in "Monte Carlo Transport of Electrons and Photons Below 50 MeV", eds. T.M. Jenkins, W.R. Nelson, A. Rindi, A.E. Nahum and D.W.O. Rogers, (Plenum Press) 287 – 306 (1989).
- [55] A.J. Cook, "Mortran3 user's guide," SLAC Computa. Res. Group Tech. Memo. No. CGTM 209 (1983).
- [56] R. Brun, O. Couet, C. Vandoni and P. Zancarini, "PAW User Guide," CERN Computer Center, Geneva, Switzerland (1992)
- [57] A.F. Bielajew, and P.E. Weibe, "EGS-Windows - A Graphical Interface to EGS," NRC Report: PIRS-0274 (1991).
- [58] Y. Namito, W.R. Nelson, S.M. Seltzer, A.F. Bielajew and D.W.O. Rogers, "Low-energy x-ray production studies using the EGS4 code system," *Med. Phys.* **17** (abstract) 557 (1990).
- [59] J. Cygler and D.W.O. Rogers, "Monte Carlo calculations of tungsten mesh electron beam modifier," *Phys. Med. Biol.* **39** (abstract) 677 (1994).
- [60] D.W.O. Rogers, "How Accurately Can EGS4/PRESTA Calculate Ion Chamber Response?," *Med. Phys.* **20** 319 – 323 (1993).
- [61] B.A. Faddegon, C.K. Ross and D.W.O. Rogers, "Forward Directed Bremsstrahlung of 10 – 30 MeV Electrons Incident on Thick Targets of Al and Pb," *Med. Phys.* **17** 773 – 785 (1990).

- [62] H. Kahn, "Use of different Monte Carlo sampling techniques," in Symposium on Monte Carlo Methods, (H.A. Meyer ed.) (John Wiley and Sons, New York) 146 - 190 (1956).
- [63] E.D. Cashwell and C.J. Everett, "Monte Carlo Method for random walk problems," (Pergamon Press, New York) (1959).
- [64] B. Nilsson and A. Brahme, "Absorbed dose from secondary electrons in high energy photon beams," *Phys. Med. Biol.* **24** 901 - 912 (1979).
- [65] D.W.O. Rogers and A.F. Bielajew, "The Use of EGS for Monte Carlo Calculations in Med. Phys.," Report PXNR-2692, National Research Council of Canada, (Ottawa, Canada K1A 0R6) (1984).
- [66] D.W.O. Rogers, "Low energy electron transport with EGS," *Nucl. Inst. Meth.* **227** 535 - 548 (1984).
- [67] C.M. Wells, T.R. Mackie, M.B. Podgorsak, M.A. Holmes, N. Papanikolaou, P.J. Reckwerdt, J. Cygler, D.W.O. Rogers, A.F. Bielajew, D.G. Schmidt, and J.K. Muehlenkamp, "Measurements of electron dose distribution near inhomogeneities using a plastic scintillator detector," *Int. J. Radiat. Oncol. Biol. Phys.* **29** 1157 - 1165 (1994).
- [68] J.O. Deasy, "Electron energy and angular distributions in radiotherapy," PhD Thesis University of Kentucky (1992).
- [69] P. Andreo and A. Fransson, "Stopping-power ratios and their uncertainties for clinical electron beam dosimetry," *Phys. Med. Biol.* **34** 1847 - 1861 (1989).
- [70] C.K. Ross and K.R. Shortt, "Energy Calibration of the 90-B Beam Line of the IRS Linac," NRC Report Report PIRS-0021 (1985).

- [71] D.W.O. Rogers, "The role of Monte Carlo simulation of electron transport in radiation dosimetry," *Int'l J of Appl. Radiation and Isotopes*, **42** 965 – 974 (1991).
- [72] G.X. Ding, D.W.O. Rogers, and T.R. Mackie, "Calculation of electron mean energy in water using realistic electron beams," *Med. Phys.* **21** (abs) 894 (1994).
- [73] G.X. Ding, D.W.O. Rogers and T.R. Mackie, "Mean energy, energy-range relationships and depth-scaling factors for clinical electron beams," submitted to *Med. Phys.* (April, 1995).
- [74] A. Kassae, X. Chen, S. Ayyalasomayajula and P. Bloch, "Electron beam characteristic with modified electron cones and scattering foil on a Varian 2100C accelerator," *Med. Phys.* **20** abs. 887 (1993).
- [75] A. Kassae, M.D. Altschuler, S. Ayyalasomayajula and P. Bloch, "Influence of cone design on electron beam characteristics on clinical accelerators," *Med. Phys.* **21** 1671 – 1676 (1994).
- [76] D. Sheikh-Bagheri, D.W.O. Rogers and T.R. Mackie, "Monte Carlo model of the 6 MV photon beam from a SL25 accelerator," *Med. Phys.* **21** 1368 (abstract) (1994).
- [77] D.W.O. Rogers, B.A. Faddegon, G.X. Ding, C.-M. Ma, J. Wei and T.R. Mackie, "BEAM: A Monte Carlo code to simulate radiotherapy treatment units," *Med. Phys.* **22** (5) (1995).
- [78] G.X. Ding, D.W.O. Rogers, and T.R. Mackie, "Electron beam spectra from different accelerators in radiotherapy," *Med. Phys.* **21** (abs) 1364 (1994).
- [79] G.X. Ding, D.W.O. Rogers, C.K. Ross, J.S. Wei, A.F. Bielajew, B.A. Faddegon, T.R. Mackie, P. Jursinic and J. Deasy, "Measured vs Monte Carlo calculated electron beam dose distributions," *Med. Phys.* **20** (abs) 933 (1993).

- [80] G.X. Ding, D.W.O. Rogers, J. Cygler and T.R. Mackie, "Electron fluence correction factors used in conversion of dose measured in plastic to dose in water," submitted to *Med. Phys.* (April, 1995).
- [81] NACP, "Procedures in external beam radiotherapy dosimetry with photon and electron beams with maximum energies between 1 and 50 MeV," *Acta Radiol. Oncol.* **19** 55 - 79 (1980).
- [82] ICRU, "Radiation Dosimetry: Electron beams with energies between 1 and 50 MeV," ICRU Report 35, Bethesda MD (1981).
- [83] A. Brahme, T. Kraepelien and H. Svensson, "Electron and photon beams from a 50 MeV racetrack microtron," *Acta Radiol. Oncol.* **19** 305 - 319 (1980).
- [84] B. Markus, "Beitrage zur Entwicklung der Dosimetrie schneller Elektronen," *Strahlentherapie* **116** 280 - 286 (1961).
- [85] M.J. Berger and S.M. Seltzer, "Quality of radiation in water medium irradiated with high-energy electron beams," *Book of Abstracts of XII International Congress of Radiology* 127 (1969).
- [86] D.W.O. Rogers and A.F. Bielajew, "Differences in Electron Depth Dose Curves Calculated with EGS and ETRAN and Improved Energy Range Relationships," *Med. Phys.* **13** 687 - 694 (1986).
- [87] B.B. Sorcini and A. Brahme, "An accurate energy-range relationship for high-energy electron beams in arbitrary materials," *Phys. Med. Biol.* **39** 795 - 811 (1994).
- [88] M.J. Berger and S.M. Seltzer, "," NBS Report NBSIR 82-2451 (1982).
- [89] A. Wu, A.M. Kalend, R.D. Zwicker and E.S. Sternick, "Comments on the method of energy determination for electron beam in the TG-21 protocol," *Med. Phys.* **11** 871 - 872 (1984).

- [90] A. Brahme and H. Svensson, "Specification of electron beam quality from the central-axis depth absorbed-dose distribution," *Med. Phys.* **3** 95 - 102 (1976).
- [91] P. Andreo and A. Brahme, "Mean energy in electron beams," *Med. Phys.* **8** 682 - 687 (1981).
- [92] B. Grosswendt and M. Roos, "Electron beam absorption in solid and water phantoms: depth scaling and energy-range relations," *Phys. Med. Biol.* **34** 509 - 518 (1989).
- [93] G.X. Ding and D.W.O. Rogers, "Energy spectra, angular spread, and dose distributions of electron beams from various accelerators used in radiotherapy," National Research Council of Canada Report PIRS-0439 (see <http://www.irs.inms.nrc.ca/inms/irs/papers/PIRS439/pirs439.html>) (March, 1995).
- [94] C. Malamut, D.W.O. Rogers and A.F. Bielajew, "Calculation of water/air stopping-power ratios using EGS4 with explicit treatment of electron - positron differences," *Med. Phys.* **18** 1222 - 1228 (1991).
- [95] ICRU, "Stopping Powers for Electrons and Positrons," Report 37, Bethesda MD (1984).
- [96] S. Duane, A.F. Bielajew and D.W.O. Rogers, "Use of ICRU-37/NBS Collision Stopping Powers in the EGS4 System," NRC Report PIRS-0177, Ottawa, March (1989).
- [97] K.A. Johansson, L.O. Mattsson, L. Lindborg and H. Svensson, "Absorbed-dose determination with ionization chambers in electron and photon beams having energies between 1 and 50 MeV," IAEA Symposium Proceedings, (Vienna) IAEA-SM-222/35 243 - 270 (1977).

- [98] M.J. Berger, S.M. Seltzer, S.R. Domen, and P.J. Lamperti, "Stopping-power ratios for electron dosimetry with ionization chambers," in *Biomedical Dosimetry*, (IAEA Vienna) 589 - 609 (1975).
- [99] S.C. Klevenhagen, "An algorithm to include the bremsstrahlung contamination in the determination of the absorbed dose in electron beams," *Phys. Med. Biol.* **39** 1103 - 1112 (1994).
- [100] A. Kosunen and D.W.O. Rogers, "DDSPR: A Code for calculating photon beam depth-dose curves and stopping-power ratios for an arbitrary spectrum," National Research Council Canada Report PIRS-298 (1992).
- [101] J.O. Deasy, P.R. Almond and M.T. McEllistrem, "The spectra dependence of electron central-axis depth-dose curves," *Med. Phys.* **21** 1369 - 1376 (1994).
- [102] P. Andreo, A. Hernandez, E. Millan and S. Millan, "Comments on the transfer of absorbed dose from plastic to water in electron beams," *Med. Phys.* **11** 874 - 876 (1984).
- [103] P.R. Almond, F.H. Attix, S. Goetsch, L.J. Humphries, H. Kubo, R. Nath and D.W.O. Rogers, "The calibration and use of plane-parallel ionization chambers for dosimetry of electron beams: An extension of the 1983 AAPM protocol, Report of AAPM Radiation Therapy Committee Task Group 39," *Med. Phys.* **21** 1251 - 1260 (1994).
- [104] H. Svensson, "unknown," *Acta Radiol. Ther. Phys. Biol.* **10** 631 - 654 (1971).
- [105] L.O. Mattsson, K.A. Johansson and H. Svensson, "Calibration and use of plane-parallel ionization chambers for the determination of absorbed dose in electron beams," *Acta Radiol. Ther. Phys. Biol.* **20** 385 - 399 (1981).
- [106] K.R. Kase, G.J. Adler, and B.E. Bjarngard, "Comparison of electron beam dose measurements in water and polystyrene using various dosimeters," *Med. Phys.* **9** 13 - 19 (1982).

- [107] L.W. Berkley and W.F. Hanson, "Measurement of electron beam ionization vs depth in various tissue-like materials," *Med. Phys.* **9** 607 (abs) (1982).
- [108] D.I. Thwaites, "Measurements of ionisation in water, polystyrene and a 'solid-water' phantom material for electron beams," *Phys. Med. Biol.* **30** 41 – 53 (1985).
- [109] I.A.D. Bruinvis, S. Heukelom and B.J. Mijnheer, "Comparison of ionization measurements in water and polystyrene for electron beam dosimetry," *Phys. Med. Biol.* **30** 1043 – 1053 (1985).
- [110] L.O. Mattsson and A.E. Nahum, "Concerning comparisons of electron beam absorbed dose measurements in water and polystyrene," *Med. Phys.* **11** 85 – 87 (1984).
- [111] S.C. Klevenhagen, D. D'Souza and I. Bonnefoux, "Complications in low energy x-ray dosimetry caused by electron contamination," *Phys. Med. Biol.* **36** 1111 – 1116 (1991).
- [112] M.J. Berger and S.M. Seltzer, "Stopping power and ranges of electrons and positrons," NBS Report NBSIR 82-2550-A (second edition) (1983).
- [113] R. Loevinger, C.J. Karzmark, M. Weissbluth, "Radiation therapy with high energy electrons Part I: Physical considerations 10 to 60 MeV," *Radiology* **77** 906 – 927 (1961).
- [114] M.S. Patterson and P.C. Shragge, "Characteristics of an 18 MV photon beam from a Therac 20 Medical Linear Accelerator," *Med. Phys.* **8** 312 – 318 (1981).
- [115] G. Luxton and M.A. Astrahan, "Output factor constituent of a high-energy photon beam," *Med. Phys.* **15** 88 – 91 (1988).
- [116] D.L. Watts and G.S. Ibbott, "Measurement of beam current and evaluation of scatter production in a 18-MeV accelerator," *Med. Phys.* **14** 662 – 664 (1987).

- [117] P.-H. Huang, J. Chu and B.E. Bjarngard, "The effect of collimator backscatter radiation on photon output of linear accelerators," *Med. Phys.* **14** 268 - 267 (1987).
- [118] S.S. Kubsad, T.R. Mackie, B.R. Paliwal and F.H. Attix, "Monte Carlo calculation of electron and secondary photon spectra from a Varian Clinac-2500," *Med. Phys.* **16** (abstract) 449 (1989).
- [119] C. Duzenli, B. McClean and C. Field, "Backscatter into the beam monitor chamber: Implications for dosimetry of asymmetric collimator," *Med. Phys.* **20** 363 - 367 (1993).
- [120] M.D. Mills, K.R. Hogstrom and P.R. Almond, "Prediction of electron beam output factors," *Med. Phys.* **9** 60 - 68 (1982).
- [121] G.X. Ding and D.W.O. Rogers, "Monte Carlo simulation of NPL linac and calculation of dose distributions and water/air stopping-power ratios," National Research Council of Canada Report PIRS 0399 (1993).

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