TOPICAL REVIEW

Monte Carlo modelling of electron beams from medical accelerators

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Abstract. Monte Carlo simulation of radiation transport is considered to be one of the most accurate methods of radiation therapy dose calculation. With the rapid development of computer technology, Monte Carlo based treatment planning for radiation therapy is becoming practical. A basic requirement for Monte Carlo treatment planning is a detailed knowledge of the radiation beams from medical accelerators. A practical approach to obtain the above is to perform Monte Carlo simulation of radiation transport in the medical accelerator. Additionally, Monte Carlo modelling of the treatment machine head can also improve our understanding of clinical beam characteristics, help accelerator design and improve the accuracy of clinical dosimetry by providing more realistic beam data. This paper summarizes work over the past two decades on Monte Carlo simulation of clinical electron beams from medical accelerators.

1. Introduction

The Monte Carlo method is a statistical simulation method. For radiation transport problems, it simulates the tracks of individual particles by sampling appropriate quantities from the probability distributions governing the individual physical processes using machine-generated (pseudo-) random numbers. Average values of macroscopic quantities such as particle fluence, energy spectrum and absorbed dose distribution can be calculated by simulating a large number of particle histories. The Monte Carlo method and its application in medical radiation physics, especially in radiation therapy physics, have been discussed in a number of publications (Burlin et al 1973, Raeside 1976, Nelson and Jenkins 1980, Rogers and Bielajew 1984, 1990, Turner et al 1985, Nahum 1985, Jenkins et al 1988, Mackie 1990, Rogers 1991, Andreo 1991).

The Monte Carlo method can precisely model the physical processes involved in radiation therapy and is powerful in dealing with any complex geometry. It is widely accepted that Monte Carlo simulation of radiation transport is one of the most accurate methods for predicting absorbed dose distributions in radiation therapy. In particular, Monte Carlo simulation can handle backscatter from high-density materials such as bone, or scatter perturbations by air cavities more accurately than any other existing dose calculation model (Rogers and Bielajew 1990, Nahum 1985, 1988, Mackie 1990, Rogers 1991, Andreo 1991, Bielajew 1994, Mohan 1997a). The major shortcoming of the Monte Carlo method, namely being computationally intensive, has become much less severe due to the rapid increase in speed and decrease in cost of computers, and the employment of innovative variance reduction techniques (Ma and Nahum 1993, Holmes et al 1993). Monte Carlo simulation is fast becoming the next generation

Over the past 20 years or so, many investigations have been carried out on the Monte Carlo simulation of photon beams from medical accelerators or $^{60}$Co teletherapy units. McCall et al (1978) investigated the effects of various targets and flattening filters on the mean energy of photon beams using the EGS3 code (Ford and Nelson 1978). Patau et al (1978) pioneered the Monte Carlo simulation of a complete photon accelerator. They simulated the generation of photons in a W–Cu target, the transport of photons through a flattening filter and the collimators, and the attenuation of photons in slabs of various materials. Nilsson and Brahme (1981) investigated the contaminant photons scattered from flattening filters and collimators. To investigate the electron contamination in photon beams, Petti et al (1983a, b) simulated a treatment machine head in great detail using a cylindrical geometry package to approximate various components of the linear accelerator. Mohan et al (1985) performed similar detailed simulations based on the EGS3 system to calculate photon spectra and fluence distributions from several accelerators. A special geometry package was also developed to model the exact shape of the flattening filter. Han et al (1987) also employed the EGS3 code to simulate in detail the treatment head of a $^{60}$Co unit. They approximated the complex geometry of a Theratron-780 $^{60}$Co unit as a source capsule, the source housing, and the collimator assembly. Similar studies of clinical photon beams were carried out using the EGS4 code system (Nelson et al 1985). Rogers et al (1988) investigated the sources of electron contamination in a $^{60}$Co beam. Chaney et al (1994) simulated a 6 MV photon accelerator to study the origins of head scatter. Lovelock et al (1994) simulated the photon beams from a Scanditronix MM50 machine to obtain the beam characteristics needed for treatment planning. An EGS4 user code, McRad, which was a generic Monte Carlo model of a photon linear accelerator, was developed by Lovelock et al (1995). Sixel and Faddegon (1995) simulated a Therac-6 treatment head in radiosurgery mode using the cylindrically symmetric EGS4 user code FLURZ with the PRESTA algorithm (Bielajew and Rogers 1987). They calculated the 6 MV radiosurgical x-ray spectra both with and without the flattening filter and compared the results with the Schiff’s thin target analytical spectra and the flattened Monte Carlo spectrum calculated by Mohan et al (1985). To study
the differential beam hardening effect of the flattening filter. Lee (1997) simulated the 6 MV beam from a Varian Clinac 2100C accelerator using the EGS4 code. Liu et al (1995) used a Combinatorial Geometry package to characterize treatment head components, which were modelled as 3D objects and combined using Boolean algebra. Simulations were performed with the ACCEPT code of the ITS 3.0 Monte Carlo system (Halbeib and Mehlhorn 1984). The MCNP Monte Carlo code (Hendricks and Briesmeister 1992, Briesmeister 1993) was also used to model the clinical photon beams. To determine the parameters in their photon source model used for dose calculation in the PEREGRINE system, Hartmann-Siantar et al (1997) simulated linacs using MCNP and the EGS4/BEAM code (Rogers et al 1995a). DeMarco et al (1998) simulated photon beams from Philips SL-15/25 linear accelerators to obtain the phase space information for the dose calculation in a patient’s CT phantom. Lewis et al (1999) also simulated a Philips SL 75/5 linear accelerator using the MCNP code. Another Monte Carlo system, PENELOPE (Salvat et al 1996), has also been used to simulate photon beams from a Saturne 43 accelerator (Mazurier et al 1999). The EGS4/BEAM code (Rogers et al 1995a) has been extensively used to simulate various types of linear accelerators for photon beams. Liu et al (1997) simulated the photon beams from a Clinac 2100C machine and, by analysing the simulation results, developed an extra-focal source model for the dose calculation using a convolution/superposition method (Mackie et al 1985). Jiang and Ayyangar (1998) simulated a Varian Clinac 1800 accelerator and studied the perturbation effect of the compensator on photon beam characteristics. Their simulation results were also used to investigate the feasibility and necessity of developing a radiosurgical Monte Carlo treatment planning system (Ayyangar and Jiang 1998) and to study the dose perturbation caused by high-density inhomogeneities in small radiosurgical beams (Rustgi et al 1998). A more detailed report on the simulation of clinical photon beams using the EGS4/BEAM code was given by Sheikh-Bagheri (1998).

Most recently, Balog et al (1999) studied the multileaf collimator (MLC) interleaf transmission by simulating the NOMOS MIMiC MLC attached to a GE Orion 4 MV linear accelerator. Their results were incorporated into the design of an MLC used in a prototype tomotherapy machine (Mackie et al 1993, 1995). By simulating the treatment head for a Siemens MXE accelerator, Faddegon et al (1999) designed a new flattening filter for the 6 MV photon beam for this machine. Verhaegen et al (1999) applied the EGS4/BEAM code to the simulation of radiotherapy kV x-ray units. Some of the above work has been reviewed by Mohan (1988) and Andreo (1991). Ebert et al (1996) also reviewed some recent work on the modelling of both clinical photon and electron beams using both analytical and Monte Carlo methods.

In this review paper, we focus on the Monte Carlo simulation of clinical electron beams from medical accelerators. Electron beam Monte Carlo dose calculation was considered to require more detailed beam phase space information on the patient surface than photon beams (Mackie 1990). The difficulty was considered to be that, compared with photon beam simulations, the simulation of electron beams was more sensitive to the details of the accelerator model (Rogers 1991). The work on Monte Carlo simulation of the clinical electron beams can be roughly grouped into two categories. The simulations included in the first category were mainly performed over 10 years ago with limited computation power. A medical accelerator had to be simplified to one or two components of most dosimetric significance in order to perform the Monte Carlo simulation within a reasonable timeframe and acceptable accuracy (Berger and Seltzer 1978, Borrell-Carbonell et al 1980, Rogers and Bielajew 1986, Manfredotti et al 1987, Andreo et al 1989). Even more recently, some studies were still carried out in this way to investigate the influence of an accelerator component on the electron beam characteristics, at a relatively low computing expense (Keall and Homan 1994, Ebert and Homan 1995b). The second category includes the Monte Carlo simulation of a complete accelerator (Udale 1988, Udale-Smith 1990, 1992, Kassaei et al 1994). This category has been greatly
enhanced since the OMEGA BEAM system was developed (Rogers et al 1990, 1995a, b, Mackie et al 1990). Since then, the Monte Carlo modelling of medical accelerators has become easier and more systematic. This review is organized to cover these two categories separately. Because of its significance and the large number of applications, the OMEGA BEAM system will be described in detail in a separate section. We also discuss the modelling of Monte Carlo simulated electron beams using different source models, which is one approach to form the link between the Monte Carlo simulation of clinical electron beams and Monte Carlo treatment planning (Ma and Rogers 1995a, b, c, Ma et al 1997a, Faddegon et al 1998, Ma 1998, Jiang et al 1999). Finally, we discuss some existing problems and possible future directions for Monte Carlo simulation of the clinical electron beams generated by medical accelerators.

2. Simulation of the accelerator components

The application of the Monte Carlo technique to the simulation of electron beams has a long history. In the early years, mainly due to the low speed of computers and the relatively simple Monte Carlo codes available, a full simulation of the accelerator geometry was an extremely difficult task. Simulations were usually carried out for treatment heads with great simplification; only one or two components of dosimetric importance, for example the scattering foil or intervening air, in an electron treatment machine head were simulated and the influence of these components on the beam characteristics and dose distributions was studied.

2.1. The work of Berger and Seltzer

Berger and Seltzer (1978) studied the effect of scattering foils on the dose distributions in a water phantom for electron beams with energies from 5 to 40 MeV. A two-step approach was proposed; the accelerator treatment head, simplified as the scattering foil and intervening air, was simulated first and then the simulation results were used for the dose calculation in the phantom. A monoenergetic electron pencil beam was assumed to be incident perpendicularly on the lead scattering foils of thickness between 0.1 mm and 0.6 mm. The effect of 100 cm of air between the scattering foil and the phantom was simulated using a layer of condensed air-equivalent material. The simulation results were recorded at the phantom surface, including the distribution in energy and angle of the electrons and bremsstrahlung photons. The correlation between the energy and direction was ignored, and therefore the energy spectrum and angular distribution were recorded separately. A superposition approach was developed to calculate the dose distribution in the phantom. For both electrons and bremsstrahlung photons, dose kernels for monoenergetic pencil beams were pre-calculated. Then the dose distribution in the phantom was the superposition of different pencil beam dose kernels weighted with different energies, directions and incident positions at the phantom surface. By limiting the summation over directions to certain cones around the central axis, the effect of collimation to finite field size was simulated. Figure 1 shows the energy spectra calculated by Berger and Seltzer (1978) for electrons passing through 0.1 mm lead foil alone or 0.1 mm lead foil plus 100 cm air. It is clear that the effect of intervening air on the electron energy spectrum is significant.

2.2. The work of Borrell-Carbonell et al

Borrell-Carbonell et al (1980) simulated three GGR MeV/AECL accelerators, i.e. Therac 40 Sagittaire, Therac 20 Saturne and Therac 10 Neptune, using a Monte Carlo code based on the condensed history scheme. Both Sagittaire and Saturne machines generated scanned electron beams controlled by electromagnets while in the Neptune machine electron beam flatness was
MC modelling of electron beams

Figure 1. Energy spectra of electrons transmitted through lead foil, or through lead foil plus 100 cm air layer. Electrons with energy $T_0$ are assumed to be incident perpendicularly onto the foil, and $T$ is the energy with which the transmitted electrons emerge from the foil, or from the layer of air behind the foil. The spectra include all transmitted electrons that emerge with directions within a cone of 20° half-angle around the direction of incidence. The curves are normalized to have unit area under each curve (Berger and Seltzer 1978).

produced by a pencil beam incident on a scattering foil. For all the machines, a simplified treatment head configuration was used in the Monte Carlo simulation. A monoenergetic and isotropic point source was assumed at the exit window of the accelerator. The beam defining system was modelled as a diaphragm with a square opening and zero thickness. Therefore, there was actually no simulation of the interactions of electrons with the components of the treatment machine head, except for the intervening air. Also there was no simulation of the secondary particles generated by electron interactions in the accelerator components.

The transport of the particles through air was simulated for a given collimator opening. In their work, Borrell-Carbonell et al (1980) used a two-step approach similar to that proposed by Berger and Seltzer (1978). The simulation results were recorded at the phantom surface, which included the electron energy, direction and position. The electron energy was calculated by subtracting the mean energy loss in air from the initial energy. Using the recorded electron phase space information, the dose distributions in a water phantom were calculated using a superposition approach similar to that of Berger and Seltzer (1978), i.e. by superposition of the pencil beam dose kernels pre-calculated with Monte Carlo simulation.

By comparing the calculated depth dose curves with measured data, Borrell-Carbonell et al (1980) checked their simple model for the treatment head. They found that the monoenergetic source was a good approximation for the scanned beams from Therac 40 Sagittaire and Therac 20 Saturne, but not for scattered beams from Therac 10 Neptune. For very high-energy (25 MeV) beams, they found that scattering from the beam defining system contributed significantly to the depth dose distribution, even for the scanned beams; therefore they concluded that the collimating system could not be approximated as a thin diaphragm.

2.3. The work by Rogers and Bielajew

Rogers and Bielajew (1986) calculated electron depth dose curves in a water phantom using the EGS4 Monte Carlo code (Nelson et al 1985) by assuming point sources of monoenergetic
electrons passing through 100 cm of vacuum. Compared with the corresponding measured data, 
they found that the calculation underestimated the dose near the surface and predicted a steeper 
fall-off beyond $d_{\text{max}}$, the depth of the maximum dose. Trying to explain the discrepancies, they 
modelled the clinical electron beam as a monoenergetic beam of electrons passing through the 
0.0127 cm Ti exit window of an accelerator, the 0.0127 cm Pb scattering foil, 102 cm of air 
and 3.9 cm of styrofoam. The electron energy spectrum near the central axis of the beam was 
scored at the phantom surface. The depth dose curves in the phantom were then calculated for 
incident broad parallel beams with the scored energy spectra. They found that the calculated 
depth dose distributions were improved by using the realistic spectrum. They also noticed that 
the mean energies of the spectra were significantly influenced by the very small low-energy 
tails. The depth of 50% maximum dose and the practical range were not ideal indices for 
the electron mean energy and the most probable energy respectively. Comparisons of dose 
distributions near inhomogeneities between Monte Carlo calculations and measurements were 
reported in a separate paper (Shortt et al 1986), which gave the first clear proof that Monte 
Carlo could handle electron heterogeneity problems correctly.

2.4. The work of Manfredotti et al

Manfredotti et al (1987) used the EGS3 code system (Ford and Nelson 1978) to simulate the 
electron collimator and the intervening air. A source of diameter 2 mm emitting monoenergetic 
electrons of 17 MeV was used in the simulation. The collimator was simulated with a 10 mm 
 thick slab with a square window to obtain a $10 \times 10$ cm$^2$ field at the phantom surface. The 
two-step approach was used; the simulated results, including energy, charge, impact points 
and direction cosines of particles that arrived on the phantom surface, were scored and later 
used in the three-dimensional phantom dose calculations.

2.5. The work of Andreo et al

Andreo et al (1989) studied the influence of energy and angular spread on the depth dependence 
of the stopping power ratio for clinical electron beams using the Monte Carlo method. In a 
similar way to Berger and Seltzer (1978), the accelerator treatment head was simplified as a 
lead scattering foil and 100 cm air. The lead thicknesses were chosen to give energy and angular 
distributions equivalent to those produced by all materials (except for air) within the treatment 
head. The EGS4 code (Nelson et al 1985) was used to simulate a monoenergetic electron 
pencil beam passing through the lead scattering foil and the air volume. The beam collimation 
was simulated by restricting the direction of the electrons emitted from the scattering foil to a 
forward cone, as done by Berger and Seltzer (1978). The electron energy spectrum was scored 
at the phantom surface and then used for the dose calculation in the phantom by combining 
with the Gaussian angular distributions of various root mean square angles. The phantom 
simulation was done using the MCEF Monte Carlo code (Andreo 1980, Andreo and Brahme 
1984).

2.6. The work of Keall and Hoban

Keall and Hoban (1994) calculated the energy and angular distributions for primary electrons 
as well as contaminant bremsstrahlung photons for a Siemens KD2 machine using the EGS4 
code (Nelson et al 1985). Like Berger and Seltzer (1978) and Andreo et al (1989), the 
accelerator treatment head was simplified as the scattering foils and intervening air while 
the influence of other components was ignored. For this type of accelerator, the scattering
A foil system consists of two foils, i.e., the first scattering foil (stainless steel at 6 MeV and gold at other energies) and the second scattering foil (aluminium of a pyramid shape). The transport of an initially monoenergetic electron pencil beam through the scattering foils and air in the treatment head was simulated and the resulting energy and angular distributions were scored at the phantom surface. The simulated electron angular distribution was found to agree closely to those calculated with the Fermi–Eyges theory (Fermi 1941, Eyges 1948, Keall and Hoban 1996, Jiang et al 1998, 1999), indicating that a Gaussian approximation to the incident angular distribution of the primary electron beam might be adequate for use in treatment planning algorithms. The depth dose curve in water for the bremsstrahlung photons alone was calculated using the simulated bremsstrahlung spectrum. They found that the photon component of a 15 MeV electron beam had a similar $d_{\text{max}}$ to the 18 MV photon beam depth dose and similar fall-off to the 10 MV photon beam depth dose. The influence of electron energy and angular spread on the depth dose distribution was also studied. They found that, compared with the measured data with the applicator removed, the calculated depth dose curve using a monoenergetic beam with no angular spread had a lower surface dose and a sharper fall-off. When the simulated energy spectrum was used, excellent agreement in both the build-up and fall-off regions was achieved. The inclusion of the initial angular spread into the phantom dose calculation had a negligible effect on the depth dose curve. However, the initial angular spread was important in determining the penumbral width at the surface.

2.7. The work of Ebert and Hoban

Ebert and Hoban (1995b) performed a detailed study on the effects of electron beam cones and cerrobend cutouts on clinical electron beams using the EGS4 system (Nelson et al 1985). The EGS4 user code they used, RTPCART_XYZ, was based on another user code RTPCART (Murray 1990) with elements of XYZDOS (Rogers and Bielajew 1990, Bielajew and Rogers 1992). The energy and angular characteristics of Siemens KD-2 electron beams were taken from the Monte Carlo simulations of Keall and Hoban (1994).

In their study, simulations were performed for a variety of beam energies (6, 12 and 18 MeV) and beam angles incident on elements of the trimming plates or cutouts. The simulated element was a semi-infinite edge of the applicator or cutout in air. The area of the element irradiated by the electron beam was limited by a 5 mm width along the edge, the thickness of the edge and the maximum lateral range of electrons from the edge. They found that there were two processes significant to the dose distribution in the water phantom beneath the edge (see figure 2). One is the scattering of the primary electrons from the vertical face of the edge, the other is contaminant photon transmission through the edge. Other processes, such as transmitted primary electrons, and photons and electrons generated within the edge, etc., are negligible. The effect of the edge on the photon component in the primary beam can be treated as simple attenuation. The electrons scattered from the vertical face of the edge are dominantly forward directed and mostly inside the field. By studying the effect of beam incident angle, they also found that the scattered electron fluence was significantly influenced by the area of the vertical face of the exposed edge; more electrons were scattered from a large exposed area. This finding indicated that cutouts with diverging edges would produce fewer scattered electrons and thus flatter profiles than those with vertical edges. Ebert and Hoban (1995b) also studied the influence of beam energy on the scattered electron fluence, as well as the energy spectrum and mean energy of scattered electrons at various distances from the applicator edge. This information helped them to build a model for applicator scatter (Ebert and Hoban 1995a).
3. Simulation of the complete accelerator treatment head

A full Monte Carlo simulation of the medical accelerator geometry is a very difficult task. A large number of physical data (e.g. component dimensions and materials) need to be obtained and validated with high precision. A large amount of careful coding has to be done to accurately and efficiently model the individual accelerator components. Most of all, the simulation results are very sensitive to the details of the model. Therefore, it requires Monte Carlo expertise and deep understanding of electron beam characteristics to obtain good agreement with measurements.

3.1. The work of Udale/Udale-Smith

This difficult problem was first tackled by Udale/Udale-Smith (1988, 1990, 1992), who did very extensive coding (about 18,000 lines of FORTRAN) based on the EGS4 code system (Nelson et al 1985). Udale (1988) simulated a 10 MeV electron beam from a Philips SL75-20 accelerator. In this work, all the treatment head components of dosimetric importance were modelled, including the exit window, primary collimator, scattering foil, monitor chamber, mirror, movable jaws, accessory ring and applicator. In order to see the effects of different parts of the treatment head on the beam, she simulated five cases (denoted from case A to case E) at various levels of gradually increasing sophistication (see figure 3). An additional part of the beam defining system was introduced into the simulation in each successive case. In case A, a monoenergetic pencil beam was incident directly on the phantom. In case B a
Figure 3. Schematic diagram of five simulation geometries with gradually increased sophistication used by Udale (1988) to study the influence of treatment head components on beam characteristics at the phantom surface. In case A, a monoenergetic pencil beam was incident directly on the phantom. In case B a monoenergetic isotropic point source at 95 cm from the phantom was used and the effect of intervening air was simulated. In case C the simulation was performed for a monoenergetic pencil beam incident on the electron window 95 cm above the phantom and interacting with the scattering foil, ionization chamber, mirror and air. In case D the movable jaws and accessory ring were added to case C and in case E the electron applicator was added.

Some approximations were made in the modelling of accelerator components in the work of Udale (1988). The primary collimator system was described as a series of concentric cylinders. The monitor ionization chamber and primary collimator extended to infinity radially, and movable jaws were semi-infinite. The initial energy of the pencil electron beam at the exit window for a nominally 10 MeV beam was determined by matching the 50% dose depth (R_{50}) and the electron practical range (R_{p}) between calculated and measured depth dose curves. Some variance reduction techniques were employed, such as the range rejection technique, i.e. particles were discarded if they did not have enough energy to reach the scoring plane at the bottom of the treatment head.

The two-step approach was adopted in the simulation of Udale (1988). In the first step, the treatment head was simulated and the simulation results (the particle phase space data) were recorded at a scoring plane at the phantom surface. The phase space information included particle type, energy, angle to the beam central axis and position at the phantom surface, which were stored in a four-dimensional array. In the second step, the dose distributions in the phantom were calculated using the scored phase space data as input.
To reduce the number of storage data about the beam exiting the treatment head and to simplify the phantom simulation, Udale (1988) used a so-called planar approximation. The energy spectrum was integrated over radius and angle, and the angular distribution was integrated over radius and energy. The angular distribution was defined relative to the beam central axis and there was no dependence on the azimuthal angle at any point. Under this simplification, beam divergence and the correlation among particle energy, angle and position were ignored. The beam was considered to comprise identical pencil beams all incident parallel to the beam central axis. Each pencil beam had the same energy and angular distributions. The dose distribution for each pencil beam was radially symmetric with respect to the beam central axis. A circular field was used to approximate the square field. Using the reciprocity principle, the depth dose to a region centred on the central axis with a radius of 0.25 cm was calculated by simulating a beam of radius 0.25 cm and scoring dose deposition in a cylindrical region with a cross-sectional area equal to that of the cylindrical beam. In this way the calculation efficiency was significantly improved, which was crucial at the time when this work was done.

Another technique used by Udale (1988) to reduce the statistical uncertainty and thus to speed up the simulation was to simulate primary electrons, secondary electrons and contaminant photons separately in a water phantom. The resulting dose distributions were scaled according to the relative numbers of each particle type and summed to give the total distribution. An extra benefit of using this approach was that some correlation between particle energy and angle was retained; for primary and secondary electrons energy spectra and angular distributions were used which were significantly different.

By simulating five cases with gradually increasing sophistication, Udale (1988) studied the influence of treatment head components on beam characteristics at the phantom surface. The effect of cone size on the beam characteristics was also investigated. The contributions to the central axis depth dose from the secondary electron component and photon component were calculated. The calculated depth dose curves were compared with the measurement to verify the simulation. The agreement was better for the smaller fields. Range–energy relationship was also checked.

In a follow-up study, Udale-Smith (1992) extended the EGS4 Monte Carlo simulation to two other Philips machines, the SL75-14/N and SL15, in addition to the SL75-20. One of the purposes of this work was to verify the dosimetric improvement of these two new machines over the SL75-20. The major improvement of the SL75-14/N over the SL75-20 included the use of a dual-foil scattering system rather than the single-foil scattering system, a low-mass ionization chamber and the aperture-plate applicator rather than the tubular electron applicator. As in the previous work (Udale 1988), the information on the component dimensions and materials for all machines was supplied by the manufacturer. But there were two pieces of information which could not be obtained from the manufacturer. One was the source spot size. A monoenergetic pencil beam was assumed. The other was the actual energy of the pencil electron beam at the exit window. Udale-Smith (1992) chose the incident energy, which gave depth-dose curves with practical ranges close to the measured values. As pointed out by Udale-Smith (1992), the introduction of empirical procedures was not satisfactory but could not be avoided.

Some important improvements of the simulation techniques were proposed. For the simulation of the SL75-20 machine, the entire treatment head geometry was treated as a single system. It was extremely complicated to modify the geometry. For SL75-14/N and SL15 machines a modular approach was used. Each treatment head component was treated as an independent module. Along the beam direction there was no overlapping between any two neighbouring components. The transport of particles in each module was independent of the geometry of the neighbouring components. This concept of component module (CM) was later adopted and extended by Rogers et al (1995a) in the design of the EGS4/BEAM system.
For the simulations of the SL75-20 and SL75-14/N machines, the planar approximation was used as in the previous work (Udale 1988). Four particle types were scored separately: primary electrons, secondary electrons, photons and positrons. These assumptions were reasonable for the SL75-20 accelerator. However, the simulation of the SL75-14/N machine indicated that they were not adequate for cleaner beams produced by the modified collimating system. Therefore a more realistic model was used in the simulation of the SL15 machine. Five particle types were scored: applicator scattered electrons, jaw scattered electrons, direct electrons (not scattered by the secondary collimating devices), contaminant photons and positrons. For direct electrons, beam divergence and the correlation between energy and direction were taken into account. Five energy bands were used for the direct electrons and for each band there was an appropriate angular distribution. Beam divergence was considered by storing the angle between the electron direction at the scoring plane and a ray from the source incident at the same point on the scoring plane.

The phantom simulations for SL75-20 and SL75-14/N were the same as in the previous work (Udale 1988). The planar approximation was found to be inadequate for SL75-14/N and SL15. Hence, two different approximations were used to include beam divergence into the phantom simulation for the SL15 machine. In one approximation, the central axis depth dose was calculated using the reciprocity principle. Then, an inverse-square correction was applied to the resulting central axis depth dose curve for direct electrons. In the other approximation, a voxel geometry was modelled and then the divergence of the direct electron component was considered explicitly.

The simulations of Udale-Smith (1992) verified the improvement of SL75-14/N and SL15 over SL75-20. For these two new types of Philips accelerator, the beams were cleaner because (a) there were fewer secondary electrons and contaminant photons in the beams, (b) the contaminant photons had much lower mean energy thus contributed less to the dose distributions, (c) there were fewer low-energy electrons in the electron energy spectra, and (d) the electron angular distributions were narrower.

Udale-Smith (1992) benchmarked the simulations by comparing the calculated and measured depth dose distributions. It was found that the introduction of some correlation between energy and direction for direct electrons improved the calculation accuracy. Allowing for the beam divergence also had a beneficial effect. The use of a simple inverse-square correction was shown to be a useful first-order estimate of the effect of beam divergence, allowing quick calculations to be performed and then corrected. Using voxel geometry for phantom simulation took into account the beam divergence explicitly and was shown to give excellent results. It was a useful starting point for the development of CT based Monte Carlo treatment planning as the patient body can be approximated as voxel geometry built from CT numbers.

### 3.2. Work by other investigators

In addition to the work of Udale-Smith (1990) and the development of the BEAM system (Rogers et al 1995a), there were also some other efforts to simulate a complete electron medical accelerator. To evaluate changes in the electron energy spectrum, depth dose and dose profile for two cone designs for Varian Clinac 2100C accelerators, Kassaee et al (1994) performed Monte Carlo simulation for the treatment head using the cylindrical symmetry CYLTRAN option in ITS Monte Carlo code (Halbeib and Mehlhorn 1984). The treatment head components modelled were scattering foil, photon jaws, electron cones and intervening air. Cones were modelled as a series of circular apertures. Energy spectrum and beam profile were investigated for a 9 MeV monoenergetic electron beam impinging on the 0.025 mm thick Be exit window. Like Udale (1988), to study the influence of different components and
the difference of two cone designs, Kassaee et al (1994) simulated four cases of different geometrical arrangements of the beam defining system, which were (a) scattering foil only, (b) scattering foil and photon jaws, (c) scattering foil, photon jaws and new cone design, and (d) scattering foil, jaws and old cone design. It was found that the presence of the movable jaws did not significantly modify the electron energy spectrum. The low-energy secondary electron component of the beam exiting the new cone was significantly reduced because the trimming plates were thick enough to absorb all the incident primary electrons, which was verified by Rogers et al (1995a) and consistent with the simulation results of Ebert and Hoban (1995b). The dose in the beam penumbral region was lower for the new cones because the leakage of secondary electrons from the cone aperture and housing was appreciably reduced.

Burns et al (1995) briefly mentioned a full Monte Carlo simulation of an accelerator treatment head to determine the electron stopping powers and practical ranges for clinical electron beams. An EGS4 user code was developed to model the scattering foil, the collimating system and the intervening air. Unfortunately, no further details were given in the paper.

More recently, Pawlicki (1998) modelled a Varian Clinac 1800 accelerator using the geometry package PENGEO of the PENELOPE Monte Carlo code system (Salvat et al 1996). He combined the accelerator simulation with the patient dose calculation in a 3D rectilinear phantom built from CT data.

4. The OMEGA BEAM system

The work of Udale/Udale-Smith (1988, 1990, 1992) brought the Monte Carlo simulation of electron accelerators to a new stage. Her simulation was no longer restricted to cylindrical symmetry. However, without a large effort, her code could hardly be applied to other types of accelerators, although it was well structured. As pointed out by herself (Udale 1988), ‘In many ways the simulation has been very crude’. In order to supply accurate clinical beam information from a wide variety of medical accelerators for electron beam Monte Carlo simulation, a better code system with greater flexibility and extensibility was needed. This led to the development of the BEAM system (Rogers et al 1995a, b). Because of the significance of the BEAM code, it is worthwhile devoting this entire section to a description of this code system and its applications.

4.1. Descriptions of the BEAM code

The OMEGA (Ottawa Madison Electron Gamma Algorithm) project was a collaborative effort between the University of Wisconsin, Madison, and the National Research Council of Canada, Ottawa (Mackie et al 1990, 1994, Rogers et al 1990). The goal was to develop a full 3D electron beam treatment planning system using the Monte Carlo simulation technique to calculate dose distributions in a patient. A major achievement of this project was the development of the BEAM and DOSXYZ code systems (Rogers et al 1995a, b, Ma and Rogers 1995a, c, Ma et al 1995). Both BEAM and DOSXYZ were based on the PRESTA extension of the EGS4 Monte Carlo system (Nelson et al 1985, Bielajew and Rogers 1987, Rogers and Bielajew 1990, Bielajew et al 1994).

Although the original emphasis was on the simulation of clinical electron beams from medical accelerators, the BEAM code was equally applicable to all other radiation therapy units, including high-energy photon beams from accelerators, 60Co beams or kilovoltage x-ray beams. The code was also a versatile, general-purpose Monte Carlo transport package, which could be used in a wide variety of applications besides simulating radiation therapy beams. There were many important and useful new features in the BEAM code (Rogers et al 1995a).
These included the use of the component modules (CMs), scoring particles’ phase space information, tracking each particle’s history, applying various variance reduction techniques, setting up a file/structure for parallel processing and developing a user-friendly interface.

The concept of component module was used in the code design of Udale-Smith (1992). In the BEAM code, this method was extended and played a very important role (Rogers et al 1995a). Component modules were actually a variety of elementary geometric entities and could be used to represent the components of an accelerator. Each CM dealt with a specific class of geometric shape and was contained between two planes, which were perpendicular to the beam axis. No overlapping between CMs was allowed. Each CM operated completely independently of the other CMs. A CM was defined with a variety of parameters rather than explicit values related to the geometric shape and material type. The parameter values were specified in an input file given by the user to model a specific accelerator component when performing a simulation. Figure 4 shows the diagrams for nine different CMs.

The beauty of this element of the design philosophy of BEAM was that each CM acted like a ‘brick’ and the model for the whole treatment head could be easily built up by simply putting a series of CMs together according to the technical drawing of an accelerator treatment head. This feature allowed accelerators of different manufacturers and different designs to be modelled at any level of complexity with little effort, even for users without much Monte Carlo coding experience. Also, this feature reduced the chance of making mistakes during the modelling process and made the quality assurance easier. The independence of each component of the geometry package allowed each CM to be tested and debugged in isolation.

Another important element of the basic design philosophy in the BEAM code was the recording of the complete phase space information of each individual particle crossing the scoring planes (Rogers et al 1995a). The information included the charge, energy, position, direction and a history tag (see below) for each particle. The phase space data could be recorded at the back plane, perpendicular to the beam axis, of any specified component module in the accelerator model. There could be an arbitrary number of scoring planes and a phase space data file could be created at each plane.

An important parameter in the phase space data file was the history tag variable, LATCH, which was a feature of EGS4 (Nelson et al 1985). In the BEAM code, however, each bit of LATCH was manipulated separately to fully utilize this tag. Using LATCH one could easily keep track of each particle’s history and the information to score dose components separately, and then analyse the relative dose contributions from various accelerator components. In order to evaluate the influence of different accelerator components, Udale (1988) and Kassaei et al (1994) simulated several cases of different levels of complexity. With LATCH, it can be done in one simulation (Rogers et al 1995a). The phase space data file could be analysed using various data analysis programs (Ma and Rogers 1995a, b, c). LATCH provided a powerful tool for studying the characteristics of clinical electron beams with these programs.

A phase space file could be re-used by the BEAM code itself (Rogers et al 1995a). This allowed the user to simulate a treatment head in separate steps. For example, the user could simulate the treatment head from the exit window down to the upper surface of the last trimming plate of the electron applicator and then re-use the recorded phase space data to simulate the last scraper with various patient-specific cutouts. For an energy/cone combination, the first step of the simulation could be done once and for all, and thus avoid the repetition of the simulation for the subsequent simulation for different electron cutouts (Zhang et al 1998, 1999, Kapur et al 1998, Jiang et al 1999). The phase space data could also be used by the DOSXYZ code (Ma et al 1995, Rogers and Bielajew 1990, Bielajew and Rogers 1992), which was designed to perform dose calculations in a 3D rectilinear geometry to simulate various phantoms and the patient geometry built from CT data.
Figure 4. Schematic diagram of some component modules developed for the BEAM code by Rogers et al (1995a, b): (a) CONS3R, (b) CONESTAK, (c) FLATFILT, (d) CHAMBER, (e) PYRAMIDS, (f) APPSQ (which was later changed to APPLICAT), (g) JAWS, (h) MIRROR, (i) XTUBE.
The two-step approach was a common technique for simulating accelerator treatment head and performing phantom dose calculations (Berger and Seltzer 1978, Borrell-Carbonell et al 1980, Manfredotti et al 1987, Udale 1988, Udale-Smith 1990, 1992). For disk space and other reasons, the correlation between particle energy, direction and position was usually broken, which led to inaccuracies in the reconstructed incident beam as demonstrated by Andreo and Fransson (1989) and in the dose distributions as shown by Udale-Smith (1992). BEAM also used the two-step approach. However, since the full phase space information for each particle was recorded, the correlation was completely retained. Actually, because BEAM used the phase space data file, this two-step approach was almost equivalent to a complete simulation of the treatment head and the phantom; particles were only ‘frozen’ temporarily at the scoring plane, with all the information to continue for further BEAM simulation or performing phantom dose calculation. The approach of scoring a particle’s full phase space information was proved to be not only flexible but also, and even more importantly, a solution to the correlation problem.

The major drawback with the approach was that it generated large data files and thus took a large amount of disk space, hundreds of megabytes for an electron beam and several gigabytes for a photon beam (Rogers et al 1995a, Kapur et al 1998).

Some useful variance reduction techniques were used in the BEAM code (Rogers et al 1995a, b). One was the range rejection method. In the BEAM code, however, this technique was further improved. Tables of the residual ranges to the threshold energy (AE) in each medium as a function of electron energy were pre-calculated using the restricted stopping powers. If a charged particle could not escape from the current region or reach a scoring plane, it was terminated. Other variance reduction techniques used in BEAM included photon interaction forcing, bremsstrahlung splitting and Russian roulette. These techniques have been described by Bielajew and Rogers (1988) and can be easily implemented in an EGS4 user code using the Mortran macros included in the EGS4 distribution. When a photon is forced to interact in the phantom the weight of the photon and all its descendents will be reduced to reflect the actual probability for such an interaction in the phantom. This photon forcing method worked for dose calculation in the phantom but failed for fluence scoring, as those photons, which did not interact in the phantom, were not transported. Ma (1992) improved the photon forcing technique by forcing the photons to interact not only in the entire phantom but also in any selected regions of interest. A fictitious photon that carried the remaining weight of the incident photon was transported to ensure the accuracy of both dose and photon fluence calculations (Ma 1992). This new photon forcing technique was implemented in the
Figure 5. The treatment head of the Varian Clinac 2100C as shown by EGS\_windows. About 100 electrons coming from the exit vacuum window are simulated and their tracks are shown in the picture (Rogers et al. 1995a).

BEAM code and the photons could be forced to interact in any accelerator component as required (Rogers et al. 1995b). The Russian roulette technique was recommended for use with the bremsstrahlung splitting technique. In a photon beam simulation, the efficiency of the bremsstrahlung photon production can be significantly improved using the particle splitting technique. However, fewer electrons generated by these photons in the treatment head can reach the scoring plane. The simulation efficiency can be significantly improved by applying the Russian roulette technique to these electrons.

Recently, a user interface program, BEAM\_GUI, has been written which has greatly simplified the process for writing BEAM input files. A similar user interface was also developed for the DOSXYZ program (Treurniet and Rogers 1998). This has made the BEAM system more user-friendly. Furthermore, the code can be used to score data to produce a 3D display of the accelerator geometry and particle histories using a graphics package called EGS\_windows (Bielajew and Weibe 1991). Figure 5 shows a Varian Clinac 2100C machine using EGS\_windows and the simulated tracks of some electrons and photons. A file structure can be set up to run the dose calculation in parallel. Different input files can be generated for the same beam and phantom configuration but with different random number seeds to ensure the independence of the individual results. The final results can be combined using the software provided. Many efforts were also made to document the code well, especially in the source code (Rogers et al. 1995b, Ma et al. 1995, Ma and Rogers 1995a, b, c).
4.2. Benchmarks and applications of the BEAM code

4.2.1. The work by the NRCC group. The BEAM code has been first benchmarked extensively for electron beams from medical accelerators and used in a wide variety of applications by its developer, the NRCC (National Research Council of Canada) group.

Rogers et al (1995a) compared the calculated dose distributions with measurements in a homogeneous water phantom for a wide variety of accelerators, including the NRC 35 MeV research accelerator, a Varian Clinac 2100C, a Philips SL75-20, an AECL Therac 20 and a Scanditronix MM50. The NRC research accelerator has a narrow, well-known electron energy. They found that the calculated and measured depth dose curves and dose profiles were in excellent agreement using no free parameters. The energy and angular distribution of electrons at the exit window of a commercial medical accelerator is usually unknown to the user. Like Udale-Smith (1990), Rogers et al (1995a) assumed that all beams in the accelerator vacuum were monoenergetic and on-axis, and selected the incident electron energy by matching the calculated and measured values of the 50% dose depth, $R_{50}$. The depth dose curves could be calculated with an accuracy better than 2% of $D_{\text{max}}$ for almost any beams simulated. A more detailed benchmark was given in a compilation by Ding and Rogers (1995). In addition to those five linear accelerators studied by Rogers et al (1995a), Ding and Rogers (1995) also simulated a Siemens KD2 machine. For each accelerator, they calculated the dose distributions in a water phantom for up to five electron beams with an energy range of 5–50 MeV. For all cases, the calculated and measured dose distributions agreed well.

A detailed study on electron beams from a Varian Clinac 2100C machine was also presented by Rogers et al (1995a). The contributions from electrons and photons from different parts of the accelerator to the depth dose curve and dose profiles were analysed. Consistent with the results of Kassaee et al (1994), they also found that the new applicator design reduced low-energy electrons from the applicator and gave a significantly cleaner beam. As a substitute to the new cone design, they found that an equivalent dosimetric effect could be achieved by adding an additional piece of lead on top of an aluminium scraper. They found that the dose distributions were sensitive to the details of the accelerator. For example, for a Clinac 2100C machine, either using a different monitor chamber or changing the jaw setting slightly would result in distinct changes in the dose distributions.

Ma et al (1993a, 1994) used the BEAM code to study the characteristics of the electron beams from three different linear accelerators, Philips SL75-20, AECL Therac 20 and Varian Clinac 2100C. Electron beams of nominal energies 6–20 MeV were simulated using the BEAM code. The results showed that for a Varian Clinac 2100C accelerator, electrons from individual components had distinct energy and angular distributions. The normalized energy spectra for each component varied by less than 5% for the same energy bin within a collimated 10 × 10 cm² field. About 10% of the electrons from the applicator are incident at relatively large angles, peaked at 6 and 15 degrees. These observations led to the development of simplified beam models to replace the simulated phase space data (Ma and Rogers 1995b, c).

A key parameter in the simplified beam model is the virtual point source (VPS) position. Ma et al (1997a) simulated electron beams emerging from the treatment head and calculated the dose distributions in a water phantom with varying air gap thicknesses. Dose distributions were also calculated for an ideal point source with various SSD values in order to verify several commonly used methods for VPS position measurements. The results showed that the ‘pin-hole’ method and the ‘full-width half-maximum’ (FWHM) method could accurately predict the source positions while the ‘inverse square’ method usually gave smaller SSD values, especially for lower electron energies. The virtual SSD values obtained using the first two methods were suitable for direct Monte Carlo or analytical calculations, with a point source model, of dose
distributions with correct FWHM values. Based on these studies, Ma and Rogers (1995b, c) and Ma et al (1997a) proposed a multiple source model consisting of a series of sources based on the energy and angular distributions of individual accelerator components. The user could predetermine the complexity of these models so that the full phase space data for these beams could be reproduced with the desired accuracy.

Using the BEAM code, the energy and angular spectra of electrons and photons could be easily obtained. Rogers et al (1995a) calculated 18 electron spectra for four different clinical accelerators, including a non-standard Varian Clinac 2100C with a Type II applicator, Philips SL75-20 with a tubular applicator and the scanned beams from the Therac 20 and MM50 Racetrack Microtron. The results for the SL75-20 accelerator were compared with those of Udale-Smith (1990). Some differences were seen for a 10 MeV beam and could be partially attributed to the different values of AE and cutoff energy, as well as different regions from where electrons were included in the spectra. Figure 6 shows the detailed analysis of the components of the electron energy spectrum, angular distribution, depth dose distribution and dose profile for a 9 MeV beam from a Varian Clinac 2100C accelerator. These results showed that most large-angle electrons were contributed by scattered electrons from applicators while scattered electrons from jaws had a relatively smaller angular spread for beams from accelerators with scattering foils. They also found that the contaminant photons were forward directed with a small angular spread because they were mainly produced in scattering foils.
Using the BEAM simulation results for a variety of clinical accelerators, Ding et al (1996) studied the mean energy in a water phantom for realistic electron beams from a variety of accelerators. They calculated the Spencer–Attix water/air restricted mass stopping power ratio for realistic electron beams in the energy range from 5–50 MeV and compared the results with those determined according to AAPM and IAEA protocols based on monoenergetic parallel beams (Ding et al 1995). A universal correction to the stopping power ratios for electron beam reference dosimetry was presented in terms of per cent depth dose in the photon tail. Ding et al (1997) also performed detailed studies of fluence correction factors using the realistic beam data simulated with the BEAM code.

The BEAM code has also been used to study the beam monitor chamber backscatter effect for electron beams. Ding et al (1994) found that, for a Clinac 2100C machine, the amount of backscatter decreased from 8% to 3% as the beam energy went from 6 to 18 MeV. They also found that 99% of the backscatter came from a fixed shielding ring downstream of the monitor chamber in the Clinac 2100C. The rest came from the collimator and applicator, and almost none from the water phantom. Using the BEAM code, Zhang et al (1998) studied the effects of changes in stopping power ratios with field size on electron beam relative output factors. They simulated electron beams for different field size, energies and machines and then calculated the stopping power ratios. It was found that this effect was negligible, due to error cancellation, as long as the AAPM TG-25 protocol (AAPM 1991) was followed with stopping power ratio data for broad monoenergetic beams (AAPM 1983). Zhang et al (1999) calculated the output factors versus size of square cutouts for electron beams with energies between 6 and 13 MeV and from a Siemens MD2 machine. The calculated relative output factors at 100 cm and 115 cm SSD agreed within 1% with the measurements.

4.2.2. The work of the Stanford group. The research group at Stanford University has been actively working on the clinical implementation of the Monte Carlo technique for radiotherapy treatment planning (Ma et al 1999). They have investigated the various aspects in the implementation procedure: simulation of electron beams from clinical accelerators (Kapur et al 1998, Ma et al 1999), beam characterization and modelling for dose calculations (Ma 1998), beam commissioning procedures for clinical implementation (Ma et al 1997c, Jiang et al 1999), and calculations of electron output factors for small fields (Kapur et al 1997, 1998, Ma et al 1997b) and extended air gaps (Mok et al 1997, Ma et al 1999). Further studies were carried out to calculate the dosimetric data required for performing accurate dose measurements for beam commissioning (Kapur and Ma 1999). More studies were reported on the effect of heterogeneous patient anatomy on the dose distributions and comparisons with conventional dose algorithms (Ma et al 1999).

Ma et al (1999) reported on the simulation of the electron beams (4–20 MeV) from three types of Varian accelerators, namely Clinac 1800, 2100C and 2300C/D, using the BEAM code. The dose distributions in a solid water phantom were calculated using the DOSXYZ code (Ma et al 1995) and compared with measured beam data according to specially designed commissioning procedures (Ma et al 1997c). The accelerator simulation was refined by adjusting the incident energy and simulating the details of the accelerator components until the difference between the Monte Carlo results and measurements was reduced to within 2% of the maximum dose value. The Monte Carlo calculated isodose curves agreed well with the measured data (using film and ionization chamber) while the results from a commercial treatment planning system with a 3D pencil beam algorithm differed significantly in some typical cases (e.g. >5 mm shift in the depth of the maximum dose, \(d_{\text{max}}\), and the 90% isodose line). The Monte Carlo method accurately predicted the beam output (agreed to within 2% with the measured value) for small irregular fields while the film
measurements at a fixed $d_{\text{max}}$ resulted in an uncertainty of up to 10% for a 6 MeV electron beam.

Mok \textit{et al} (1997) studied the air gap factor for electron fields at extended source distances, the air gap factor being defined as the ratio of the machine output for a field defined by a cutout at an extended SSD to the output of the same field at the standard SSD. They used the BEAM code to simulate the phase space data for the electron beams (6–20 MeV) from a Varian Clinac 2100C accelerator and the DOSXYZ code to calculate the dose values in a water phantom. Comparisons of the Monte Carlo calculations and the measured results showed that the air gap factors could be accurately predicted (within 2%) for square fields of sizes from $10 \times 10$ cm$^2$ to $2 \times 2$ cm$^2$. Similar results were found for elongated fields and irregular fields of various sizes (Mok \textit{et al} 1997, Ma \textit{et al} 1999). They concluded that when set up properly the Monte Carlo program could be used for output factor calculations for small and irregular electron fields by a dosimetrist or a medical physicist with little Monte Carlo knowledge.

Ma \textit{et al} (1997c) reported on the procedures for commissioning the computer simulated clinical beams for the implementation of the Monte Carlo technique for treatment planning. The EGS4/BEAM system was used for the simulation of the beams from several clinical accelerators: Varian Clinac 1800, 2100C and 2300C/D. The EGS4/DOSXYZ code was used for the dose calculations in water and in patients (phantoms built from CT data). The simulated beam data were compared with measurements performed using ionization chambers, diode detectors, TLDs and film. The dosimeter readings were converted using the stopping power ratios calculated for the realistic beams (Kapur and Ma 1999) and the dosimeter perturbation effects were properly corrected. They were able to keep the difference between the calculated and the measured dose, anywhere in the phantom, to about 2% of $D_{\text{max}}$. The output factor (OF) calculation was more sensitive to the details of the linear accelerator; the photon jaw positions in an electron beam could significantly change the OF values which had to be simulated accurately to match the measured data to within 2%. They have developed procedures and successfully commissioned the beams used at the Stanford Medical Center. As a result, the uncertainty in the OF for small irregular field electrons was reduced from 5–10% to about 3% with Monte Carlo treatment planning.

Output factors for electron beams from a Varian Clinac 2100C accelerator were investigated systematically by Kapur \textit{et al} (1998). Electron beams of 6–20 MeV nominal energy for rectangular and square inserts varying in field size from $1 \times 1$ cm$^2$ to $20 \times 20$ cm$^2$ were simulated using the Monte Carlo calculated phase space files. The calculations for water phantoms yielded values that were consistent with the measured values to within 2% for the range of applicator and square/rectangular insert combinations used on these machines for all the beam energies. The variations of output factors for different applicator–insert combinations were quantitatively assessed by separating the contributions of the direct and indirect particles to the total output factor. Beam characteristics such as the energy spectrum, planar fluence and angular distribution were studied to correlate the observed trends qualitatively. The use of these phase space files to compute output factors for arbitrarily shaped electron cutouts such as those used clinically for electron boosts in breast or head-and-neck cancers was examined. It was found that Monte Carlo calculations reproduced the measured values in water phantoms more accurately than a commercial treatment planning system which uses a 3D pencil beam dose calculation algorithm (Hogstrom \textit{et al} 1981), especially under the conditions of compromised lateral scatter equilibrium.

Kapur and Ma (1999) also studied the restricted mass collision stopping power ratios for clinical electron beams of 6–20 MeV nominal energy from a Varian Clinac 2100C accelerator. The Monte Carlo simulations used for this purpose were performed for a variety of beam geometries including narrow to broad beams, normal to oblique angulation and various SSDs.
for homogeneous water phantoms as well as heterogeneous phantoms, for various detector materials. The homogeneous water phantom calculations in this energy range were consistent with protocol recommended values for water-to-air ratios under broad, normal incidence conditions (AAPM 1983). The ratios for narrow realistic clinical beams differed by up to 1% compared with the broad realistic clinical beams, the differences being most significant for water-to-air, film materials and silicon and much smaller for water-to-detector materials such as graphite, ferrous sulphate, PMMA, polystyrene and lithium fluoride. In heterogeneous phantoms comprising water and layered lung- or bone-like plastic materials, stopping power ratios for water-to-detector were compared with the same in identical sized water phantoms using the stopping power ratio correction factors. These factors were found to vary in depth most for water-to-air, film materials and silicon, and to a lesser extent for the other detector materials described above. The extent of variation depended on the actual construction and geometry of the heterogeneous phantoms.

4.2.3. The work of the T-SRCC group. Faddegon et al (1998) described their work on implementing Monte Carlo for electron beam treatment planning at the Toronto-Sunnybrook Regional Cancer Center (T-SRCC). Beam data for their current clinical implementation consisted entirely of full phase space data sets simulated using the EGS4/BEAM code (Rogers et al 1995a). All 14 beams on a Siemens MXE (5–14 MeV) and a Philips SL20 (4–20 MeV) were commissioned for two applicators, covering nominal 10 × 10 cm² and 15 × 15 cm² fields. After spending substantial effort in commissioning, they still had unresolved discrepancies of 3% (or 2 mm in isodose lines) in the measured and calculated dose distributions (Faddegon et al 1998). They considered it likely to be sufficient to weight the beam data with some simple, slowly varying functions of energy and position to improve the calculation accuracy from a clinical perspective. Preliminary work on deriving beam data from measured dose distributions was done using a simple beam model of a point source with spectral distribution (Faddegon 1995). An unfolding technique was utilized to extract appropriate weights for a response matrix of monoenergetic depth doses to yield the measured depth doses. The unfolded spectral distribution had a 50% excess of low-energy electrons to compensate for the assumption of a point source with no angular distribution about the ray lines. Further work was proposed to resolve the level of detail and accuracy required in a beam model to adequately represent a beam, and to determine which concise set of measurements and methodology was most clinically viable for determining the beam model parameters (Faddegon et al 1998).

4.2.4. The work of Karlsson et al. Karlsson et al (1999) investigated the electron beam characteristics for a Varian Clinac 2300C/D accelerator. The purpose was to find out how a conventional treatment head could be modified to use multileaf collimated electron beams. Using the computer controlled multileaf collimator (MLC) for both photon and electron fields, automatic setup would be possible and complex field shapes and beam matching could be achieved. The physical parameters investigated in this work were beam penumbra and virtual/effective source position which were considered to be essential in beam matching and for dosimetry calculations. A 9 MeV nominal electron energy beam was simulated using the EGS4/BEAM code for various geometry combinations. Karlsson et al (1999) reported that the beam characteristics would fulfill their preset criteria for clinical use for electron beams of nominal energies down to 9 MeV by replacing the air in the treatment head with helium, changing the primary scattering foil position, using a thinner monitor chamber and lowering the MLC. Another proposal was made based on their Monte Carlo simulations that the electron beam characteristics would still be acceptable by inserting a helium balloon between the MLC and the patient. Thus, no change to the MLC would be necessary.
5. Discussion

During the last two decades, significant efforts have been made in Monte Carlo modelling of the clinical electron beams from medical accelerators. Many major issues have been addressed through detailed Monte Carlo simulations. Such efforts have been facilitated by the development of the BEAM code. Electron beams from various clinical accelerators have been studied more systematically for different purposes by different research groups. The accelerator manufacturers have studied the beam characteristics to improve accelerator design, although they have not published any detailed reports in the literature. The accelerator users have studied the beam phase space information to derive necessary dosimetric data for accurate dosimetry measurement and to perform accurate dose calculations for radiotherapy treatment planning. However, there are still a variety of practical problems associated with the accuracy of the accelerator simulation using the Monte Carlo techniques.

5.1. Simulation accuracy

The most basic information required for a Monte Carlo simulation of a treatment head is the specifications of the accelerator parts, such as their locations, dimensions and materials. Without knowing the component specifications precisely, the simulation results may be of limited usefulness. So far, it seems difficult to get adequate information about a specific accelerator from the manufacturers (Udale-Smith 1992, Rogers et al 1995a). Because of the commercial value of the detailed specifications of the accelerator parts, manufacturers are usually reluctant to provide the information with the necessary details for Monte Carlo modelling. Furthermore, newly purchased accelerators are often adjusted on-site for individual users, for example by selection of scattering foils or tuning of the waveguide, to match the beam characteristics of the machine to be replaced, or the existing machines. The situation may be further complicated due to accelerator repair, improvement or update. Accelerators of the same model may not consist of exactly the same components. For example, the same type of accelerator may have different scattering foils, flattening filters, monitor chambers or applicators.

Another practical problem is that it is difficult for users to know the precise phase space information of electrons at the vacuum exit window, such as spot size and location, electron energy distribution and angular distribution. When performing Monte Carlo electron beam simulations it has been common practice to assume a pencil beam of monoenergetic electrons at the accelerator vacuum window. The initial energy of the pencil beam is determined by matching the calculated and measured values of some dosimetric quantities, say, $R_{50}$ or $R_p$ (Udale 1988, Udale-Smith 1990, 1992, Keall and Hoban 1994, Rogers et al 1995a, Kapur et al 1998, Ma et al 1999). It has been found that the final depth dose curve is not sensitive to the width of the incident energy spectrum, unless the spectrum was very broad (Rogers et al 1995a). This approach has been demonstrated to be practical and adequate for radiotherapy dose calculations. However, the introduction of these empirical procedures does call into question the accuracy of the simulation results, such as the characteristics of the actual beam phase space data, even though the final dose distribution agrees well with the measured data. For example, the calculated energy spectrum is not necessarily an accurate representation of the realistic spectrum. In addition, as pointed out by Rogers et al (1995a), there may be other, more subtle effects, for which this approach may fail. There is a paucity of experimental data on the fluence of clinical electron beams (Deasy et al 1996). Thus, there is still rationale for establishing a benchmark for treatment head simulation based on fluence and dose measurements using a research accelerator with well known source and treatment head details (Faddegon et al 1998).
It is clear that the required accuracy of accelerator simulation depends on the required accuracy of the end point quantities to be investigated. For Monte Carlo treatment planning, the accuracy of the electron phase space data is usually ensured in such a way that the Monte Carlo calculated dose distributions would be consistent with the measured data to within 2% of the maximum dose (Kapur et al 1998, Ma et al 1999). This criterion seems to be practical and adequate as the 1σ uncertainty on the dose distributions measured clinically is usually about 2% following the recommended dosimetry procedures (cf IAEA 1987, AAPM 1983, 1991). The 1σ uncertainty in the dose at a point in a phantom in a routine clinical measurement is more likely to be about 3% (Khan 1994). Experience has shown that agreement better than 1% could be achieved between Monte Carlo simulations and measurements by fine-tuning the parameters used in the accelerator simulations (Kapur et al 1998, Ma et al 1999).

High-precision measurements could be predicted to within 0.3% by Monte Carlo simulations (Ma et al 1993b, Ma and Seuntjens 1997). Such agreement seems to be overkill for clinical accelerator simulations if the dosimetric uncertainty remains at 2–3% in the dose data used for machine commissioning. Faddegon et al (1998) estimated the various uncertainties in electron dose delivery and concluded that an overall accuracy of 5% in dose or 5 mm in the location of the isodose lines, or in shorthand 5%/5 mm, would be achievable if relative doses were calculated to 3%/3 mm (point-to-point variation of dose within a given dose distribution) and the dose in water relative to reference conditions was calculated to 2%/2 mm.

5.2. Simulation time

The major drawback of the Monte Carlo method is the computing time required to obtain an acceptable statistical uncertainty in the simulated quantities. Ma et al (1997a) reported that to achieve a 1% statistical uncertainty in the dose distribution in a water phantom consisting of 1 cm³ cubic voxels, about $1 \times 10^4$ phase space electrons were needed for every 1 cm² area within the field. The number of particles per unit area would increase for smaller voxels to achieve the same statistical uncertainty. Rogers et al (1995a) showed that the ratio of the number of phase space electrons per cm² scored at the scoring plane for a $10 \times 10$ cm² field to the number of incident electrons at the vacuum exit window increased with incident electron energy. For a Varian Clinac 2100C accelerator, this ratio was about 0.0002 for a 6 MeV electron beam and 0.0008 for a 20 MeV electron beam. This means that one has to simulate up to $5 \times 10^7$ histories in the accelerator simulation in order to achieve the specified 1% statistical uncertainty in the dose calculation. More than $10^8$ electron histories will be needed for a 6 MeV, $25 \times 25$ cm² electron field. This may require over 300 h of CPU time on a SGI R4400 200 MHz workstation (Rogers et al 1995a) or a few days of CPU time on a Pentium Pro 200 MHz PC (Kapur et al 1998). The total CPU time required to simulate five nominal electron energies and five applicators will be close to a month. Because of the iterative process of fine-tuning electron incident energy and other accelerator parameters, the overall CPU time for commissioning all the beams properly may be several months. Clearly, most cancer centres do not possess such computing resources or the necessary Monte Carlo expertise.

5.3. Beam characterization and source modelling

An important reason for simulating the treatment head is to obtain the electron beam phase space information needed for Monte Carlo treatment planning. The phase space data generated with the BEAM code contain almost all the phase space information needed (Rogers et al 1995a). However, there are some problems associated with the direct application of such phase space
data in Monte Carlo dose calculations (Ma and Rogers 1995b, c, Ma et al 1997a, Ma 1998, Jiang et al 1999). Since the beam characteristics are usually different, even for the same type of accelerator, it is necessary to simulate each individual accelerator to obtain the phase space data for Monte Carlo treatment planning. The generation and quality assurance of phase space information by simulating the treatment head is not an easy task, even with the BEAM code. It requires both Monte Carlo simulation experience and much more time than that required for commissioning a conventional electron treatment planning system. This places a practical limit on the clinical implementation of Monte Carlo treatment planning in an ordinary cancer clinic.

Another problem is the storage of the Monte Carlo simulated phase space data. It was found that the beam characteristics for different electron applicators may differ significantly due to the variation of the electrons collimated by the movable jaws and scattered from the applicator scrapers (Zhang et al 1998, Kapur et al 1998). This means that separate phase space files have to be obtained for each electron energy/applicator combination. In order to achieve a statistical uncertainty of 1–2% in a phantom consisting of 0.1–1 cm³ voxels, $10^6$–$10^7$ phase space particles are usually required in a Monte Carlo simulation, depending on the electron beam field size (Rogers et al 1995a, Zhang et al 1998, Kapur et al 1998, Ma et al 1999). For a clinical linear accelerator with five nominal energies and five applicator sizes, this means more than $10^8$ phase space particles or a few gigabytes of computer disk space. More disk space is required if photon phase space data are also stored. This represents a significant burden for the limited computer resources at most clinical centres. Obviously a more concise description for the clinical electron beam with sufficient accuracy is necessary for performing routine Monte Carlo treatment planning.

A methodology has been proposed to replace the phase space file with a multiple source model with parameters derived from the simulated phase space data (Ma et al 1993a, 1994, 1997a, Ma and Rogers 1995b, Ma 1998, Jiang et al 1999). Based on the observations that particles from different components of an accelerator have different energy, angular and spatial distributions while the particles from the same component have very similar characteristics, it was considered that the particles from different parts of an accelerator come from different subsources. Each subsource represented a critical component in the treatment head and its geometrical dimensions were determined by the component in question. Each subsource had its own spectral and planar fluence distributions derived from the simulated phase space data. By sampling the particle position on the subsource and on the phantom surface, the correlation between the particle position and incident angle was naturally retained. A special program, BEAMDP, was written to help derive parameters for the source model from the simulated phase space data using BEAM (Ma and Rogers 1995a, c). This multiple source model has been evaluated against the full phase space data for various accelerators and beam setups (Ma et al 1997a, Ma 1998). Figure 7 gives an example which demonstrates the correlation between electron incidence direction and position on the phantom surface as reconstructed using this multiple source model. Figure 8 illustrates that using the multiple source model can reduce the accelerator simulation time by a factor of 10 to achieve the same statistical uncertainty (i.e. requiring a factor of 10 fewer original phase space particles).

A practical commissioning procedure for electron beam Monte Carlo treatment planning was proposed by Ma et al (1997c). The idea was to derive the basic source model parameters from the simulated phase space data and then to modify the source model parameters based on the measured beam data for clinical implementation. Ma (1998) demonstrated the feasibility of this approach using a multiple source model developed previously (Ma et al 1997a). By varying the bin boundaries of the energy spectra for each subsourse, the depth dose distributions calculated using the phase space reconstructed from the multiple source model for one nominal beam energy matched the depth dose distributions for another nominal energy. By varying the
bin boundaries of the planar fluence distribution for each subsource, the lateral dose profiles calculated using the phase space reconstructed from the multiple source model for one field size matched the dose profiles for another field size (Ma 1998). Jiang et al (1999) further improved the methodology for electron beam modelling and commissioning (Ma et al 1997a, c, Ma 1998). They improved the source model representation and reconstruction algorithms and investigated the minimum number of subsources of dosimetric significance to obtain the required beam commissioning accuracy. This method was applied to electron beams from a Varian Clinac 2100C machine with a type III applicator. It was found that a four-source model was adequate for all the energy/applicator combinations. The source model consisted of an electron point source, a photon point source and two extended electron sources. The scoring plane was placed directly above the upper surface of the last scraper of the applicator, rather than the phantom surface. The last scraper or a cutout placed on the scraper was not handled by the source model but simulated separately in a further BEAM simulation or in the patient dose calculation. The model was benchmarked by comparing the dose distributions calculated with the source model and the full phase space data. Figure 9 shows an example for a 20 MeV beam with a 6 × 6 cm² applicator and 120 cm SSD. The chosen Varian Clinac 2100C acted as the reference accelerator. The model established for the reference accelerator was used to commission electron beams from other machines of similar design such as a Varian Clinac 2300C/D by tuning a few parameters in the model. The calculated dose distributions agreed to within 1–2% with the measured data (Jiang et al 1999).

6. Conclusions

We have reviewed the work on the simulation of clinical electron beams from medical accelerators using the Monte Carlo method. It is well accepted that the only accurate
Figure 8. The surface dose profiles along the x-axis calculated using the original phase space data of varying size (a) and using the multiple-source model based on the corresponding phase space data (b) for an 18 MeV electron beam from a Varian Clinic 2100C accelerator. The field is 10 cm × 10 cm defined at 100 cm SSD. The voxel size in the phantom is 1 cm³ for the dose calculation. The number of phase space particles in each of the data sets is given in the figures. $10^7$ particles are reconstructed for the dose calculations in (b) while the phase space data are recycled (Ma et al 1997a).

and practical way to obtain detailed information on a clinical electron beam is the Monte Carlo simulation of the treatment head. Accurate Monte Carlo treatment planning cannot
be performed without accurate beam phase space data. In addition, detailed information about radiotherapy beams has a wide variety of applications in clinical physics and radiation
dosimetry. Monte Carlo simulation of medical accelerators can increase our understanding of clinical beam characteristics, improve accelerator design and also improve the accuracy of clinical dosimetry by providing realistic beam data.

Optimistically, accurate accelerator simulation will result in accurate dose calculation and eventually in accurate dose delivery. Accurate dose calculation also plays a vital role in dose uniformity or conformity in intensity-modulated radiation therapy. It is exceedingly difficult to match the prescribed dose throughout the treatment volume in fields with heterogeneities due to the large dose discontinuities present in such fields. Treatment plans optimized on the basis of overly simplified dose calculation algorithms can become non-physical due to the uncertainties in the beam profiles. With accurate dose calculations, on the other hand, beam modifiers, mixed beams (electrons and photons) and optimization techniques may be used with confidence to improve the uniformity and conformity of the delivered dose distribution to the ideal prescription.

The following summarizes the current status of the work on electron accelerator simulation:

(a) Although Monte Carlo simulation of electron medical accelerators has been performed by a number of investigators who focused on the effect of one or two components of dosimetric significance, more detailed beam characteristics and accurate particle phase space did not become available until the complete accelerator geometry could be simulated accurately.

(b) Since the work of Udale and the development of the EGS4/BEAM system, our knowledge of clinical electron beams has been greatly improved. A large number of scientific findings and important results have been reported based on the EGS4/BEAM studies. However, these studies have only been for a few accelerators and a few beams. Systematic simulations of all the beams on various clinical accelerators need to be carried out.

(c) The availability of electron beam phase space has led to the clinical implementation of the Monte Carlo technique for radiotherapy treatment planning dose calculations. The Monte Carlo simulated beam data have been used as input to the Monte Carlo code for accurate phantom and patient dose calculations. Accurate beam information has also resulted in more accurate dosimetric data.

(d) The computing time required to simulate the beam data from a clinical accelerator is significant. This may not be a problem for a once-and-for-all study on some dosimetric data such as the conversion factors or correction factors for radiation dosimeters but will be a burden for a small cancer clinic after major changes in beam parameters due to machine upgrade, replacement of components or any other reasons. Faster computers and parallel processing methods can be used to speed up the simulation in large research centres. Variance reduction techniques are called for to further reduce the CPU time for accelerator simulation and patient dose calculation.

(e) Accurate simulation of an electron beam from a clinical accelerator requires not only a specifically developed Monte Carlo code but also a great deal of Monte Carlo expertise. As the specifications of the accelerator components are often not known adequately and the machines are also tuned individually at the installation sites (leading to the variation in the incident electron energy), it can be a real challenge to match the Monte Carlo calculated dose distributions with the measured beam data, and dosimeter perturbation effects often need to be well understood to make this possible. Small inaccuracies in the simulation geometry can result in significant changes in the calculated dose distributions. If necessary, accurate energy, angle and fluence distributions should be used for the electrons incident on the vacuum exit window to achieve accurate end-point quantities (such as electron spectrum).
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(f) The phase space data can contain very detailed information about a clinical electron beam. However, the simulation of the phase space of a clinical beam can be time-consuming and it also requires a large amount of disk space. Well-designed beam models can accurately represent the beam characteristics and reconstruct the beam phase space. By using beam models rather than the raw beam phase space data, one can save accelerator simulation time and reduce the disk storage requirement.

(g) It is probably impossible to simulate the beam phase space for all the clinical electron beams at every cancer clinic because of the lack of computer resources and Monte Carlo expertise. However, it has been demonstrated that well-developed beam models can be used to reconstruct any clinical electron beam. A reference machine needs to be simulated accurately to derive the beam model parameters and these parameters can be further adjusted based on the measured beam data for any machines of the same type. Further studies on source modelling and beam commissioning are needed for widespread clinical application of the Monte Carlo dose calculation technique.

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