Monte Carlo simulation of the Tomotherapy treatment unit in the static mode using MC HAMMER, a Monte Carlo tool dedicated to Tomotherapy

E. Sterpin\textsuperscript{a}, M. Tomsej\textsuperscript{b}, B. Cravens\textsuperscript{b}, F. Salvat\textsuperscript{c}, K. Ruchala\textsuperscript{b}, G. H. Olivera\textsuperscript{b}, and S. Vynckier\textsuperscript{a}

\textsuperscript{a} Department of Radiotherapy, St-Luc University Hospital, 10 av. Hippocrate, 1200 Brussels, Belgium

\textsuperscript{b} Tomotherapy Inc., 1240 Deming Way, Madison, Wisconsin 52717

\textsuperscript{c} Facultat de Fisica (ECM), Universitat de Barcelona, Societat Catalan de Fisica (IEC), Diagonal 647, 08028 Barcelona, Spain

esterpin@yahoo.fr

Abstract. Helical tomotherapy (HT) is designed to deliver highly modulated IMRT treatments. The concept of HT provides new challenges in MC simulation, because simultaneous movement of the gantry, the couch and the multi-leaf collimator (MLC) must be simulated accurately. However, before accounting for gantry, couch movement and multileaf collimator configurations, high accuracy must be achieved while simulating open static fields (1x40, 2.5x40 and 5 x 40 cm\textsuperscript{2}). This is performed using MC HAMMER, which is a graphical user interface allowing MC simulation using PENELOPE for various configurations of HT. Since the geometry of the different elements and materials involved in the beam generation are precisely known and defined, the only parameters that need to be tuned on are therefore electron source spot size and electron energy. Beyond the build up region, good agreement (2%/1mm) is achieved for all the field sizes between measurements (ion chamber) and simulations with an electron source energy set to 5.5 MeV. The electron source spot size is modelled as a gaussian distribution with full width half maximum equal to 1.4 mm. This value was chosen to match measured and calculated penumbras in the longitudinal direction.

1. Introduction

Intensity Modulated Radiation Therapy (IMRT) allows better sparing of the healthy tissue while increasing dose conformity and homogeneity to the target [1]-[3]. Helical Tomotherapy (HT) is designed to deliver highly modulated IMRT treatments, yielding significant reduction of dose to critical tissues without compromising target coverage. The irregular fluence is obtained through the optimization process results of the simultaneous movement of the gantry, table and multileaf collimator. The resulting dose distribution is calculated with the treatment planning system developed by Tomotherapy and based on a collapsed cone convolution algorithm [4], [5].
This algorithm has shown great accuracy for simple configurations in inhomogeneous media [6], but the scientific community is still concerned when small fields or a superposition of small fields i.e. IMRT fields, are used, especially in low density inhomogeneities [7] - [10]. Convolution/superposition algorithms are known to be more accurate than older correction-based algorithms, even though deviations were reported at interfaces and in low density regions [8], [11], [12].

Monte Carlo (MC) simulation is considered as the most accurate method to calculate dose distributions in tissue [13]. Moreover, the validity of MC simulation is not limited by the complexity of the treatment, provided that the treatment delivery is simulated in a realistic manner. Therefore, the precise modeling of HT is necessary to evaluate the convolution/superposition algorithm accuracy in complex situations.

HT modeling is divided into three parts. Firstly, the elements of the treatment head involved in the beam generation and broad beam collimation devices (jaws) must be accurately and completely modeled. Secondly, the multileaf collimator (MLC) geometry has to be defined precisely. Finally, the helical mode need to be simulated by calculating, translating and rotating the phase spaces delivered statically according to the pitch setup and the MLC configuration at each time of the treatment. When all these steps are accomplished, dose distribution inside patients for clinically delivered treatments may be calculated and then compared with the convolution/superposition algorithm. The code used for the simulations PENLELOPE [14] which allows the user to determine any geometry configuration provided that all the bodies of the geometry are delimited by quadric surfaces. PENLELOPE incorporates elaborate physical models which have been used for radiotherapy applications [15] and benchmarked experimentally [16] and with other MC codes [17]. A dedicated graphical user interface, called MC HAMMER, for HT simulations was developed.

This paper covers the first part of the simulation, i.e. static fields. The model was commissioned by comparing calculated and measured data in reference conditions. Of course, the highest accuracy must be achieved, since the resulting phase spaces will be used as an input when the helical mode will be turned on. A previous paper was already published on the Monte Carlo modeling of HT [18], principally on radiation characteristics aspects. However, the present work included a more detailed model of HT treatment planning for future MC treatment planning purposes. It was also based on the last HT system and uses PENLELOPE code instead of MCNP.

2. Material and methods

2.1. Tomotherapy treatment unit

HT delivers intensity modulated radiotherapy. The treatment head is composed by a linear accelerator tube (SIEMENS), beam formation elements (target, beam hardener), collimating devices (jaws and primary collimator) and a binary MLC to provide modulation. The head can rotate continuously and simultaneously with the translation of the treatment table (helical delivery), in the manner of a CT scanner. HT can operate in two modes. In the treatment mode, the nominal energy is equal to 6 MV. HT has also the ability to acquire CT scan images with the patient in the treatment position, with a nominal energy reduced to 3.5 MV in that case.

When all the leaves are open, the position of the jaws can be set to define three field sizes, 40x1, 40x2.5 and 40x5 cm² at the isocenter (SSD = 85 cm). Lower field sizes permit higher longitudinal modulation of the dose distribution but at the cost of higher treatment times. Due to the absence of a flattening filter, the transverse profile has a typical cone shape. The binary MLC equipping the system is composed by 64 movable leaves with a projected nominal thickness at the isocenter of 6.25 mm. Apart from those leaves, there are two additional stationary leaves at each extremity of the MLC that affect the beam collimation, even when a static open field is delivered.

2.2. Measurements
Most of the measurements were performed in a water tank, with a SSD set to 85 cm. The dimensions of the water tank were 60x27x27 cm³, where 60 cm refers to the lateral dimension. The detector was an Extradin A1SL ion chamber (Standard Imaging). The collecting volume and its diameter equaled 0.056 cm³ and 4.1 mm. Profiles were measured at several depths (1.5 cm, 5 cm and 10 cm). Lateral profiles refer to profiles measured in the direction where the field size is equal to 40 cm. In the longitudinal direction, the field width is determined by the jaws and may have three values (5 cm, 2.5 cm, and 1 cm). The simulations were compared to a “gold standard” (GS) which was obtained by acquiring profiles and depth dose distribution from several Tomotherapy units, with the Extradin A1SL ion chamber. All the data were then averaged to provide only one depth dose and one set of profiles. Currently, each new released unit is tuned in order to match the GS within 1%/1mm for depth dose and lateral profiles and within 1% of full width half maximum (FWHM) for longitudinal profiles of open fields. Basic measurements of the HT unit of our institution, e.g. depth dose curves and open fields’ profiles, are in agreement with the GS within the same limits. Therefore, the parameters of the MC simulation could be used for both the GS and the HT unit of our institution with the same level of accuracy. It is worthwhile to mention that the choice of the detector was made implicitly, since one of the purposes of the MC simulation is to match the GS.

Some ion chamber designs may be not adequate to measure accurately the dose in the build up region [19]. Even though MC simulations should reproduce as close as possible commissioning data (i.e. the GS, measured with an ion chamber), additional measurements in Solid Water™ were performed using EDR-2 films placed at several depths, perpendicular to the beam axis, to study the build up region. Films were placed each 2 mm from 0.1 to 1.5 cm depth, and each 5 mm from 1.5 to 5 cm depth. EDR-2 films were chosen because of their better resolution and because several studies show that EDR-2 films yield generally accurate results with deviations generally within 2%, for perpendicular [20] and parallel [21], [22] depth dose measurements. High depths (>10 cm) and large fields may lead to substantial deviations because of the potential over-response of the film to scattered photons with lower energies [20], [21].

2.3. Monte Carlo simulation

2.3.1. Geometry definition

Through the PENGEOM subroutine package, PENEOLOPE provides great freedom to the user to define easily complex geometries, allowing an exact reproduction of some unique features of HT, like the jaw collimation system. Since the geometry and the materials were known in detail and PENEOLOPE was able to reproduce it faithfully, no modifications of the geometry should have been introduced further during the tuning and the commissioning processes of the model. Furthermore, the program GVIEW provided with the PENEOLOPE package enabled the verification of the geometry as it was defined in the MC simulation, as illustrated in figure 1.

However, it remained a small uncertainty in the geometry, which was the exact position of the target along the beam axis. When the machine is assembled, the target is mounted up into the linear accelerator, shifting systematically a little the target from its original position, which explains the uncertainty. The order of magnitude of this displacement with respect to the original position is around a few millimeters. The lateral profile width is very sensitive to the position of the target, whereas it is relatively insensitive to the other parameters. Therefore, the position could be accurately determined by matching calculated and measured lateral profile width during a standard iterative process. The last parameters for MC tuning were the energy and the geometric distribution of the electron source. The energy was determined matching cone profiles and percent depth doses (pdd) for the largest field (5 cm). The cone profile shape has a high sensitivity to the electron energy since it depends on the angular distribution of the photons emitted by the target, which is a function of electron energy. The
electron source spot size was determined by comparing calculated and measured penumbras. During all these steps, the acceptance criteria are, beyond the build up region, local dose differences below 2% of dose maximum for all the points, and, in high dose gradients, differences in millimeters had to be similar as the criteria used for the matching of a given Tomotherapy treatment unit with the GS.

Even though the MLC was not included in the simulations presented in this paper, some modeling of the MLC was necessary to account for the two external stationary leaves. The region occupied by the MLC in the treatment head is shown in figure 1.

Dose distributions were calculated in a water phantom, with a voxel size of 2x2x2 mm³. However, some averaging was allowed, depending on which profile was acquired. For depth dose calculations, the voxel size was increased up to 1 cm in the transverse direction. For profiles shown in this paper, the voxel size was increased up to 6 mm in the depth direction and 6 mm in the direction perpendicular to the direction of the profile considered. The number of simulated histories was such that the statistical uncertainty is smaller than 1% (one sigma) of maximum dose for all the voxels.

Two phase-space scoring planes were used in our simulations, one below the jaws and the other one below the beam stopper, as depicted in figure 1. In this paper, the dose calculations were then always performed from the phase space file representing the particle fluence at the second plane.

2.3.2. Particles transport

The transport of particles is simulated by PENELOPE (an acronym for PENetration and Energy LOss of Positrons and Electrons). While photons are tracked using the conventional detailed method (i.e., interaction by interaction), electrons and positrons are simulated using a mixed (class II) scheme. Firstly, hard collisions are simulated individually from the corresponding differential cross sections. Soft interactions between consecutive hard collisions are, however, simulated globally as a single “artificial” event using multiple-scattering approximations. The simulation is controlled by four user-defined parameters: $C_1$, $C_2$, $W_{cc}$ and $W_{cr}$. The parameter $C_1$ determines the mean free path for hard elastic events; $C_2$ sets the upper limit for the average fractional energy loss in a single step. The parameters $W_{cc}$ and $W_{cr}$ are energy-loss cutoff values that separate soft and hard events, for inelastic collisions and bremsstrahlung emission, respectively. Thus, inelastic collisions (bremsstrahlung events) with energy loss larger than $W_{cc}$ ($W_{cr}$) are simulated individually. In the present simulations we adopted the following set of simulation parameters: $C_1 = C_2 = 0.05$ and $W_{cc} = W_{cr} = 10$ keV. These values are somewhat conservative for a routine simulation, but here it was wise to sacrifice simulation speed to ensure reliability of the whole tuning process.
2.3.3. **MC HAMMER**

MC HAMMER is a graphical interface that allows the user to change easily some parameters of the simulation, like the energy, the geometric distribution of the source, the jaws settings … The geometry definition can be visualized in two or three dimensions. Once the parameters of the simulation were defined, phase space files could be calculated and analyzed. In its current version, the interface was configured for static mode and phase space calculations only. Dose calculations were therefore performed using the user code “penmain” in the PENELLOPE distribution. The interface permits to run the simulation on a single CPU where MC HAMMER is installed (standalone job) or on a cluster. In our case, a cluster with 16 computers was used.

3. **Results and discussion**

3.1. **Energy and target position tuning**

The target position was assumed to be correctly determined when the calculated cone profile width agreed with the measured one within 1 mm, as illustrated in the inset in figure 2. When the target is not displaced, the lateral profile width is 5 mm overestimated, which is not the case when the target is shifted 2.5 mm further from the isocenter. Therefore, the distance from source to isocenter is 85.25 cm in our simulation. However, in the rest of the paper, the values of the SSD are given according to the nominal value of the SSD (source to isocenter distance of 85 cm).

The energy was first determined by comparing transverse profiles. When the energy increases, bremsstrahlung photons are preferentially emitted forward at higher energies and, therefore, the ratio between the dose on the central axis and a dose at a given point off-axis is increased (the lateral profile is more “pointed”). The best agreement (within 1%/1mm) was obtained for a monoenergetic electron source with an energy equal to 5.5 MeV, as illustrated in figure 2, where measured and calculated profiles are normalized on the central axis. The energy spread of the electron source was a rather
Insensitive parameter, as stated already by other authors for conventional linear accelerators [13], [23]. For the sake of clarity, MC points were drawn each cm only, except at the edges of the profiles. In figure 3 (a) measurements performed with ion chamber are compared with MC calculations. It is worthwhile to mention that the depth dose curves were scaled such that measurements and MC were equal at 2 cm depth. Outside the build up region, good agreement was obtained with deviations smaller than 1%. In the build-up region, ion chamber measurements deviated substantially from MC results. To resolve this discrepancy, which could have originated from a non adapted ion chamber design for measurements in the build up region, EDR-2 films measurements in Solid Water™ were compared to MC simulations in the same medium. Results are shown in figure 3 (b) where good agreement was obtained between EDR-2 measurements and MC with deviations smaller than 1%/1mm. As the GS, measurements were normalized at depth of maximum dose (1.5 cm) and MC was normalized in the same manner as in figure 3 (a).

The resulting photon energy spectrum of the second phase space file was shown in figure 4. The maximum energy was of course 5.5 MeV, while the average energy equaled 1.4 MeV. Moreover, the peak due to electron-positron annihilation is clearly visible (511 keV peak). The emerging beam consisted mainly of photons, with a negligible contribution of charged particles (electrons and positrons). Indeed, the ratio between the maximum (energy) fluence of charged particles and the maximum fluence of photons (excluding the annihilation peak) was around 0.25%.

Good agreement was also obtained for other field sizes, but, for conciseness, these results are not shown in this paper but will be included in a forthcoming publication.

Figure 2  Lateral profiles comparison between Monte Carlo simulation (MC, open circles) and the gold standard (GS, solid line) for a 40x5cm² field. The profiles are acquired in a water phantom at 1.5, 5 and 10 cm depth and SSD = 85 cm. In the insets, measured (GS, solid line) and calculated profiles (MC) at 5 cm depth, SSD = 85 cm are compared for different positions of the target in the simulation (with tuned displacement, open circles, without displacement, open squares)
Figure 3  40 x 5 cm² field depth dose comparison between MC simulations (open circles) and measurements (GS, solid line; EDR-2, crosses).MC calculations is firstly (a) compared to ion chamber measurements (MS, solid line) in a water phantom (SSD = 85 cm) and secondly (b) to EDR-2 film measurements in Solid Water™ (crosses) (SSD = 85 cm). In each figure, MC simulations are performed according to the experimental conditions (water phantom or Solid Water™ phantom).

Figure 4 Total spectrum of the second phase space file. The maximum energy is equal to 5.5 MeV and the average energy is around 1.4 MeV.

3.2. Jaws settings and electron source geometry

In the simulation, the position of the jaws must be accurately defined according to the GS in the longitudinal direction for the three field sizes. This position was determined comparing the FWHM at a given depth (5 cm). The results are shown in figure 5. The calculated and measured FWHMs agreed within 1% of FWHM.

The penumbra depends on the electron source spot size and the active volume of the detector shape used in the measurements. It can be quantitatively defined as the distance between 80% and 20% of
the dose at the central axis. The active volume of the ion chamber has a diameter of 4 mm. To reproduce partially the averaging effect of the chamber, each point in the MC curve represents the average of a given number of neighboring voxels, which corresponds to an active volume of 8 mm³ times the number of such voxels since the original voxel size is 8mm³. Of course, this is an approximation, since the ion chamber is cylindrical whereas the voxels are cubic. To illustrate the effect of both electron source spot size and voxel size, figure 6 shows the calculated dose profile for a point source, and with a gaussian distribution (FWHM = 1.4 mm) with voxel sizes of 2, 4 and 6 mm in the longitudinal direction. As expected, increasing voxel size and electron source distribution lead to a larger penumbra and a smoother fall-off. The best agreement was achieved for FWHM = 1.4 mm and a voxel size of 4 mm in the longitudinal direction, which corresponds to the diameter of the ion chamber, as clearly shown in figure 6.

Figure 5 Comparison between measured (GS) and calculated (MC) longitudinal profiles, for the three static field sizes available in the tomotherapy treatment unit (5cm, 2.5 cm, 1 cm). The profiles are acquired at SSD = 85 cm and 5 cm depth. The electron source (E=5.5 MeV) has a gaussian geometric spread with FWHM = 1.4 mm.

Figure 6 Comparison between calculated (MC) and measured (GS) penumbras (longitudinal profile), for the 40x5 cm² field. The profiles were acquired at 5 cm depth, SSD = 85 cm. The effects of both the electron source spot size and the voxel size are illustrated.
4. Conclusions

PENELOPE is well suited to model the unique features of HT because of the freedom let to the user to define new geometries. Thanks to the detailed definition of the geometry, only three parameters needed to be determined: the energy, the spot size of electron source and the target position. The typical cone profile shape, due to the absence of flattening filter, was found to be sensitive to the energy of the primary electron beam and the target position. Percent depth dose and longitudinal profiles were also calculated in water phantom and compared to measurements. Good agreement was achieved with deviations below 1 mm and 2% of dose maximum.

Collimation by the jaws was also simulated accurately for all the field sizes. Particles can be stored in a phase space placed below the jaws, thus yielding to a significant reduction of computation time, since calculations through the beam formation elements and the jaws are not necessary anymore. Therefore, the phase space files for the three field sizes can now be used for more complex configurations using the MLC, these simulations have already been validated but not shown here for conciseness.

Acknowledgements

This work was supported by a grant from the Belgian ‘Fonds National pour la Recherche Scientifique’ (F.N.R.S., grant number FC 73512)

References

[1] Sultanem K, Shu HK, Xia P, Akazawa C, Quivey J M, Verhey L J and Fu K K 2000 Three-dimensional intensity-modulated radiotherapy in the treatment of nasopharyngeal carcinoma: the University of California-San Francisco experience Int. J. Radiat. Oncol. Biol. Phys. 48 711-22
[2] Hsiung C-Y, T H-M, Huang H-Y, Lee C-H, Huang E-Y and Hsu H-C 2006 Parotid-sparing intensity-modulated radiotherapy for nasopharyngeal carcinoma: preserved parotid function after IMRT on quantitative salivary scintigraphy and comparison with historical data after conventional radiotherapy Int. J. Radiat. Oncol. Biol. Phys. 66 451-61
[3] Luxton G, Hancock S K and Boyer A L 2004 Dosimetry and radiobiologic model comparison of IMRT and 3D conformal radiotherapy in treatment of carcinoma of the prostate Int. J. Radiat. Oncol. Biol. Phys. 59 267-84
[4] Mackie T R, Olivera G H, Reckwerdt P J and Shepard D M 2000 Convolution/superposition photon dose algorithm General Practice of Radiation Oncology Physics in the 21st Century, ed A Shiu and D Mellenberg (College Park, MD: American association of Physicists in Medicine) 39-56
[5] Mackie T R, Reckwerdt P J, Olivera G H, Shepard D and Zachman J, 2001a The convolution algorithm in IMRT In 3-D Conformal and Intensity Modulated Radiation Therapy, ed J Purdy, W III Grant, J Palta, B Butler and C Perez (Madison, WI: Advanced Medical Publishing) 179-90
[6] Lu W, Olivera G H, Chen ML, Rechwerdt P J and Mackie T R, 2005 Accurate convolution/superposition for multi-resolution dose calculation using cumulative tabulated kernels Phys. Med. Biol. 50 655-80
[7] Carrasco P, Jornet N, Duch M A, Weber L, Ginjaume M, Eudaldo T, Jurado D, Ruiz A and Ribas M, 2004 Comparison of dose calculation algorithms in phantoms with lung equivalent heterogeneities under conditions of lateral electronic disequilibrium Med. Phys. 31 2899-911
[8] Jones A O and Das I J, 2005 Comparison of inhomogeneity correction algorithms in small photon fields Med. Phys. 32 766-76
[9] Arnfield M R, Siantar C H, Siebers J, Garmon P, Cox L and Mohan R, 2000 The impact of electron transport on the accuracy of computed dose Med. Phys. 27 1266-74
[10] Martens C, Reynaert N, De Wagter C, Nilsson P, Coghe M, Palmans H, Thierens H and De Neve W, 2002 Underdosage of the upper-airway mucosa for small fields as used in intensity-modulated radiation therapy: a comparison between radiochromic film measurements, Monte Carlo simulations, and collapsed cone convolutions Med. Phys. 29 1528-35
[11] Ahnesjö A, 1989 Collapsed cone convolution of radiant energy for photon dose calculation in heterogeneous media Med. Phys. 16 577-92
[12] Aspradakis M M, Morrison R H, Richmond N D and Steele A, 2003 Experimental verification of convolution/superposition photon dose calculations for radiotherapy treatment planning Phys. Med. Biol. 48 2873-93
[13] Verhaegen F and Seuntjens J, 2003 Monte Carlo modeling of external radiotherapy photon beams Phys. Med. Biol. 48 R107-64
[14] Salvat F, Fernandez-Varea J M, and Sempau J 2006 PENEOLOPE-2006: A Code System for Monte Carlo Simulation of Electron and Photon Transport OECD Nuclear Energy Agency, Issy-les-Moulineaux, France
[15] Sempau J, Fernandez-Varea J M, Acosta E and Salvat F 2003 Experimental benchmarks of the Monte Carlo code PENEOLOPE Nucl. Instrum. Methods Phys. B 207, 107–23
[16] Sempau J, Sanchez-Reyes A, Salvat F, Ould ben Tahar H, Jiang S B and Fernandez-Varea J M, 2001 Monte Carlo simulation of electron beams from an accelerator head using PENEOLOPE Phys. Med. Biol. 46 1163-86
[17] Ye S J, Brezovich I A, Pareek P and Naqvi N A, 2004 Benchmark of PENEOLOPE codes for low-energy photon transport: dose comparisons with MCNP4 and EGS4 Phys. Med. Biol. 49 387-97
[18] Jeraj R, Mackie T R, Balog J, Olivera G, Pearson D, Kapatoes J, Ruchala K and Reckwerdt P, 2004 Radiation characteristics of helical tomotherapy Med. Phys. 31 396-404
[19] Abdel-Rahman W, Seuntjens J P, Verhaegen F, Deblois F and Podgorsak E B, 2005 Validation of Monte Carlo calculated surface doses for megavoltage photon beams Med. Phys. 32 (1) 286-98
[20] Zhu X R, Jursinic P A, Grimm D F, Lopez F, Rownd J J and Gillin M T, 2002 Evaluation of Kodak EDR-2 film for dose verification of intensity modulated radiation therapy delivered by a static multileaf collimator Med. Phys. 29 (8) 1687-92
[21] Olch A J, 2002 Dosimetric performance of an enhanced dose range radiographic film for intensity-modulated radiation therapy quality assurance Med. Phys. 29 (9) 2159-68
[22] Dogan N, Leybovich L B and Sethi A 2002 Comparative evaluation of Kodak EDR2 and XV2 films for verification of intensity modulated radiation therapy Phys Med Biol 47 4121-30
[23] Sheikh-Bagheri and Rogers D W O 2002 Monte Carlo calculation of nine megavoltage photon beam spectra using the BEAM code Med. Phys. 29 391-402