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Verification of Dose Calculation Algorithms in Treatment Planning Systems for External Radiation Therapy: A Monte Carlo Approach

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2006

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1

A virtual linear accelerator for verification of treatment planning systems

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Abstract. A virtual linear accelerator is implemented into a commercial pencil-beam-based treatment planning system (TPS) with the purpose of investigating the possibility of verifying the system using a Monte Carlo method. The characterization set for the TPS includes depth doses, profiles and output factors, which is generated by Monte Carlo simulations. The advantage of this method over conventional measurements is that variations in accelerator output are eliminated and more complicated geometries can be used to study the performance of a TPS. The difference between Monte Carlo simulated and TPS calculated profiles and depth doses in the characterization geometry is less than ±2% except for the build-up region. This is of the same order as previously reported results based on measurements. In an inhomogeneous, mediastinum-like case, the deviations between TPS and simulations are small in the unit-density regions. In low-density regions, the TPS overestimates the dose, and the overestimation increases with increasing energy from 3.5% for 6 MV to 9.5% for 18 MV. This result points out the widely known fact that the pencil beam concept does not handle changes in lateral electron transport, nor changes in scatter due to lateral inhomogeneities. It is concluded that verification of a pencil-beam-based TPS with a Monte Carlo based virtual accelerator is possible, which facilitates the verification procedure.

1. Introduction

An accurate determination of the absorbed dose in radiation therapy is very important. According to ICRU 50 (ICRU 1993) the deviation in the dose delivered should be kept within −5% and +7% of the prescribed dose in the planning target volume (PTV). Deviation limits as low as ±3% can also be found in the literature (Brahme *et al* 1988). To ensure this, all involved steps must be minimized regarding uncertainties. This work emphasizes dose calculations using treatment planning systems (TPS). ICRU 42 (ICRU 1987) states that computer-produced dose distributions can be considered accurate enough if they differ from relative dose measurements by less than 2%, or 0.2 cm in position of isodose curves in very steep dose gradients, e.g. in the penumbra and at interfaces between different densities.

The accuracy of treatment planning systems can be determined by performing phantom measurements. Several detector methods have been used, e.g. thermo-luminescence dosimeters (TLD) (Knöös *et al* 1986), film (van Bree *et al* 1994), diodes or ionization chambers (Hurkmans *et al* 1995, 1996). The selection of method, determined by the purpose of the study, also limits the number of dose points, the shape and the composition of the phantoms. The type of detector used can also introduce dosimetric problems, such as violation of the Bragg–Gray relation. Collecting the input data, i.e. the characterization set for a TPS, as well as those measurements needed for verification, is usually rather difficult and time-consuming,

consequently these measurements are not always performed together. The requirements on the accelerators, i.e. stability of output, energy, flatness and symmetry, are of the same order as those put on the TPS, thus other conditions may be present when the verification measurements are performed.

The aim of this work is to study whether measurements for the verification procedure of a pencil-beam-based TPS can be replaced with simulations using a Monte Carlo technique. This could facilitate the commissioning process of a new TPS, and also simplify the determination of limits in the TPS. A common approach is to adjust the Monte Carlo simulated model of a linear accelerator, in clinical use, to reproduce measurements from a real accelerator (e.g. Udale-Smith 1992, Rogers *et al* 1995, Holmes *et al* 1997, van der Zee and Welleweerd 1999). This can, however, be rather cumbersome. This new approach is based on a virtual Monte Carlo simulated treatment unit, which eliminates the previously mentioned drawbacks associated with measurements and drastically reduces the person-hours needed. With this method it will be possible to evaluate the TPS in geometries and points in which it is impossible to do measurements, and the whole volume of interest can be easily studied. Since the variations of the accelerator with time are eliminated there are no problems with adding new geometries for evaluations of the TPS. The data needed to characterize a treatment unit in the TPS are generated using Monte Carlo simulations, a virtual accelerator. These are then implemented into the TPS by the vendor, according to the same procedure as used for 'real' accelerators, and made available for treatment planning. The virtual accelerator and its corresponding unit in the TPS can then be used as a self-consistent data set for studies of dose distribution variations between Monte Carlo data and the TPS. The importance of self-consistent data sets is discussed by the AAPM Task Group 53 (Fraass *et al* 1998). In this study, the emphasis lies on the dose modelling, thus the method is chosen so as to eliminate head-scatter variations. A classical example with an inhomogeneous geometry in which it is usually rather complicated to perform measurements is used to illustrate the advantage of this approach.

Monte Carlo methods are today widely used in different medical radiation applications, since the complexity of photon and electron transport makes it nearly impossible to solve some problems analytically (Andreo 1991). The point-spread functions, which are the basic data in many modern radiotherapy dose calculation algorithms using convolution mathematics, are also in fact generated using Monte Carlo techniques (e.g. Mackie *et al* 1988).

2. Materials and methods

2.1. The calculation algorithm

The treatment planning system used in this study, HELAX-TMS version 4.1A†, is based on fluence modelling and a pencil-beam convolution concept introduced by Ahnesjö *et al* (1992). In short, a pencil beam describing the dose distribution around an infinitesimal ray in water is convolved with the impinging energy fluence, which is shaped and/or modulated by the treatment head. To accomplish this, a number of measurements of the beam have to be performed (characterization). For the pencil beam, a set of depth doses is required from which a laterally invariant depth-dose effective spectrum is derived (Ahnesjö and Andreo 1989). The spectrum is then used to assemble a polyenergetic pencil beam from linear combinations of Monte Carlo based monoenergetic pencil beams.

The total energy fluence is determined by deconvolving a two-dimensional dose distribution, obtained for the largest possible field at an average clinical depth, with the polyenergetic pencil beam, which gives the total energy fluence (Ahnesjö and Trepp 1991). The

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† Treatment Management System, TMS version 4.1A from Helax AB, Uppsala, Sweden.

Table 1. The characterization data set used together with the machine geometry to implement a treatment unit into the treatment planning system.

	SPD (cm)	Field size (cm)	Depth (cm)
Central depth dose	90	5, 10, 15, 20	—
Profiles	90	5, 10, 15, 20	1.5, 5.0, 10.0, 20.0
Star shaped profiles†	90	40	10
Output factors in air	100‡	5, 10, 15, 20	—
Output factors in water	90	5, 10, 15, 20	10

† Profiles at every 10° and the diagonals through the corners of the field.
‡ Source–detector distance.

primary energy fluence is obtained by subtracting calculated head-scatter energy fluence from the total energy fluence. The head-scatter energy fluence is modelled by fitting an analytical expression to measured output factors in air.

In summary, the characterizing data consist of depth doses for the pencil-beam compilation, star profiles for the primary fluence matrix, output factors (OF) in air for head-scatter modelling, absolute dose for monitor unit calculations and profiles in water for penumbra and filter determination. These data are listed in table 1. To be able to implement a treatment unit into the TPS, the machine geometry together with a measured characterizing data set for each beam quality has to be supplied to the vendor. Wedged beams are considered as a separate beam quality. Earlier versions of this TPS have been extensively described and verified in several papers (Knöös *et al* 1994, 1995, Hurkmans *et al* 1995, 1996, Weber *et al* 1996, van't Veld 1997, Basran *et al* 1998, Hansson *et al* 1999).

2.2. Monte Carlo calculations

The Monte Carlo calculations are performed with EGS4, electron gamma shower version 4 (Nelson *et al* 1985). The most suitable EGS4 module for this project is XYZP with PRESTA implemented (Bielajew and Rogers 1987), in which the possibility of using a photon spectrum and divergent beam has been added.

In principle, it is possible to simulate a photon spectrum by allowing electrons to impinge on a target and produce bremsstrahlung. This is, however, time-consuming, and outside the scope of this work. For the purpose of this study, any spectra could be used. However, to get beam qualities in the same range as those used clinically, depth-dose equivalent spectra extracted from the TPS are used. Throughout the study, the EGS4 transport parameters AP = Pcut = 10 keV and AE = 521 keV are used, i.e. photons and electrons with energy above 10 keV are set in motion and transported, and photons with energy lower than 10 keV are considered locally absorbed. Two different values of Ecut are used, 600 and 700 keV (which includes the electron rest mass). Thus, electrons with kinetic energy lower than 89 and 189 keV respectively are considered locally absorbed. The simulations are performed with 10 batches and the statistical uncertainty, estimated as one standard deviation, is kept below 0.8% inside the primary radiation field.

The TPS uses the correct linear attenuation coefficient of the tissue in question during the ray-trace procedure, but calculates the dose to water, which is a consequence of the point-spread functions being generated in water. EGS4, on the other hand, calculates the dose to the medium specified. The difference between μ_{en}/ρ for muscle and water is 1–2% for the energies of interest. Therefore, to achieve consistency, water, with density according to the actual medium in the TPS, is used in the simulations.

Table 2. The voxel size, Ecut and number of photon histories used for the characterizing data simulations for 6 and 18 MV.

	X voxel (cm)	Y voxel (cm)	Z voxel (cm)	Ecut (keV)	Histories (millions)
Depth doses	0.75/4/5/6†	0.75/4/5/6†	0.25/0.5‡	600	35–150
Profiles	0.5	0.50/4/5/6†	0.5	700	200–250
Star	1	1	2	700	440–400

† $5 \times 5/10 \times 10/15 \times 15/20 \times 20$.
‡ 0.25 for depths from 0 to 3.25 cm.

2.2.1. Beam characterizing data for the virtual accelerator. The virtual accelerator is set up to produce two photon energies, 6 and 18 MV x-rays, covering most conventional beam qualities. All the characterizing data for each beam quality (cf table 1) are generated using the modified XYZP in a water phantom with dimensions $54 \times 54 \times 44$ cm^3 (characterization geometry) at a source–phantom distance (SPD) of 90 cm. A layer of air in front of the water is added to yield a more accurate simulation where the generation of electrons in the air column between the bremsstrahlung target and the phantom is accounted for. All photons are randomly selected from a distribution describing the spectra and they are emitted from a point source into the rectangle defined by the beam opening. The emission probability is kept constant over the whole beam area. Selecting the x and y coordinates in this plane from a rectangular distribution fulfills this and consequently no flattening filter is required to homogenize the fluence.

For each field in the characterization set, the dose could be scored in an equally spaced 3D voxel grid covering the whole phantom volume. The dose along the central axis, profiles etc can then be extracted. However, to reduce CPU time and disk space required the Monte Carlo runs are divided into specific depth dose, profile and star simulations comparable to data collection based on measurements. In this way, the voxel geometry can be optimized for each simulation. For example, the lateral voxel extension for depth doses is not critical and can be much larger than for profiles. This makes it possible to keep the uncertainty low and the CPU time reasonable. Voxel size, Ecut and number of histories are summarized in table 2. The output factors in water, OF$_{water}$, are obtained directly from the depth dose and star profiles since all doses scored in EGS4 are normalized to the incoming fluence. The absolute dose calibration (dose per monitor unit) is obtained from the depth dose simulation. There is no need to simulate output factors in air, OF$_{air}$, since the virtual accelerator does not use a flattening filter or collimator jaws, thus no variation in head scatter is present for various field sizes. The only variation in the Monte Carlo simulations may be contributions from 'phantom-scatter' in the air column since different air volumes being radiated distinguish the output for different field sizes. This effect is negligible and therefore all OF$_{air}$ are set to unity.

The implementation of data into the treatment planning system for this virtual linear accelerator is tested by comparing depth doses, profiles and OF for square fields with field sides ranging from 5 to 40 cm from the virtual Monte Carlo based accelerator, with the result from the TPS for exactly the same characterization geometry. All data are normalized to a reference situation (isocentric set-up with 10 cm depth for a 10×10 cm^2 field). Using this approach, both the relative distribution and the absolute level can be evaluated with the same data set.

2.2.2. Application of the virtual accelerator in an inhomogeneous case. The virtual accelerator based on Monte Carlo and its corresponding unit in the TPS is used to calculate the dose distribution for a mediastinum-like geometry with water and lung (water with the relative

density 0.3) (Rice *et al* 1988, Ahnesjö 1989, Knöös *et al* 1995, Blomquist and Karlsson 1998). A 10×10 cm^2 open photon field with an SPD of 100 cm for 6 and 18 MV x-rays is used. The number of photon histories followed is 600×10^6 for both energies, and Ecut is equal to 600 keV. The dose is scored along the x-axis in a Cartesian voxel system with $\Delta x = 0.25$, $\Delta y = 4.00$ and $\Delta z = 0.25$ cm. Isodose lines in the central plane from the TPS and the Monte Carlo simulation are used for comparison.

3. Results and discussion

3.1. Implementation

The TPS vendor has implemented the virtual accelerator according to the same procedure as used for 'real' accelerators. The virtual accelerator is extreme in the sense that the target is infinitely small and head scatter is absent. On the whole, the implementation process turns out successfully. Two small disagreements associated with target size and head scatter are, however, found which are eliminated by minor adjustments of parameters in the TPS database.

Output factor normalized depth doses and profiles for 6 and 18 MV calculated by the TPS are in good agreement with the simulations for the characterization geometry. Deviations are, however, present in the build-up region. Depth doses and profiles for 10×10 and 20×20 cm^2 fields are seen in figure 1. The difference between the TPS calculated and Monte Carlo simulated depth doses is included. The deviations for the depth doses and profiles, for both energies, are less than 2% and for many points less than 1% except in the build-up region where the deviation is larger. This is of the same order as previously reported results based on measurements (Knöös *et al* 1994). The larger deviation in the build-up region is due to insufficiencies in the TPS when modelling charged particle contamination, which has been previously reported (Knöös *et al* 1994). Consequently, in the characterization geometry, the TPS fulfils the requirements for a computer-produced dose distribution according to ICRU 42 (ICRU 1987), i.e. within 2% or 2 mm.

Comparison of OF$_{water}$ for both beam qualities shows a deviation of up to 2% (figure 2). This is actually more than expected since a laterally invariant spectrum is used in the simulations as well as in the treatment planning system. For a physical accelerator this deviation could be several per cent due to the off-axis softening, which is not modelled by the TPS (Knöös *et al* 1994, Hurkmans *et al* 1995).

3.2. Inhomogeneities

Results for the mediastinum case, which is an example of a complicated geometry with regard to measurement techniques, are presented for the TPS calculation and the Monte Carlo simulation in figure 3. A 10×10 cm^2 field and an SPD of 100 cm are used. A penumbra widening and decrease of the dose in the lung is observed in the simulated data due to the prolonged range of the electrons in low-density regions, which causes gradual loss of lateral electron equilibrium. This agrees with results in the literature (Kornelsen and Young 1982, Mackie *et al* 1985, Knöös *et al* 1995). It is notable already for 6 MV (panel (*a*) and (*b*)) that a small penumbra widening is present, which the TPS neglects to model. This mis-modelling is much larger for the higher energy, 18 MV (panels (*c*) and (*d*)). The overestimation made by the treatment planning system is an effect of not scaling the lateral transport of electrons. It always uses the same lateral transport of energy as for water, i.e. the medium for which the pencil beam was determined initially. Thus, when the medium differs from water, changes in the range of the electrons will not be accounted for. The difference between the TPS and the Monte
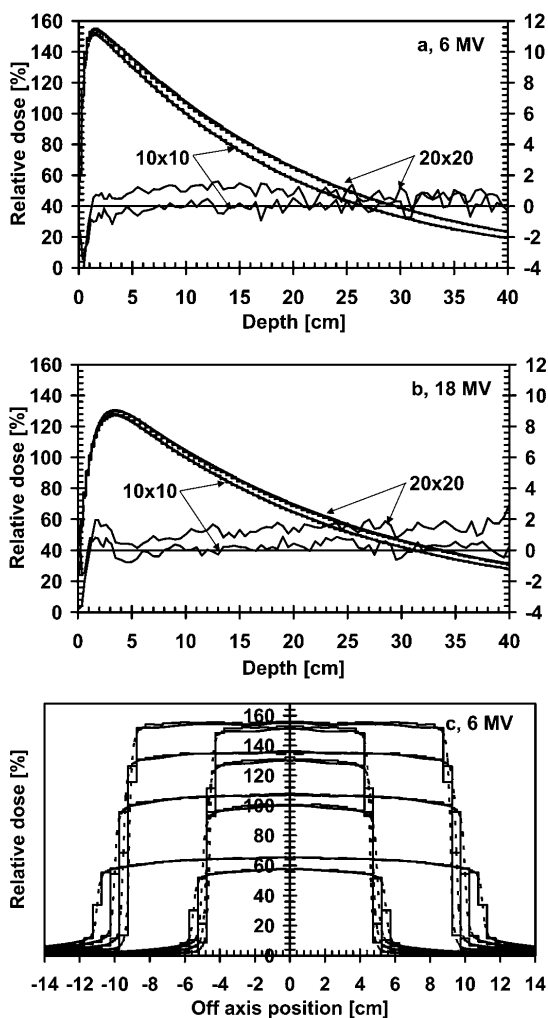
Figure 1. Comparison between depth doses (*a*), (*b*) and profiles (*c*), (*d*) at 1.5, 5.0, 10.0 and 20.0 cm depth, for two fields with equivalent field side 10 and 20 cm respectively, belonging to the characterization set. Step-shaped curves represents Monte Carlo simulations and dashed lines TPS calculations (left scale). The deviations between depth doses are shown in the lower part (right scale).

Carlo simulations in the lung region increases with energy and is about 9.5% for 18 MV compared to 3.5% for 6 MV. Another limitation is that the TPS does not account for changes in scatter due to lateral inhomogeneities. It always uses water as the lateral scatter medium.
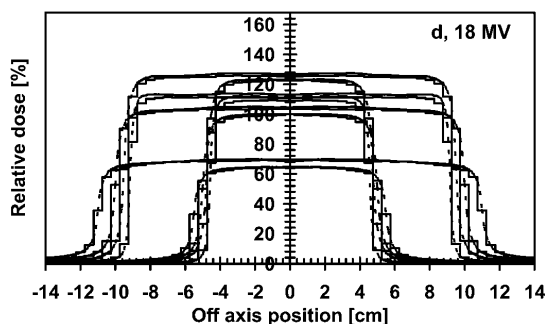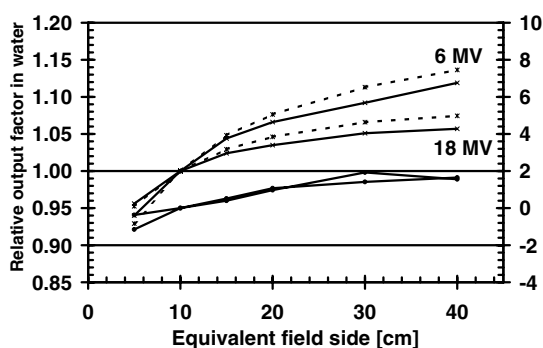
Figure 1. (Continued)

Figure 2. Output factors (OF) in water for 6 and 18 MV (left scale). TPS calculation, dashed curve, Monte Carlo simulation, full curve. The two lower curves represent the deviation between them (right scale).

The relative importance of this approximation decreases with increasing energy due to more forward scattering of the photons.

The developed approach makes it possible to study the absorbed dose in the whole volume of interest in contrast to conventional measurement where the dose is determined in a small number of points (Mackie *et al* 1985, Kosunen *et al* 1993, Hurkmans *et al* 1995); the performance of a treatment planning system in much more complicated geometries can also be studied (Hurkmans *et al* 1995, Kosunen *et al* 1993).

The EGS4 code used for the Monte Carlo simulations does not account for scatter from irradiated parts of the treatment head such as the flattening filter, collimator jaws and auxiliary modulators. The virtual accelerator therefore cannot be used for verification studies focused on head-scatter modelling. The lack of head scatter affects the photon energy fluence, and consequently the virtual accelerator can be used to study phenomena associated with the irradiated media, e.g. phantom scatter, inhomogeneities etc and is therefore suited to isolated studies of the dose modelling part of a treatment planning system.
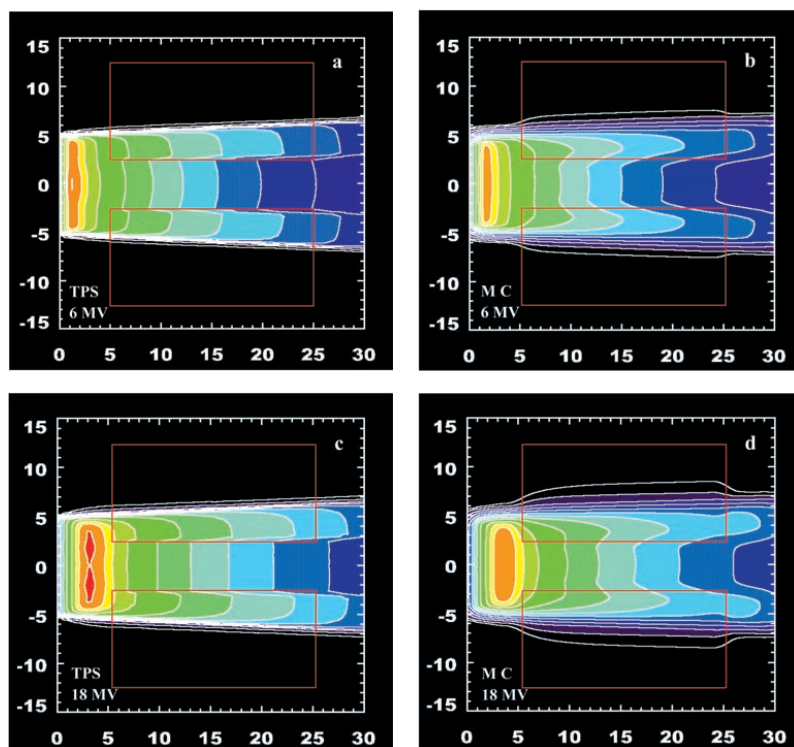
Figure 3. Dose distribution in the central plane in the mediastinum geometry. The two left panels (*a*) and (*c*) show the result from the TPS and the two right panels (*b*) and (*d*) the Monte Carlo result. Both 6 ((*a*) and (*b*)) and 18 MV ((*c*) and (*d*)) are shown. The isodose curves shown are 5, 10, 20, 30, 40, 50, 60, 70, 80, 90, 95, 97.5, 100, 102.5 per cent.

4. Conclusions

The results presented in this study show that it is possible to create a data set for characterization of a treatment planning system with a Monte Carlo based virtual accelerator. Contradictory to conventional measurements, additional verification geometries can easily be added without, for example, any worries concerning performance stability of the accelerator. Thus a self-consistent data set can be maintained through the whole process. The lack of head scatter in the virtual accelerator makes it possible to study the dose modelling process separately. The advantages with this approach are that the problems with fluctuating accelerator performance with time, i.e. dose per monitor unit, energy, symmetry and flatness, are avoided as well as other uncertainties associated with measurements. The person-hours needed for the verification procedure are drastically reduced, and no accelerator time is required. One can also more easily identify and isolate model deficiencies in this way since all uncertainties related to the data collection during characterization as well as verification are avoided. The absorbed dose in

the whole volume of interest can easily be scored for comparison with the TPS. Advanced and complex geometries in which conventional measurements are complicated or even impossible can be studied as well. The mediastinum geometry is an example of a complex geometry in which it is hard to do accurate measurements. The results verify the widely known fact that the pencil-beam concept does not handle changes in lateral electron transport, nor changes in scatter due to lateral inhomogeneities. Converting a three-dimensional computer tomography set suitable for the Monte Carlo code will also make it possible to do comparisons on real patient data. Doing measurements for these situations is impossible at the moment.

For advanced treatment planning systems, i.e. the one used in this study, the modelling of head scatter must be included for a complete verification. This can be done by the BEAM code (Rogers *et al* 1995). For less advanced systems where head scatter is not modelled explicitly or when only phantom scatter integration is studied, the approach used here is sufficient. This study has also shown that it will be possible to implement a full characterization including head scatter into HELAX-TMS.

In the future, data sets for several energies and for different designs of TPS can be made available for both users and vendors for verification of treatment planning systems. This could complement the current available data sets based on measurements such as the Task Group 23 dosimetric verification package developed by the American Association of Physicists in Medicine (AAPM).

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