Accurate model of photon beams as a tool for commissioning and quality assurance of treatment planning calculations

Haydee M. Linares Rosales, Elier Lara Mas, Rodolfo Alfonso Laguardia
Instituto Superior de Tecnologías y Ciencias Aplicadas (InSTEC)
Ave. Salvador Allende y Luaces. La Habana, Cuba
haydeemarialinares2709@gmail.com

Abstract
Simulation of a linear accelerator (linac) head requires determining the parameters that characterize the primary electron beam striking on the target which is a step that plays a vital role in the accuracy of Monte Carlo calculations. In this work, the commissioning of photon beams (6 MV and 15 MV) of an Elekta Precise accelerator, using the Monte Carlo code EGSnrc, was performed. The influence of the primary electron beam characteristics on the absorbed dose distribution for two photon qualities was studied. Using different combinations of mean energy and radial FWHM of the primary electron beam, deposited doses were calculated in a water phantom, for different field sizes. Based on the deposited dose in the phantom, depth dose curves and lateral dose profiles were constructed and compared with experimental values measured in an arrangement similar to the simulation. Taking into account the main differences between calculations and measurements, an acceptability criteria based on confidence limits was implemented. As expected, the lateral dose profiles for small field sizes were strongly influenced by the radial distribution (FWHM). The combinations of energy/FWHM that best reproduced the experimental results were used to generate the phase spaces, in order to obtain a model with the motorized wedge included and to calculate output factors. A good agreement was obtained between simulations and measurements for a wide range of field sizes, being all the results found within the range of tolerance.

Key words: *Monte Carlo method; quality control; photon beams; linear accelerators; tolerance*

Modelo de haces de fotones como una herramienta para el control de calidad de los cálculos en la planificación de tratamientos

Resumen
La simulación del cabezal de un acelerador lineal requiere de la determinación de los parámetros que caracterizan el haz primario de electrones que incide en el blanco de radiación, los cuales juegan un papel importante en la exactitud de los cálculos con Monte Carlo. En este trabajo se realizó la habilitación de los haces de fotones (6 MV y 15 MV) de un acelerador Elekta Precise, empleando el código de Monte Carlo EGSnrc. De forma adicional se estudió la influencia que ejerce cambios en las características del haz primario de electrones sobre la distribución de dosis absorbida en diferentes campos de radiación. Basado en la dosis absorbida, curvas de dosis en profundidad y perfiles de dosis se calcularon y compararon con valores experimentales medidos en un arreglo similar a las simulaciones, empleando criterios de aceptabilidad. Los perfiles de dosis para campos pequeños resultaron ser fuertemente dependientes de la distribución radial (FWHM). Las combinaciones de energías/FWHM que mejor se ajustaron a las mediciones se emplearon en la generación de espacios de fases, para obtener un modelo con la cuña motorizada y para el cálculo de los factores de campo. Se obtuvo muy buena correspondencia entre las mediciones y las simulaciones realizadas, encontrándose todos los resultados dentro de los márgenes de tolerancias.

Palabras clave: *método de Monte Carlo; control de calidad; haces de fotones; aceleradores lineales; tolerancia*

Introduction

Accurate radiation dose distributions in patients are required to plan radiation treatments, to assess the potential for local control of tumors and radiation-induced complications, to develop accurate radiation response data, and to reliably compare treatment plans and techniques [1-3].

The development of fast codes specifically designed for dose calculation in radiotherapy and the considerable increase of processors speed, have enabled the establishment of three-dimensional 3-D Monte Carlo methods (MC) for routine clinical treatment planning [4,5].

MC dose calculation systems for radiotherapy treatment planning of electron beams may soon be commonplace in the clinic. The demand for a reliable calculation of the absorbed dose distribution with a therapy planning system is dependent on the accuracy of modeling the simulated system. MC calculations of dose in a patient requires a good estimation of the distributions of charge, energy, position, and direction the phase-space data of particles emerging from the linac treatment head for its use on the incident source [4,6].

Uncertainties associated with the above characteristics will directly result in a systematic error in dose calculation [4]. Limited information exists in the literature regarding the determination of the initial electron beam characteristics. It has been shown that small changes in the geometry and its composition alter the simulation results [1]. The commissioning process should then be able to deal with this problem and still provide a description of the radiation source that leads to an accurate linac reproduction of those fields according to the user interests. The accuracy of a MC simulation relies heavily on an appropriate selection of the parameters that define the radiation source [1].

For a photon beam produced by a medical linac, the primary radiation source is the electron beam that impinges on the target. Its actual shape and spectra are rarely known and except for very specific measurements conducted, it is impossible to obtain this information experimentally. The direct use of the full phase-space data from treatment head simulations typically stored in phase-space files has the potential to provide the most accurate beam characterization [6].

The main goal of this paper is to obtain and validate a precise MC model of a clinical linear accelerator, which eventually will be used for validation of complex dose calculations performed by commercially available treatment planning systems, when measurements are not feasible or very difficult to set up. Another important application of the model could be the independent verification of intensity modulated plans, in order to reduce the workload associated to patient specific quality assurance base on measurements.

Materials and Methods

In order to commission the MC modeling of the 6 MV and 15 MV photon beams of an Elekta Precise accelerator, a methodology was used in which the energy and spot size from the primary electron beam are varied. To determine the best combination of energy–spot size, three representative field sizes were selected (2 x 2, 10 x 10, 40 x 40 cm^2) that cover the relevant clinic range. Depth dose curves and lateral profiles were experimentally measured and calculated with MC, in order to evaluate their differences as part of the optimization procedure.

Monte Carlo simulations

MC calculations were performed using BEAMnrc and DOSXYZnrc tools from EGSnrc code (v4-2.4.0, 2013). The first of these tools is used to simulate the head of the accelerator, whose components are shown in figure 1. Initially the geometric model of an Elekta Precise linac generic head type was built. All details related with the geometry and composition of each component was obtained from technical specifications of the equipment. The primary electron beam was modeled as a monoenergetic beam with normal incidence in the target using the values of average energy and spot size supplied by the manufacturer. DOSXYZnrc was used to generate voxelized geometries in an array of Cartesian coordinates.

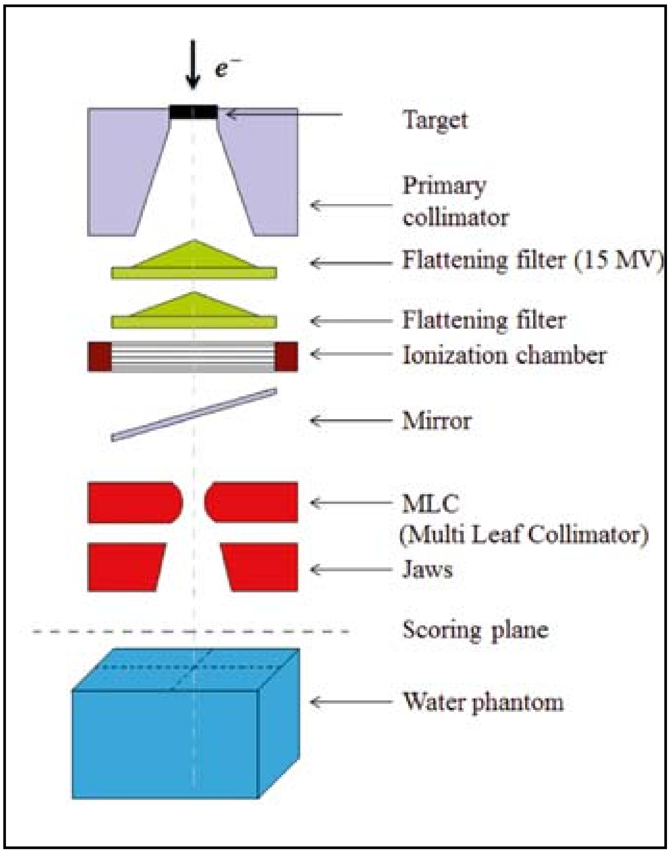

Figure 1. Elekta Precise accelerator scheme, used for MC simulations.

The motorized physical wedge used in this kind of linac was also included in the MC simulation, as they are frequently used in the clinic. Its composition and detailed geometry were provided by the manufacturer.

In these simulations, electrons at the nominal accelerating potential are initiated immediately upstream of the x-ray-producing target. Bremsstrahlung photons and secondary electrons produced are stochastically transported to a plane just after the monitor chamber, but before any secondary collimators. At this point all particle properties are saved to a phase space file. Phase space files generated in the first stage are read as the inputs for calculations that transport particles through jaws defining various field sizes at isocenter. Particles reaching isocenter are saved to files for each modeled field size. These particles are then transported into a water phantom for dose calculations using the DOSXYZnrc tool.

The phase space characteristics and their influence on dose deposited in the phantom, depend on the initial beam characteristic that impinges in the target. Sometimes the beam mean energy differs by at least 0.5 MV from the energy provided by the manufacturer. In order to allow an accurate MC calculation, the user must match the defined beam energy with the real energy of the electron beam impinging the target.

Methodology for optimal beam parameters selection

In order to determine by MC the relevant parameters of the primary electron beam (mean energy and spot size), that allowed to obtain the most realistic model of the actual beam, a sensitivity study was performed, to establish how the variations in mean energy and spot size affect the depth dose curves and lateral dose profiles. Previous studies [1, 7, 4] show that depth dose curves are practically insensitive to variations in spot sizes, similar to dose profiles are to small energy variations. Changes in PDD and dose profiles are noticeable; a minimum variation of 0.25 MeV and 0.5 mm in mean energy and spot size respectively is needed.

Based on the above, an optimal parameters selection procedure for both beams (6 MV and 15 MV) was performed, consisting of two parts: firstly, decide the optimal main energy for each field size, making comparisons between calculated depth dose curves and the experimental data, fixing the value of spot sizes (1 mm), and secondly, once fixed the optimal main energy, varying the dimensions of the beam spot size, to calculate dose profiles and compare them with measurements. The range of values used for both magnitudes, main energy and FWHM is shown in table 1.

Table 1. Range of values to determine the beam optimal parameters

Parameter	Interval	Step
6 MeV	5.50-6.50	0.25
15 MeV	10.50-11.50	0.25
FWHM (cm)	0.5-2.0	0.5

As the optimal mean energy and spot sizes may vary for the same photon beam, depending on the field size, a selection criteria was established for each field, in order to give different weights to each field size. The criterion for energy optimization was based on the usage of the field [1], assigning 50 % weight to the 10 x 10 cm field and 25 % for the other two fields. In case of spot sizes, the fact that the decrease in field size increases sensitivity to variations [1; 4; 7] was taken into account, giving 50 % weight to the field of 2 x 2 cm^2 and 25 % at two other fields.

The procedure described above was used to generate data for both photon beam energies. Depth dose curves were calculated for depths between 0 and 30 cm at the beam axis. Dose profiles were calculated at different depths: 3, 5, 10, and 20 cm respectively. Measured and simulated depth doses are also corrected for small relative displacements by calculating the position of 80 % and 90 % of the PDD maximum in both curves.

Depth sose

Pilot simulations were performed for 2 x 2, 10 x 10 and 40 x 40 cm^2 field sizes, keeping the FSW at a value of 0.1cm and varying the energy in steps of 0.25 MeV from 5.0 M eV to 6.5 MeV. Depth dose was determined in a cubic water phantom with voxel dimensions 1 x 1 x 0.2 cm^3 (x,y,z dimensions where z is the depth), at the central axis, for field sizes 10 x 10 cm^2 and 40 x 40 cm^2. In case of 2 x 2 cm^2 field size, the voxel dimension was reduced to 0.25 x 0.25 x 0.2 cm^3. The maximum z-boundary was set to 40 cm. To verify the chosen optimum parameter set, depth dose simulations were performed for all field sizes.

Dose profiles

For field sizes of 2 x 2 cm^2 and 10 x 10 cm^2 a homogeneous phantom was constructed with voxel dimensions of 0.12 cm and 0.25 cm, respectively, in the x, y axes. In the case of 40 x 40 cm^2 field size, in order to obtain a better agreement of simulation/measurement, the dimensions of the phantom were defined in a different way. For this field a square voxel 0.5 cm wide was set in the plateau region of the field and 0.25 cm in the penumbral region.

Pilot simulations were performed for all field sizes with the fixed optimum energy value previously obtained and varying the FSW from 0.5 cm to 2.0 cm, with step of 0.5. Dose profiles were extracted at 3, 5, 10 and 20 cm depths, for each field size.

Dosimetric set and linac

Experimental verification of the study in question tries to find how close to reality the MC algorithms are under various simulation conditions. Like other types of algorithms, experimental verifications are necessary to ensure both safety and accuracy. This work has been reproduced by MC simulations of the conditions of experiments performed for beams 6MV and 15 MV Elekta Precise accelerator of the INOR.

PDD and dose profiles were measured in the PTW MP3 water scanning system using a PTW semiflex chamber type 31013 for larger field sizes and a PTW dosimetry unshielded diode type 60017 for the smaller fields. The dimensions of the water tank are 50 x 50 x 40.8 cm^3. Measurements were performed for 6 and 15 MV photons incident at 100 cm SSD, for field sizes of 2 x 2 cm^2, 10 x 10 cm^2 and 40 x 40 cm^2, defined at the isocenter. Profiles were measured at depths ranging from 1.5 cm for 6 MV and 3.0 cm for 15 MV to 20 cm. The diode was chosen for these measurements because of its superior spatial resolution, which is necessary for accurately measuring small field profiles, especially in the penumbra region.

Overall uncertainties

As mentioned above, accelerator simulation was performed with EGSnrc code. Previous works [1] have demonstrated that the methodology used in this work, brings better reproducibility of MC calculations and measurements, specifically for small fields.

To evaluate the accuracy of the simulation process, the Venselaar criteria was considered. Once you have obtained a physical description of the beam, it is necessary to evaluate the accuracy of the algorithm used in terms of differences in actual clinical settings. It is well known that: a) there are differences between measurements and calculations, b) these differences are dependent on the location within the beam and the patient's geometry, c) simple statements about criteria of acceptability (tolerances) cannot be made, d) a useful way to compare calculations and measurements is to analyze the deviations statistically, e) any general table of tolerances or expectations depends on the state of the art of the dose calculation algorithms and on the kind of situations (i.e. beams, patients) considered [10].

In this work the test data set consists of relative dose values, measured at specified points in a number of representative beam geometries of the treatment machine. All measurements were done in the same date and time, with the same experimental setup, and is highly consistent with the previously simulated beam data.

Some authors [10; 2] have shown that deviations between results of calculations and measurements can be expressed as a percentage of the locally measured dose:

$$\delta = 100 * \frac{D_{calc} - D_{meas}}{D_{meas}}$$

where δ is a percentage magnitude, D_{calc} is the calculated dose at a particular point in the phantom and D_{meas} is the measured dose at the same point in the phantom.

The level of acceptance of the results is the uncertainty associated with the procedure, which are the result of the measurements themselves, constraints (expected) beam pattern as well as the algorithm used for calculating the dose. According to the above-mentioned, a criteria of acceptability was established depending of the dose region similar to Venselaar et al. [2] studies.

Results and discussion

Previous studies [1] have demonstrated the existing marked dependences on the initial beam parameters of all the field sizes selected in the present work. It's well-known that they are typical beam sizes used in radiotherapy treatment.

Depth dose curves

Figure 2 shows the influence of mean electron energy on depth- dose curves for 6 MV and 15 MV in the 2 x 2 cm^2 field size. In both cases the study field sizes selected were 2 x 2, 10 x 10, 40 x 40 cm^2 respectively.

For 6 MV beam, the percentage local dose differences in the region beyond the depth of maximum were less than 1 % for 5.75 MeV in all the field sizes studied. In case of 15 MV beam, the same behavior was observed for field sizes of 2 x 2 and 10 x 10 cm^2 corresponding to a 11.25 MeV energy value, but in the 40 x 40 cm^2 field size the energy that best reproduced the model corresponded to 10.75 MeV. Table 2 shows the maximum local dose differences for all examined energies in the region where electron equilibrium is reached.

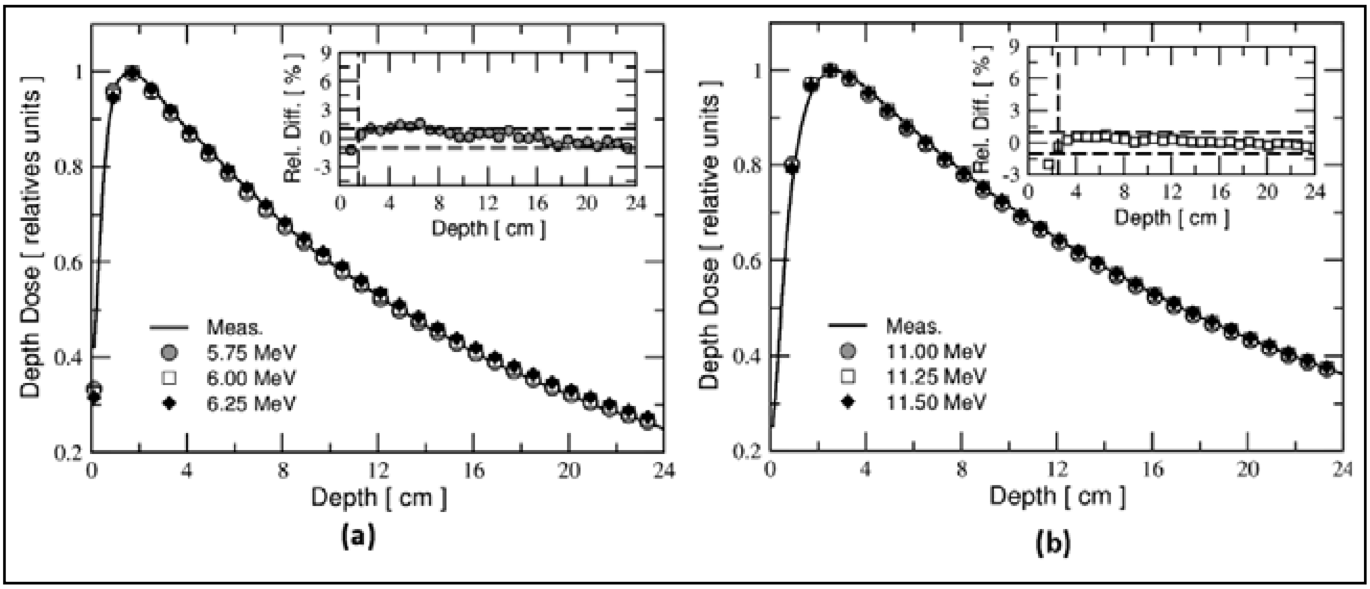

Figure 2. Mean electron energy influence in depth dose curves for (a): 6 MV and (b) 15 MV for the 2 x 2 cm^2 field sizes. Embedded graph shows the relative dose difference respect to (a) 5.75 MeV, (b) 11.25 MeV and the measurements respectively.

Table 2. Percentage local dose differences between measurements and calculated values of depth-dose curves for mean electron energies in the 6 MV and 15 MV photon beams

6MV				15MV			
Energy (MeV)	**2 x 2 cm²**	**10 x 10 cm²**	**40 x 40 cm²**	**Energy (MeV)**	**2 x 2 cm²**	**10 x 10 cm²**	**40 x 40 cm²**
5.5	1.31	1.12	1.21	**10.5**	1.63	1.14	1.59
5.75	0.46	0.27	0.34	**10.75**	1.16	0.44	0.74
6	1.12	0.72	1.44	**11**	0.71	0.26	1.61
6.25	1.26	1.3	1.51	**11.25**	0.23	0.24	2.76
6.5	1.43	1.88	1.73	**11.5**	0.43	0.54	3.57

Lateral dose profiles

The influence of the radial distribution of electron beam on the lateral dose profiles was studied at D_{max}, 5 cm, 10 cm and 20 cm of depth. Some of the results obtained are shown in figure 3 for 6 MV and 15 MV in the 2 x 2 cm^2 field size at D_{max} depth. To obtain these results the energy was fixed in 5.75 and 11.25 MeV respectively. As it has been discussed in several papers [4, 1] the increase in the FWHM of radial intensity resulted in a flattening of the dose profile "horns". The influence of FWHM is more representative in small field sizes, and no graphical differences were observed in the largest field sizes.

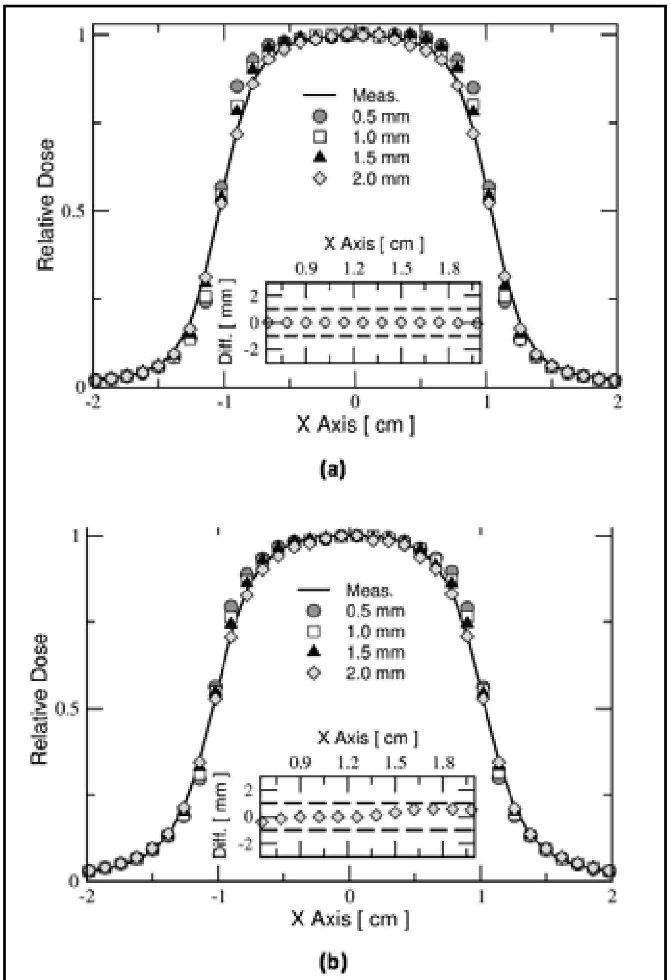

Figure 3. The lateral dose-profile curves as function of the radial intensity distribution given at a FWHM of (a) 6 MeV and (b) 15 MV electron beam at 3 cm depth for field size of 2 x 2 cm^2. Embedded in the graph are shown the discrepancies expressed as distance to agreement for a FWHM of 2.0 mm.

For a FWHM value of 2.0 cm we found the best correspondence between MC calculations and measurements, obtaining differences smaller than 1 mm, as it is shown in the embedded graph, which evaluatesdiscrepancies only in the penumbra region. In 10 x 10 field size, the influence of changing the radius of the electron beam cannot be evaluated separately from the effect of changes in the mean energy, except in regions near the penumbra. Otherwise, 2 x 2 cm^2 is roughly dependent only from changes in the FWHM of the primary beam.

Following the previously described procedure, the optimal combination energy/FWHM was accomplished for both 6 MV and 15 MV beams of the Elekta Precise accelerator head. The best energy/FWHM combination was determined to be 5.75 MeV/2 mm for 6 MV beam and 11.25 MeV/2 mm for 15 MV. The achieved agreement in depth doses and lateral profiles was excellent. Excellent result was also achieved concerning the agreement with measurements in simulations with the motorized wedge model included. The differences in all study cases are quantified through the acceptability criteria based on confidence limits.

Acceptability criteria

Table 3 shows the set tolerances for the accuracy of photons beams dose calculations used in this paper, corresponding to the best combination energy/spot size, applying the criteria proposed by Venselaar et al. [2].

Table 3. Summary of accuracy assessment of MC simulations for open fields at 10 cm depth. Percentages between brackets represent acceptability criteria corresponding to each beam region, (terminology adapted from [2])

	Tolerance	**2 x 2 cm²**	**10 x 10 cm²**	**40 x 40 cm²**
6 MV	δ1 (2 %)	0.94	0.94	0.87
	δ2 (10 %-2mm)	2.51 - 0.36	1.23 - 0.26	1.60 - 0.41
	δ3 (3 %)	1.15	0.50	0.42
	δ4 (30 %)	5.59	13.36	25.67
	RW50 (2 mm)	0.26	0.78	0.81
	δ50-90 (2 mm)	0.41	1.13	1.38
	Tolerance	**2 x 2 cm²**	**10 x 10 cm²**	**40 x 40 cm²**
15 MV	δ1 (2 %)	0.66	0.58	0.95
	δ2 (10 %-2 mm)	3.22 - 0.16	1.60 - 0.28	3.09 - 0.41
	δ3 (3 %)	1.03	0.39	0.38
	δ4 (30 %)	4.95	9.55	19.50
	RW50 (2 mm)	0.26	0.47	0.51
	δ50-90 (2 mm)	0.28	0.92	1.04

In addition, table 4 shows the set tolerances when the motorized wedge is included in the simulation model. All parameters are within the range of tolerance. These tables demonstrate all the previously discussed results. A comparison was made between the deviations that would be allowed in a number of points according to these recommendations. The comparison relative to the same reference dose value was performed at the specified points.

Table 4. Summary of accuracy assessment of MC simulations for wedged fields at 10 cm depth Percentages between brackets represent acceptability criteria corresponding to each beam region (terminology adapted from [2])

	Tolerance	2 x 2 cm²	10 x 10 cm²	30 x 30 cm²
6 MV	δ1 (2 %)	-	-	-
	δ2 (10 %-2 mm)	3.10- 0.70	2.4 - 0.43	5.4 - 0.82
	δ3 (3 %)	1.90	1.20	2.46
	δ4 (30 %)	11.90	11.10	12.31
	RW50 (2 mm)	0.23	0.31	0.534
	δ50-90 (2 mm)	0.721	0.814	0.932
	Tolerance	**2 x 2 cm²**	**10 x 10 cm²**	**30 x 30 cm²**
15 MV	δ1 (2 %)	-	-	-
	δ2 (10 %-2 mm)	3.60 - 0.89	4.50 - 0.53	5.80 - 0.96
	δ3 (3 %)	1.23	1.10	2.35
	δ4 (30 %)	12.34	14.35	15.24
	RW50 (2 mm)	0.29	0.32	0.64
	δ50-90 (2 mm)	0.70	0.62	1.12

Output factors

Table 5 summarizes the results of the output factor from MC calculations and corresponding measurements for fixed monitor units (100 MU) in reference conditions (i.e. SSD = 100.0 cm, depth = 10.0 cm). These values correspond to different field sizes for 6 MV and 15 MV photon beams respectively. The last column of the table shows differences in percentages between simulations with MC and measurements.

In table 5, it is worth observing that the simulated output factors for 6 MV do not deviate more than 1.5 % from the measured output factors. In case of 15 MV, photon beam deviations are less than 1% in all cases, except for 2 x 2 cm^2 field size, which shows a difference of 1.762 % from measurements.

In both cases, the major difference is associated to the smallest field size (2 x 2 cm^2), in which the diode detector tends to overestimate the response. This phenomenon could be explained in the following way: as the material surrounding the diode detector reduces its lateral electronic disequilibrium, its lateral scattering increases, thus resulting in the important role played by secondary electrons in small fields.

Table 5. Output Factors for 6 MV y 15 MV photon beams

	Measurement	MC	Diff. (%)
6 MV			
2 x 2	0.795	0.787	1.074
3 x 3	0.838	0.8437	0.581
5 x 5	0.901	0.902	0.125
7 x 7	0.946	0.951	0.464
10 x 10	1.000	1.000	0.000
15 x 15	1.057	1.063	0.596
20 x 20	1.097	1.100	0.334
25 x 25	1.122	1.129	0.661
30 x 30	1.140	1.146	0.567
40 x 40	1.148	1.165	1.038
15 MV			
	Measurement	**MC**	**Diff. (%)**
2 x 2	0.769	0.783	1.762
3 x 3	0.858	0.866	0.908
5 x 5	0.921	0.925	0.440
7 x 7	0.960	0.962	0.183
10 x 10	1.000	1.000	0.000
15 x 15	1.045	1.053	0.818
20 x 20	1.070	1.080	0.917
25 x 25	1.092	1.102	0.937
30 x 30	1.103	1.111	0.783
40 x 40	1.116	1.120	0.352

Conclusions

To obtain accurate results from MC simulations in radiotherapy calculations, precise modeling of the linac head and a sufficiently large number of particles are required. The simulation of an Elekta Precise accelerator was carried out in this work using the MC code EGSnrc, based on the manufacturer's information.

The geometrical-physic model used in simulations was fitted to reproduce with precision the measurements. In that way depth dose curves, lateral dose profiles and outputs factors were obtained not only for all the above-mentioned field sizes, but for both energy values as well. The influence of the primary electron beam characteristics overdose distribution was studied. In all cases a good agreement was obtained with the experimental data set, being all the results within tolerance.

Once the model of the open beam was validated, the motorized wedge can be simulated in a straightforward way, if a detailed information of it geometry and composition is available.

The linac model has been tested for other geometries with excellent agreement. These results create the foundation to establish a MC-based system for the validation of dosimetric accuracy of calculation engines in commercial treatment planning systems. This is a common task during the commissioning of these TPS, where the lack of adequate measuring devices (detectors, phantoms) that simulate complex clinical conditions is frequent. The availability of an accurate beam model could be used as a substitute for the measurements, thus facilitating the commissioning process.

A methodology for optimizing the modeled beam is proposed and validated, based on the use of confidence limits with tolerances according to beam region and eventually complexity. The accurate beam model developed here has opened the possibility of establishing a MC-based system for patient-specific quality assurance of IMRT plans, consequently reducing the medical physics workload in a busy radiotherapy department as those found in low-income countries.

References

[1] PENA J, GONZÁLEZ CASTAÑO DM & GÓMEZ F. Automatic determination of primary electron beam parameters in Monte Carlo simulation. Med. Phys. 2007; 34(3): 1076-1084.

[2] VENSELAAR J, WELLEWEERDB H & MIJNHEER B. Tolerances for the accuracy of photon beam dose calculations of treatment planning systems. Radiother. Oncol. 2001; 60(2): 191-201.

[3] LIBBY B, SIEBERS J & MOHAN R. Validation of Monte Carlo generated phase-space descriptions of medical linear accelerators. Med. Phys. 1999; 26(8): 1476-1483.

[4] TZEDAKIS A, DAMILAKISJ E, MAZONAKIS M. et. al. Influence of initial electron beam parameters on Monte Carlo calculated absorbed dose distributions for radiotherapy photon beams. Med. Phys. 2004; 31(4): 907-913.

[5] INDRIN J, CURRAN B, CYGLER J, et. al. Report of the AAPM Task Group No. 105: Issues associated with clinical implementation of Monte Carlo-based photon and electron external beam treatment planning.. Med. Phys. 2007; 34(12): 4818-4853.

[6] KAWRAKOW I & WALTERS BRB. Efficient photon beam dose calculations using DOSXYZnrc with BEAMnrc. Med. Phys. 2006; 33(8): 3046-3056.

[7] SHEIKH-BAGHERI D & ROGERS DWO. Sensitivity of megavoltage photon beam Monte Carlo simulations to electron beam and other parameters. Med. Phys. 2002; 29(3): 379-390.

[8] ROGERS DWO, WALTERS B & KAWRAKOW I. BEAMnrcUsers Manual. NRCC Report PIRS-0509(A). 2011.

[9] KAWRAKOW I, ROGERS DWO & WALTERS BRB. Large efficiency improvements in BEAMnrc using directional bremsstrahlung splitting. Med. Phys 2004; 31(10): 2883-2898.

[10] International Atomic Energy Agency. Commissioning and Quality Assurance of Computerized Planning Systems for Radiation Treatment of Cancer. IAEA Technical Report Series 430. Vienna: IAEA, 2003.

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