Application of a Monte Carlo linac model in routine verifications of dose calculations

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Abstract
The analysis of some parameters of interest in radiotherapy Medical Physics based on an experimentally validated Monte Carlo model of an Elekta Precise lineal accelerator was performed for 6 and 15 MV photon beams. The simulations were performed using the EGSnrc code. As reference for simulations, the values of the previously obtained optimal beam parameters (energy and FWHM) were used. Deposited dose calculations in water phantoms were done, on typical complex geometries commonly are used in acceptance and quality control tests, such as irregular and asymmetric fields. Parameters such as MLC scatter, maximum opening or closing position, and the separation between them were analyzed from calculations in water. Similarly simulations were performed on phantoms obtained from CT studies of real patients, making comparisons of the dose distribution calculated with EGSnrc and the dose distribution obtained from the computerized treatment planning systems used in routine clinical plans. All the results showed a great agreement with measurements, finding all of them within tolerance limits. These results allowed the possibility of using the developed model as a robust verification tool for validating calculations in very complex situations, where the accuracy of the available TPS could be questionable.

Key words: *Monte Carlo method; accuracy; linear accelerators; radiation dose distributions; radiotherapy*

Aplicación de un modelo de Monte Carlo de un acelerador lineal en la verificación de los cálculos dosimétricos de rutina

Resumen
El análisis de algunos parámetros de interés en la física médica de la radioterapia, basado en un modelo de Monte Carlo de un acelerador Elekta Precise, fue realizado en este trabajo para los haces de fotones de 6 y 15 MV. Las simulaciones se realizaron con el código EGSnrc. Como referencia para las simulaciones se emplearon los parámetros óptimos (energía y FWHM) previamente calculados. Los cálculos de la dosis absorbida se realizaron con maniquíes de agua sobre geometrías complejas, comúnmente empleadas en las pruebas de aceptación y control de calidad en la clínica. Parámetros de interés como la dispersión en las MLC, la máxima posición de apertura o cierre y la separación entre estas se analizaron a partir de los cálculos en agua. De forma similar se realizaron cálculos en maniquíes construidos a partir de los estudios tomográficos y comparaciones con los resultados reportados por el sistema de planificación en dichos casos. Los resultados obtenidos evidenciaron una gran concordancia con las mediciones, encontrándose dentro de los límites de tolerancias reportados. Estos resultados crean la base para el empleo del modelo de Monte Carlo como una herramienta robusta para verificar y validar los cálculos de dosis en situaciones de gran complejidad, donde la exactitud de los sistemas de planificación es cuestionable.

Palabras clave: *método de Monte Carlo; precisión; aceleradores lineales; distribución de las dosis de radiación; radioterapia*

Introduction

Nowadays one of the most discussed topics is how to obtain an accurate dose distribution calculation in Radiotherapy. Also getting dose values from the treatment planning system (TPS) under the level of acceptance is an every day challenge in the clinical environment. Absorbed dose distribution calculations with a TPS are strongly linked to the accuracy of the simulated system. The application of those calculations requires a good estimation of the system specifications such as energy, charge distribution, direction and position of the particles generated from the head source [1, 2].

On the other hand at present, there is a widespread interest in the clinical implementation of modern radiotherapy technologies, such as RapidArc, VMAT, TomoTherapy and CyberKnife. Given the complexity of these technologies and the sophistication of the dose calculation engines used by their respective commercial treatment planning systems (TPS), Monte Carlo (MC) methods have proved being very useful for patient-specific treatment quality assurance (QA), TPS commissioning, or for clinical site-specific treatment technique commissioning. However, in order to represent as realistically as possible the beam delivery and dose deposition to patients, MC simulations of these technologies require the capability of continuously modeling variable beam configurations and complex treatment geometry and kinematics with respect to the patient [3].

The MC calculation process is not exempt of uncertainties, which leads to a resultant systematic error [2]. The main uncertainties sources are the intrinsic simulation uncertainty and the available dosimetric set used in the validation of the MC model obtained. When a dosimetrically validated MC model is available, it could be used as reference to perform the response evaluation of different detectors that could present troubles such as energetic or angular sensibility response, partial volumes effects, dose rate dependence, etc. In addition, the referred model could be used as tool for verification in the dosimetric calculations performed by TPS, although with limitations in determined clinical conditions such as small fields, boundaries between regions with different densities, etc.

The main goal of this work was the application of a previously obtained Elekta Precise linac MC model for the simulation of complex geometries and patients. Furthermore, we intend to evaluate the reproducibility capacity of such model in conditions where the response of the conventional evaluation methods is questionable.

Materials and Methods

MC linac model

Monte Carlo simulations were performed using the EGSnrc V4-2.4.0 code, from a linac model previously obtained and dosimetrically validated. The main physical parameters used are shown in table 1. Those values correspond to the optimal parameters (mean energy and FWHM) of the 6 MV and 15 MV electron primary beams.

Table 1. MC model optimal beam parameters employed during simulations

Nominal Beam	Mean Energy (MeV)	FWHM (mm)
6 MV	5.75	2.0
15 MV	11.25	2.0
FWHM (cm)	0.5 - 2.0	0.5

BEAMnrc was used in phase space generation with enough statistics below the collimation system. The scheme and the components' modules from BEAMnrc used in simulations are shown in figure 1. Phase space files were used as sources in DOSXYZnrc to calculate depth dose in both, water phantoms and CT phantoms from patients' studies.

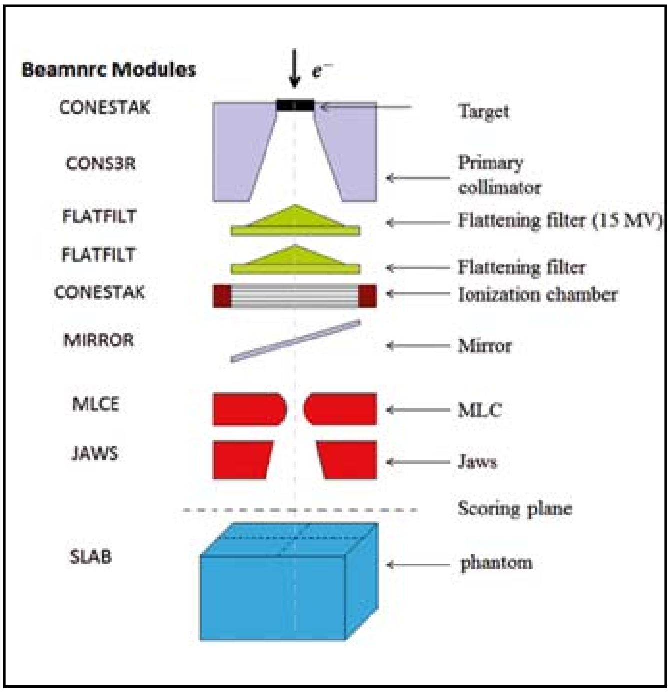

Figure 1. Linac scheme and component module names used in each case in BEAMnrc simulations.

Calculations in water were performed using a phantom with 0:2 x 0:2 x 0:2 cm^3 dimensions (x, y, z where z represents depth in the coordinate system). The maximum z limit was established at 40 cm depth. The physical parameters in simulations were established according to previous publications [1, 4, 5] to ensure the best reproducibility with measurements as well as to get the best compromise between accuracy and simulation speed.

Measurements

The experimental verification was performed using the PTW MP3 water scanning system and the PTW dosimetry unshielded diode type 60017 ("electron diode"). 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 source to surface distance (SSD).

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. Silicon diodes have the sensitive volume small enough (typically < 0.2 mm^3) so that the volume averaging effects can be avoided. However, their angular dependence is not uniform due to the internal construction and materials used and can vary by 3 % in magnitude [6]. Diodes are known to over respond to low energy photons due to the differences in mass energy absorption coefficients of silicon and water at keV energies. However, in small fields, where the scattered radiation is reduced, the contribution of low energy photons is rather low. Care must be taken to select an adequate type of diode. Unshielded diodes ("electron diodes") were reported to have more adequate properties for small field dosimetry than shielded ("photon diodes") [7-10]. Shielded diodes are energy compensated, to absorb some of the low energy scattered photons, and contain high density material (e.g. tungsten) [8]. However, the presence of tungsten increases the fluence of secondary electrons in silicon due to the higher mass energy absorption coefficient of tungsten, for lower energy photon beams. This causes over response of a diode. It was shown that the response of shielded diodes is not completely independent of changes in field size and the depth of measurement [8]. The increase in the contribution of low energy scattered photons with depth results in an over-response of shielded diodes. However, some diodes have been reported to exhibit under response at large depths [11]; which was attributed to the dose rate dependent response. In small fields, like used in IMRT, the use of unshielded diodes is recommended.

For profile measurements, a diode should be oriented parallel to the beam axis and two scans in opposite directions should be made to resolve potential asymmetry due to directional dependence of the diode response. For measurements in very small fields stereotactic diodes should be used. Diodes have a limited lifetime and their sensitivity depends on accumulated dose. Consequently, they should be periodically re-calibrated [12].

Application of the MC linac model

Once the MC model was dosimetrically validated, simulations were performed in complex geometries with clinical interest such as those which could be of interest for acceptance testing and commissioning purposes. To evaluate the application range of the model, irregular and asymmetric fields were constructed.

Figure 2 shows MC screenshots of some of the above-mentioned geometries. Parameters such as MLC scatter, maximum opening or closing position, and the separation between them were analyzed from calculations in water. To perform simulations where the MLC scatter will be evaluated, two special geometries configurations were created. The first is a geometry in which the left and right MLC banks are separated 1cm as shown in figure 3 (a). For convenience, in the future we will refer to this geometry as "GEOM 1". In this case a dose profile along the MLC system was used to evaluate differences between measurements and calculations. The second geometry created consisted in an open square field 20 x 10 cm^2 dimensions, in which a MLC pair was kept closed as figure 3 (c) shows. This geometry is referred as "GEOM 2". In this case a dose profile across the MLC was used to evaluate the differences. In both Figures the green line indicates the position in which the doses' profiles were obtained.

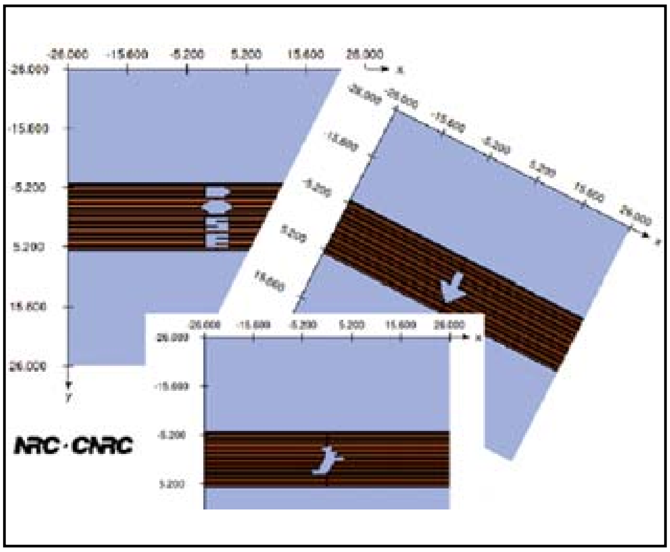

Figure 2. Irregular geometries created using the MC model. Preview of the leaf configuration surface.

Taking into account the equation, deviations between the results of calculations and measurements can be expressed and evaluated as a percentage of the locally measured dose, 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.

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

The level of acceptance of the results is determined by the uncertainty associated with the procedure, which results from the measurements themselves, constraints (expected) beam pattern as well as the algorithm used for calculating the dose. According to that statement, a criteria of acceptability was established at δ = 1 % for all the study cases.

The fact of having enough statistics below the collimation system and precision in the geometry creation, allows performing dose calculations using the source number 20 or 21 in DOSXYZnrc. Source 20 uses a phase space file as source to perform dose calculations in DOSXYZnrc. This source greatly enhances the capabilities of the phase space source incident from multiple directions and allows the user to simulate continuous motion of the phase space source relative to the DOSXYZnrc phantom over specified ranges of incident directions, SSD and isocentre coordinates. Moreover, the source allows the user to interpose a geometry, generated using either a BEAM accelerator or a non-EGSnrc code (likely simulating an MLC geometry) compiled as a shared library, between the source plane and the DOSXYZnrc phantom.

Source 21 defines a beam treatment head simulation (compiled as a shared library) source incident over multiple ranges of continuous motion with respect to angle, SSD and isocentre. The source motion can be synchronized with the settings of any synchronized component modules (CMs) in the accelerator. There is also an option to run the source through geometry (usually MLC) defined, compiled as a shared library, placed between the treatment head and the DOSXYZnrc phantom [13]. Having the ability with source 20 or 21 of combining couch movements, collimator and gantry rotations, multiple dynamic and statics beams were simulated, both in water and CT phantoms.

Patient simulations

Calculations in patients were also performed for the most clinically representative cases. Taking into consideration the TPS specifications, the corresponding input files were constructed through the DICOM files information. In each case the CT sets were resampled to 0.4 cm^3 voxels with the average density based on the Hounsfield numbers. From these numbers, the materials of the voxels were also mapped, using the respective calibration ramp. The following materials were used: air, lung, tissue and bone. A routine step-and-shoot IMRT plan was selected for MC simulation of dose deposition on a head and neck patient's CT scan.

Results and Discussion

Simulations with the mentioned linac model were performed, which was previously commissioned and validated against measurements. The range of application of such MC linac model is wide, in which all the routine clinical parameters can be evaluated. In the present work the analysis of some parameters has been performed, but also calculations in patients were performed, which is one of the main challenges nowadays.

Figure 3 shows the geometries configurations which were used to evaluate the MLC scatter. These configurations were used in both 6 MV and 15 MV beams. In figure 3 (a) and (c) exhibit an X-Y scatter obtained with BEAMdp from the 6 MV beam linac model; while (b) and (d) are the dose profiles obtained from calculations with DOSXYZnrc (black line) and measurements (grey line) corresponding to (a) and (c) respectively. In both cases the major dose difference observed was in the penumbra region, but the value never exceeds the 1.83 %. In both energies the dose differences are rounding the 0.97 % as an average. Regarding the displacement between points (distance-to-agreement), the maximum difference observed between calculations and MC simulations was 1.4 mm. The MLC leakage analysis showed leakage picks from which the maximum dose observed does not exceed the 4.14 % from the open field 10 x 10 cm^2.

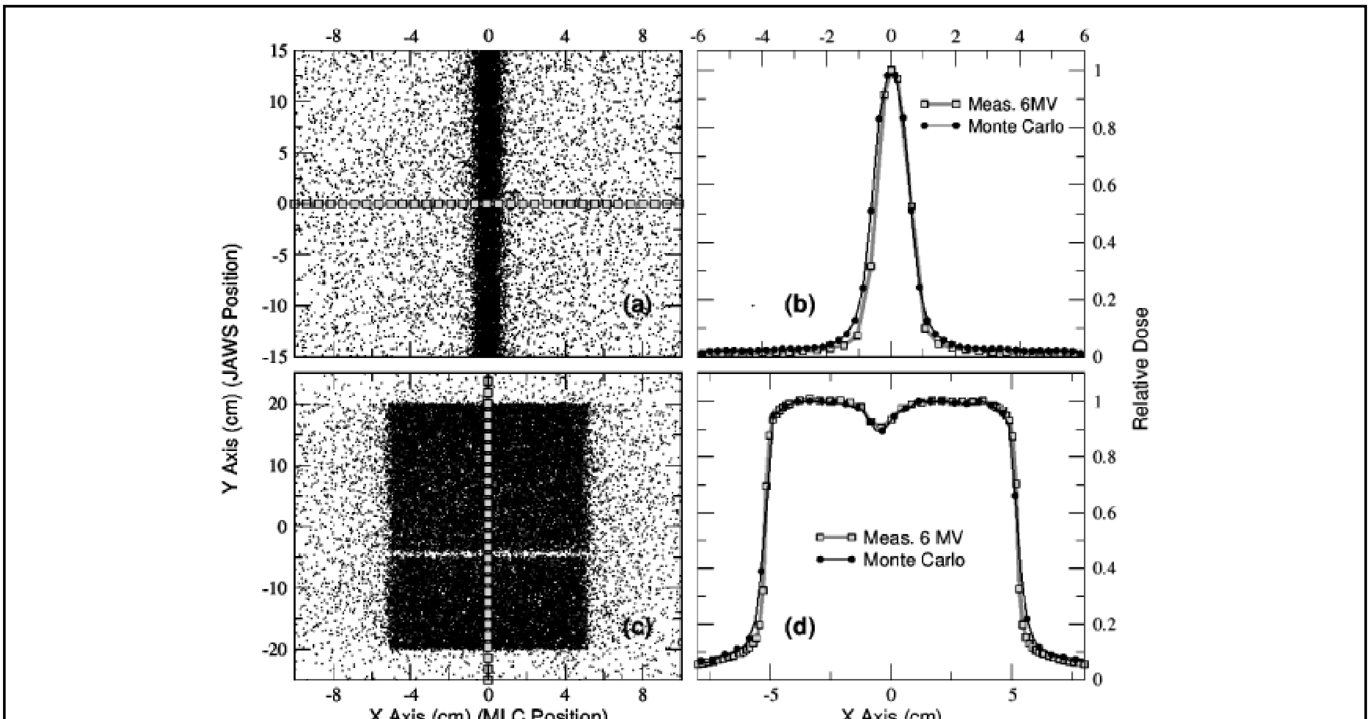

Figure 3. Evaluation of the MLC scatter (a) and (c) exhibit an X-Y scatter obtained with BEAMdp from the 6 MV beam linac model. That geometry had been for convenience named "GEOM 1" and "GEOM 2" respectively (b) and (d) are the dose profiles obtained from calculations with DOSXYZnrc and measurements corresponding to (a) and (c) respectively. The grey lines shows the level at which the dose profile were evaluated at 5 cm depth.

To evaluate the accuracy and precision in the design of geometry with shaped irregular field of high complexity, a geometry named "DOSE" was created for both 6 and 15 MV, which form is a DOSE poster. Figure 4 shows the results obtained from calculations in the "DOSE" geometry. Using the BEAMdp tool from EGSnrc, the figure 4 (a) was obtained, which is an X-Y scatter graph from the phase space file generated above the collimation system for 15 MV beam MC linac model. The green and red lines were intentionally added to the picture to represent the two positions (X = 3 cm and X = 3 cm) where the dose profiles will be evaluated. In figure 4(b) and (c) show the dose profiles obtained from calculation over a water phantom for the geometry showed in (a) at 5 cm depth in the green and red positions respectively. The orientation of the dose profiles in figure 4 b and c was intentionally changed to show how well the model reproduces the shape of each letter.

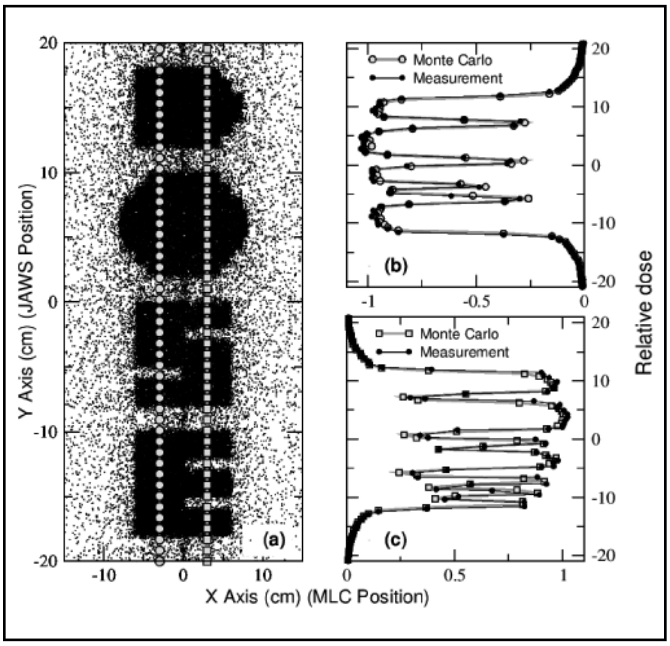

Figure 4. MC linac model simulation in a complex geometry (a) shows an X-Y scatter obtained from a phase space generated by the MC linac model for the "DOSE" irregular fields, (b) shows a dose profile corresponding to grey line position (X = -3 cm) in (a) and (c) shows a dose profile corresponding to the black line position (X = 3 cm) in (a).

Figure 4 evidences the high level of correspondence between calculations and measurements. The model is able to reproduce any geometry configuration, regardless of the level of complexity or the size. Through the equation, the maximum difference observed in this case was 0.85 % - 0.53 mm which is an evidence of the above mentioned geometry reproducibility. The MLC opening-closing position and separation among them were also evaluated. Figure 5 illustrates the isodoses lines from calculations in DOSXYZnrc using the geometry configuration shown in figure 4.

Table 2 summarizes the maximum deviation values obtained in all the simulations performed in water phantoms. The values expressed in percentages are the relative dose differences considering the equation.

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

Field	6 MV Beam	15 MV Beam
5 x 20 cm^2	0.80 % - 0.9 mm	0.61 % - 0.85 mm
20 x 5 cm^2	0.81 % - 0.86 mm	0.62 % - 0.79 mm
10 x 10 cm^2 (offaxis)	0.74 % - 0.65 mm	0.54 % - 0.61 mm
20 x 10 cm^2 (half blocked)	0.78 % - 0.85 mm	0.58 % - 0.78 mm
GEOM 1	1.02 % - 1.4 mm	0.98 % - 0.81 mm
GEOM 2	0.99 % - 0.96 mm	0.94 % - 0.81 mm
DOSE	0.89 % - 0.54 mm	0.85 % - 0.53 mm

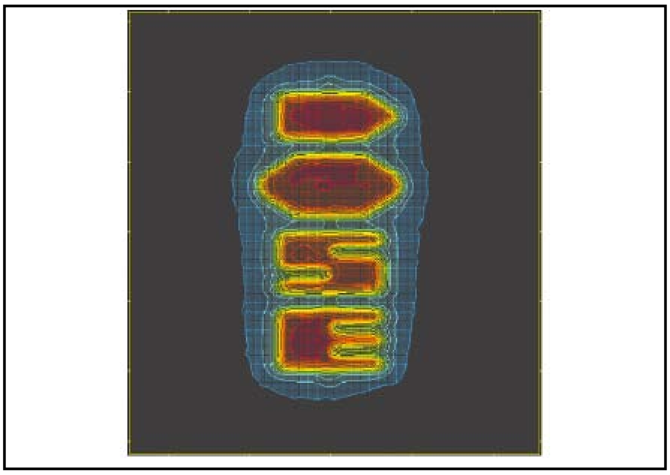

Figure 5. Isodose curves obtained from the application of the MC model over a water phantom.

Combining all the above-mentioned possibilities and extending calculations to CT from TPS, figure 6 presents the benchmarking results obtained, showing the dose distribution for the described head and neck IMRT plan. Figure 6 (a) represents the dose distribution corresponding to the axial and sagital views obtained from the Elekta Precise Plan treatment planning. An screenshot of the DOSXY Z show code from DOSXYZnrc is shown in figure 6 (b), corresponding to the dose distribution obtained from MC calculations using the 6 MV linac beam model.

The IMRT step and shoot case was simulated, using both sources 20 and 21. Using source 20 the simulations took about four to six hours to perform for 1 x 109 histories with an uncertainty of about 0.2 % in the high dose region, using an Intel Core i7 with four processors at 2.0 GHz. Simulations with source 21 took about 36 to 48 hours with the same number of histories, and as result, uncertainty values of about 0.2 % were obtained in the high dose region just as in source 20 simulations.

The main difference between those sources is associated to the simulation time. Source 21 requires a full BEAMnrc simulation as was reported in [3, 4] and as result to perform 1 x 109 histories in DOSXYZnrc, hundreds of thousands of histories must be transported through each component in BEAMnrc.

The high level of correspondence in both, the TPS and the MC dose distribution, as well as the homogenei-

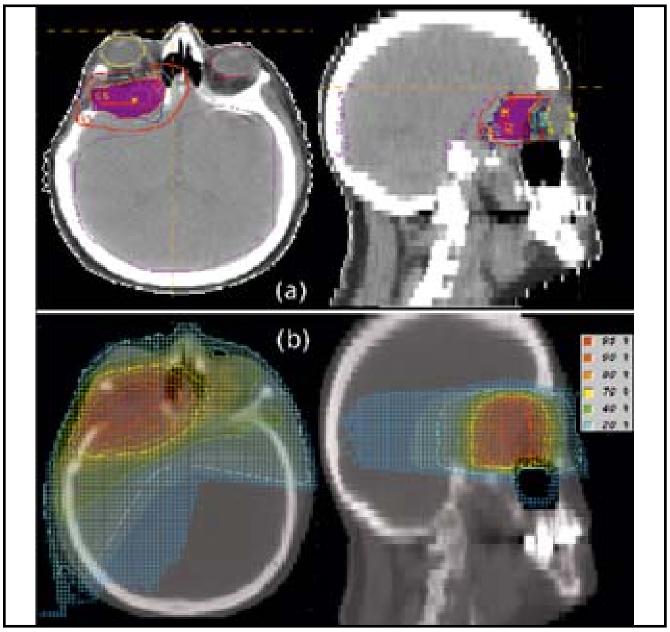

Figure 6. IMRT case dose distribution calculated with: (a) Elekta Precise Plan treatment planning and (b) DOSXYZnrc code.

ty, are aspects that could not be obviated from the case shown. Quantitatively the differences between MC dose distribution calculations and the TPS could be evaluated from the Dose Volume Histogram (DVH) presented in figure 7. This figure shows the DVH information corresponding to each of the structures contoured in the TPS and also the DVH information from MC corresponding to the CTV and PTV structures.

In comparison with the TPS DVH obtained for the PTV, the major difference observed in MC calculations was in the range of 80-90 % of the relative dose, with a maximum difference value of 1.8 % of ratio of total structure volume. In case of the CTV the maximum difference value observed was 1.1 % of ratio of total structure volume, in the region of 87-92 % of the relative dose. The major difference observed in case of the left eye, was in the region of 73-83 % of the relative dose, with a value of 2.1 % of ratio of total structure volume. For visualization purposes, the MC DVH corresponding to the remaining structures were not shown, but in all cases a good agreement was observed.

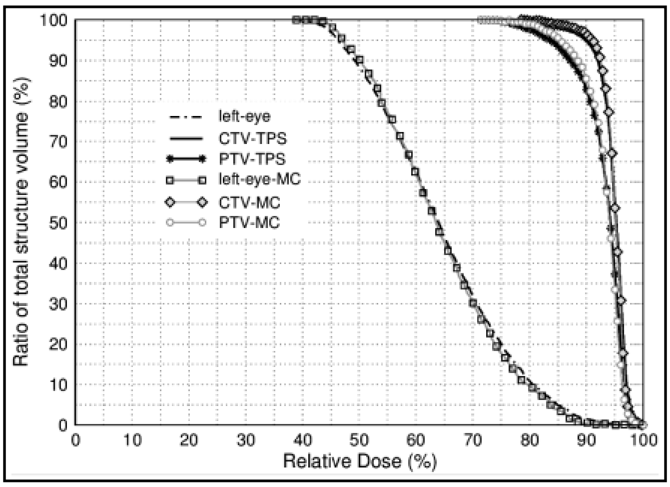

Figure 7. Dose volume histogram obtained for the head and neck IMRT case.

Conclusions

The application of a MC linac model in routine verifications of dose calculations was performed. The geometrical reproducibility test showed differences in values always below 2 % - 2 mm. In all the examined geometries, the MLC leakage values reported by the MC calculations were below 5 %, being in concordance with the reported information concerning that parameter. Dose calculations in water and CT phantoms showed uncertainty values below 0.5 %, and the maximum deviation observed does not exceed 1.5 % - 1.5 mm. The results showed that the model can be used for validation of dose calculations of available TPS, especially in complex clinical scenarios, where experimental verifications are difficult to be implemented and measurements are prone to larger uncertainties.

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