Verification of dose calculations in a heterogeneous phantom using PRIMO

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Abstract: Treatment planning systems (TPSs) are routinely used in radiotherapy to calculate the monitor units (MU) required to deposit a certain amount of energy per unit mass in the patient and the dose distribution in it due to their high calculation speed. To reduce calculation time, TPSs adopt several approximations which are optimized for low atomic number materials such as those found in the human body. These approximations may lead to substantial deviations between calculated and measured dose in the presence of heterogeneities. In the present TFG we employ Monte Carlo simulations to study these discrepancies and verify the TPS calculations. The program used is PRIMO, which is based on the PENELOPE code system.

I. INTRODUCTION

X-ray beams are widely used in radiotherapy to treat the majority of malignant tumours. This ionizing radiation is delivered to the patient using a specific type of particle accelerator called linear accelerator (linac) [1]. In order to ensure the proper deposition of energy, by the interaction of the radiation with the tissues of the patient, several tests are made periodically in the hospitals. One of the aims of these tests is verifying if the dose calculated with Treatment Planning Systems (TPSs) agrees with the dose measured in different types of phantoms which reproduce parts of human body, both geometries and tissues. The proper calculation of dose into the patient in the planning process is extremely important because an undue energy deposition in healthy tissues could compromise tumour local control or cause damage to healthy tissues.

Commonly, there are no substantial discrepancies between dose calculations with TPSs and measured doses in homogeneous media as long as low atomic number elements are involved (up to $Z \approx 12$). However, the result of simulations in media with abrupt changes of mass density, like lung tumours (air/tissue), bone-crossing fields (tissue/bone) and treatments of patients with metallic prosthesis (tissue/Ti), differs in respect of measured values. In view of the relevance of this topic, dose discrepancies must be investigated.

A study done at the *Unitat de Radiofísica de l'Hospital Clínic de Barcelona*, using the TPS *Eclipse* distributed by *Varian*, and the heterogeneous phantom CIRS CBCT 062 [2], which reproduces the geometry of an abdomen with different tissue equivalent plugs, shows these discrepancies. The aim of the present TFG is to compare the dose calculated with *Eclipse* and the measured one with the prediction of the program PRIMO [3], which is based on the Monte Carlo (MC) code PENELOPE [4]. This comparison will be useful to determine if TPSs like *Eclipse* have shortcomings on dose calculations in heterogeneous media and the subsequent investigation of discrepancy sources.

To this end, in the present work the nominal photon beam energy considered is 6 MV owing to the wide use of this energy in radiotherapy treatments. Discrepancies using higher energy beams are less important.

II. MATERIALS AND METHODS

A. Simulation setup

The program PRIMO was installed on an Intel Xenon server (8 cores) with the Windows operating system.

The simulation process in PRIMO is divided into three steps or segments. In the first segment (s1) the production of Bremsstrahlung photons in the W target is simulated. Segment two (s2) simulates the movable collimators conformation. In the third segment (s3) the dose distribution in a phantom or patient is tallied. The segments are simulated sequentially and the results of s1 and s2 (i.e., particles states) are written to the hard disk in files called phase space files (PSFs). The linac geometry is defined choosing a linac model. In the present work a Varian 2100 linac is used.

Before running the simulations, each segment must be configured. Beam parameters which can be modified in s1 are the nominal energy, the initial electron energy, the beam angular divergence and, assuming a Gaussian distribution for the energy and for the radial distribution, the corresponding FWHM of the energy distribution and focal spot size. The field conformation is defined in s2. The aperture of the jaws and leafs of the multi-leaf collimator (MLC) and the isocenter position modify the size and position of the radiation field. Finally, in s3 one can choose either a homogeneous water phantom or a computed tomography (CT) volume, which has to be imported. The size of the patient model and binned grid dimensions can be modified in the water phantom while the sizes of CT volumes are determined by the images imported and cannot be changed. The voxelized geometry

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is created by associating a material and a mass density to each voxel in the CT volume. The density of each bin is assigned through a calibration curve, which relates physical densities and Hounsfield units (HU) delivered by the CT scan, and the volume segmentation is done by assigning each material to a HU interval. In this segment, the source-to-phantom-surface-distance (SSD) is adjusted.

Analogue simulation [5] of all segments may require several months of computing time to achieve low average statistical uncertainties. In order to reduce the amount of CPU processing time, PRIMO uses so-called variance-reduction techniques in all segments. These techniques rely on the idea that it is possible to change the probability distribution function of deposited energy keeping its mean value in each bin unbiased. Thereby, the distribution of energy deposited in each bin is altered reducing the variance maintaining the average value of dose unaltered. A weighting method is used to modify the probability distribution correcting the statistical weight of each particle proportionally to their contribution to the final result. These techniques are used separately in each segment. Two splitting techniques, splitting-roulette [6] and rotational-splitting [7], and Bremsstrahlung interaction forcing in the target are available in s1. A movable-skins technique [8] is used in s2 and PRIMO selects automatically the skin thickness in the jaws and MLCs depending on the beam energy. These techniques are combined with a simple splitting in s3 for the dose tallying. The splitting factor can be chosen too depending on the statistical uncertainty we want to achieve. Increasing the value of this factor does not ensure the reduction of uncertainty insomuch as variance depends on the number of particles in the PSF too, the so-called "latent variance" [9]. Since simulation time is proportional to the splitting factor value, choosing it wisely is extremely important in order to maximize the efficiency ϵ, defined as $\epsilon = 1/\delta^2 t$ where δ is the statistical uncertainty. Combination of variance-reduction techniques in s1 and s3 does improve the efficiency considerably. Since only 6 MV beams were simulated in this work, the variance-reduction technique used in s1 was splitting-roulette as recommended by the authors of PRIMO for nominal energies below 15 MV.

All uncertainties reported by PRIMO are given as 2 standard deviations.

B. Tuning of the initial beam parameters

The accurate characterization of initial beam parameters of each linac is necessary because they may be different, even for the same model of linac, and MC methods need these parameters as input. Although the nominal energy of the x-ray beam can be chosen, the energy of the accelerated electrons, which produce photons by Bremsstrahlung in the W target, depends on the assembly and electronics of the tuned-cavity waveguides used to accelerate the electrons. As a consequence, the energy distribution of the photons produced and delivered to the patient may differ somewhat from the nominal energy selected.

The penetration of the photon beam in the patient depends on the photon energy distribution, and therefore the energy of the electrons that hit the W target must be set correctly. In this work we consider a mono-energetic electron beam as a previous study of the commissioning of the same model of linac proposed [10]. Since the focal spot size and beam divergence affect only the penumbra of the radiation field in the patient and, furthermore, we focus on measurements in the linac isocenter, these two parameters will not be tuned in the present TFG.

Commonly, the percent depth dose (PDD) curve in water is used to characterize the quality and penetration power of the photon beam. $\mathrm{PDD}_{20/10}$ is defined as the ratio of absorbed doses at depths of 20 cm and 10 cm,

$$\mathrm{PDD}_{20/10} = \frac{D(20\ \mathrm{cm})}{D(10\ \mathrm{cm})}, \tag{1}$$

for a 10×10 cm^2 field and a SSD of 100 cm.

To set the correct initial electron energy, it must be varied until the simulated $\mathrm{PDD}_{20/10}$ ratio agrees with the experimental value. Only changes larger than 200 keV induce perceptible changes in the energy deposition. Instead of wasting CPU time making several simulations with different values of the initial energy, owing to the small variation of the PDD curve slope with beam energy changes, we can approximate the relation between dose gradient and beam energy as linear. Then, to determine the energy for which the $\mathrm{PDD}_{20/10}$ value agrees with measurements we want to reproduce interpolating linearly, only two simulations are needed.

The initial electron energies in these two simulations were 5.60 MeV and 7.00 MeV. For the sake of reproducing the field conformation and phantom features used to make experimental measurements, the SSD was 100 cm and a field of 10×10 cm^2 was used for all energies. The dose tallying region was a $40\times40\times50$ cm^3 water phantom. The CPU time of these two simulations was 80 h. To get $\mathrm{PDD}_{20/10}$ values with a relative uncertainty below 2% in 80 h and characterize the dose gradient in the z axis, the phantom was gridded in bins of $3.33\times3.33\times0.2$ cm^3.

C. Simulations

Once the initial beam parameters were determined, a simulation using them was done so as to check whether the simulated dose distribution agrees with the measurements. In this simulation the dose tallying region was divided in $2\times2\times2$ mm^3 bins to ensure a high resolution in all directions. Except for the bin size, field conformation and phantom features were the same as those used to tune the initial beam parameters. The time spent in this simulation was 336 h (14 days) to achieve an average statistical uncertainty of 1.8% in the dose.

The initial beam parameters that characterize the linac employed at the Hospital Clínic were used in the sub-

sequent simulations. In these simulations, the patient models were six CT volumes of a heterogeneous phantom (Figure 1) with various material inserts. Inserts 1 and 2 are made of the same material. These plugs were changed for other material inserts in each simulation and the attenuation of materials was studied. The inserts are equivalent to body tissues, see Table I. The phantom structure is made of Plastic Water® (ρ =1.029 g/cm^3). We can simulate it as solid water because both materials have a similar composition of low atomic numbers elements.

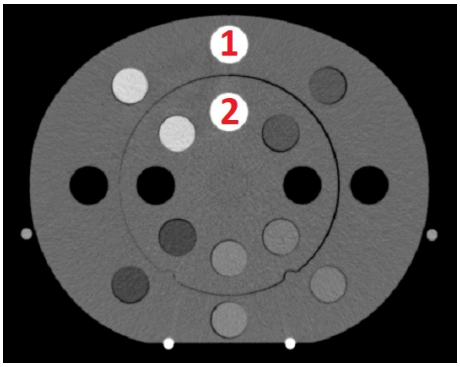

FIG. 1: CIRS CBCT 062 Phantom.

TABLE I: Tissues reproduced by phantom inserts. HA stands for hydroxyapatite.

Tissue	ρ **(g/cm^3)**
Lung (Inhale)	0.205
Lung (Exhale)	0.507
Breast (50% gland / 50% adipose)	0.99
Muscle	1.06
Solid Trabecular Bone (200 mg/cm^3 HA)	1.16
Solid Dense Bone (800 mg/cm^3 HA)	1.53

The field size in these simulations was 4×4 cm^2 and the linac and collimator isocenter was positioned at the point in the volume corresponding to its position in the phantom during the measurements. The bin size, as determined by the imported CT images, was 1.6×3×1.6 mm^3.

A calibration curve, needed to assign physical densities to HU, was created associating HU of each insert to its mass density by means of values obtained with a CT scanner. Due to the filtered back-projection method used to reconstruct CT volumes, HU values distribution was not homogeneous in all the insert, values on the edges were smaller than in the center. In order to avoid this reconstruction artefact, the HU values used to determine the calibration curve were those of the insert center. HU ranges for the material assignment were determined from the standard deviation of HU values in the whole inserts. Once the voxelized geometry was created, one simulation was run for each of the six CT volumes.

The value of splitting factor in s3 was 60 for all simulations so as to reach a 2% relative uncertainty in the dose, the minimum value achievable considering the latent variance of the PSF.

III. RESULTS AND DISCUSSION

The initial electron energy, determined from the tuning process, was 6.15 MeV. Using this value we found $PDD_{20/10}$=0.584±0.018 (2 SD), which agrees with the experimental result, 0.575. The agreement between the simulated and measured PDD curves is shown in Figure 2. The PDDs have been normalised at a depth of 5 cm.

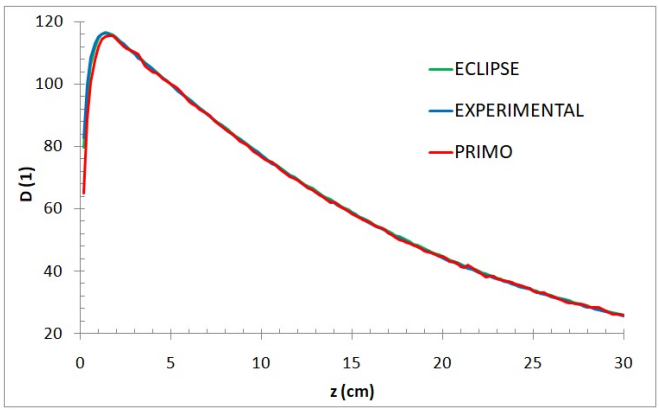

FIG. 2: Measured and simulated depth dose curves comparison for a 10×10 cm^2 field.

The deviations of *Eclipse* and PRIMO results with respect to the measured values for the 10×10 cm^2 and 4×4 cm^2 fields are listed in Table II. These differences correspond to values between the maximum of the PDD at 1.5 cm and a depth of 30 cm. In the case of the 10×10 cm^2 field, both *Eclipse* and PRIMO match with the measured curves. The average deviation of the PRIMO simulation is practically zero. This means that the discrepancies of PRIMO are just simulation statistical noise. On the other hand, the discrepancies of *Eclipse* increase with z (results not shown). The PRIMO results for the 4×4 cm^2 field agree with the experimental values too, but the discrepancies are slightly greater than for the 10×10 cm^2 field.

TABLE II: Eclipse and PRIMO PDD comparison for the 10×10 cm^2 and 4×4 cm^2 fields.

Field (cm^2)	Error (%)	**PRIMO**	**Eclipse**
10×10	Average error	0.035	0.635
	Average absolute error	0.506	0.638
4×4	Average error	−0.325	0.380
	Average absolute error	0.590	0.426

While the initial beam parameters determination leads to good agreement between simulated results and measurements, the dose profile along the x axis at a depth of

10 cm for the 10×10 cm^2 field shows substantial discrepancies in the penumbra (Figure 3). The simulated doses in the penumbra do not match with the experimental values, even considering the statistical uncertainty. This is a consequence of not having determined the focal spot size in the initial beam parameters tuning process. The comparison of dose deviations between *Eclipse* and PRIMO and the measurements, corresponding to the 90% of field size values, are shown in Table III.

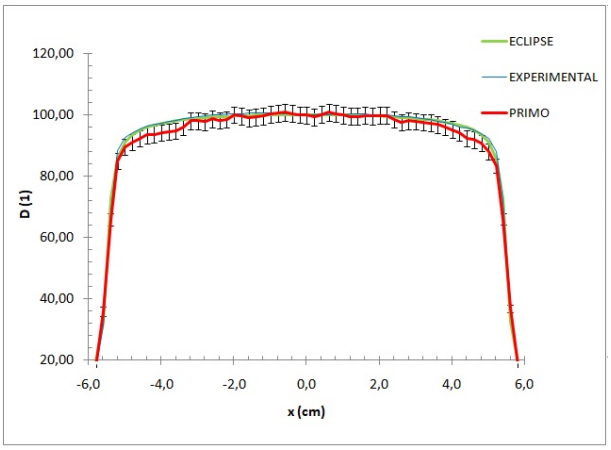

FIG. 3: Comparison of dose profiles at 10 cm depth for a 10×10 cm^2 field.

TABLE III: Eclipse and PRIMO dose profiles comparison at 10 cm depth for a 10×10 cm^2 field.

	PRIMO	**Eclipse**
Average error (%)	1.371	−0.338
Average absolute error (%)	1.436	0.383

These discrepancies in the penumbra may explain why the dose along the z axis differs with respect to the experimental values more in 4×4 cm^2 than in 10×10 cm^2 fields, see Table II. Photons scattered in the penumbra contribute more to the energy deposition in points located on the z axis for smaller fields because of the proximity of these two regions. Thereby, a wrong dose calculation in the penumbra region leads to small discrepancies in dose on the z axis which are not negligible for 4×4 cm^2 fields.

The method used to verify the TPS simulations, with respect to measurements, in Hospital Clínic was planning with *Eclipse* how many monitor units (MU) were required to obtain 1 Gy at the isocenter point in each phantom insert distribution. The measurements show a disagreement between the planned dose, 1 Gy, and the measured values when the linac was configured with MU planned by *Eclipse*. The attenuation calculated by *Eclipse* leads to substantial discrepancies between these values for inserts whose density differs from that of water, see Table IV. In order to compute these deviations, the measurements in heterogeneous media were compared with the value measured in the homogeneous situation, which agree with experimental values, see Table II. To reproduce an homogeneous medium, the insert whose composition and mass density are the closest to these of phantom structure was used. Hence, the breast tissue equivalent plug was chosen as the reference material and the deviations were calculated with respect to the measured value using this insert.

The experimental values are expressed in nC because the deposited energy is obtained from the charge collected in the air that fills an ionisation chamber placed in water. Each ionisation chamber is calibrated and the charge can be converted to absorbed dose in water. As ionisation charge is proportional to dose and due to the same dose value was planned by TPS for all inserts distributions, 1 Gy, the ionisation charge for all measurements must be the same. In these simulations HU were converted into electron densities through the *Eclipse* default calibration curve.

TABLE IV: Measured values at the isocenter position.

Insert	**Measured value (nC)**	**Deviation (%)**
Breast	3.597	—
Lung (Inhale)	3.627	0.83
Lung (Exhale)	3.601	0.11
Muscle	3.604	0.19
Bone200	3.646	1.36
Bone800	3.727	3.61

PRIMO can only be used as a dose verification tool [11], therefore it cannot plan the MU required to deposit a certain amount of energy in a patient model. Hence, to compare measurements and dose values calculated by TPS and by PRIMO, it is necessary that the experimental and *Eclipse* values were measured and calculated, respectively, from the same MU for all insert distributions. The values calculated by *Eclipse* are expressed in Gy to water, experimental ones were measured in nC, as it has been mentioned above, and PRIMO reports dose in units of eV/g per primary particle as dose expressed in the medium where the measurement point is situated. To compare these three values, dose was expressed as a ratio in respect of breast tissue values, $r_i = D_i/D_{\text{breast}}$. The three values are the dose at the isocenter point.

The deviations with respect to the measured values are calculated as

$$\epsilon_{\text{MC}} = 100 \times \frac{r_{\text{MC}} - r_{\text{exp}}}{r_{\text{exp}}} \tag{2}$$

for each insert and are shown in Table V.

Bone tissues discrepancies between PRIMO results and measured values may be the consequence of not having adjusted the focal spot size. As mentioned above, photons scattered in the penumbra region contribute to the absorbed dose along the z axis. For 4×4 cm^2 fields, field conformation in these simulations, wrong dose calculations in penumbra lead to disagreements in z axis points

TABLE V: Discrepancies of calculated values with respect to experiment.

Insert	$\epsilon_{Eclipse}$ (%)	ϵ_{PRIMO} (%)
Lung (Inhale)	−0.82	0.81
Lung (Exhale)	−0.08	−0.03
Muscle	−0.17	0.95
Bone200	−1.33	1.16
Bone800	−3.47	1.01

dose. Due to that effect and taking into account that more photons are scattered in high atomic number materials than in water, the discrepancies are larger in a heterogeneous phantom than in a water phantom.

Regardless this source of discrepancies, the PRIMO results for Bone800 tissue differ less than TPS calculated values in respect of measured ones. *Eclipse* approximations in cross-sections calculations may be the cause of the large disagreement in regions with abrupt changes of density, like at the edge of solid bone inserts.

The discrepancies of PRIMO results, equation (2), are lower than those of *Eclipse* in the majority of inserts distributions except for muscle tissue. This disagreement might be because the *Eclipse* calibration curve reproduces more realistically the material assignment in this HU region than the curve calculated by us.

IV. CONCLUSIONS AND OUTLOOK

The study of depth-dose curves using the $\text{PDD}_{20/10}$ parameter, which specifies the beam quality, has allowed an accurate determination of the initial electron energy of the studied linac. The appropriate adjustment of this initial beam parameter has made it possible to compare PRIMO simulations with the predictions of *Eclipse* and experimental data considering several field conformation and patient models. The discrepancies between these values have revealed some shortcomings of the *Eclipse* TPS, especially in regions with abrupt changes of mass density like in the heterogeneous phantom used in the present TFG.

The simulation of different field sizes highlights the importance of the focal spot size determination in Monte Carlo methods because a wrong value of this initial beam parameter leads to non-negligible calculation errors in dose values, even in regions out of the penumbra, for fields smaller than 10×10 cm^2. The calibration curve plays an important role in CT patient models dose calculations because the attenuation depends on the density assignment. A more accurate adjust of it could have been possible with more HU values associated to known mass densities.

The following steps in the tuning of the initial beam parameters involve a commissioning process [10, 12] which requires more computational resources and CPU time than those we had in this TFG. This determination will lead to a better examination of the weaknesses of TPS and Monte Carlo codes to improve the efficiency of dose planning processes.

Acknowledgments

I thank the helpful guidance and advice of my tutors Dr. José M. Fernández-Varea and Antonio Herreros (Hospital Clínic), without whom the present TFG could not have been completed. I would also like to thank my family for their constant support.

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