Radiation protection study in Intraoperative Radiotherapy: calculation of peripheral dose around a Mobetron using Monte Carlo simulations

Belén Cabañero[1], Belén Juste[1*], Rafael Miró[1], Gumersindo Verdú[1]

[1]*Institute for Industrial, Radiophysical and Environmental Safety (ISIRYM).*
Universitat Politècnica de València. Camí de Vera s/n 46022 Valencia, Spain
*bjuste@upv.es

Abstract. Intraoperative radiotherapy consists of external irradiation during a surgical intervention and is currently one of the most used treatments against cancer. Despite the shielding structures that protect the beam, there is a dispersed dose component that escapes from the accelerator head. The purpose of this work is to study the scattered radiation that is produced by the intraoperative linear accelerator *Mobetron*, installed in a Spanish hospital, San Jaime, in Torrevieja (Alicante). One of the main problems when installing an intraoperative accelerator is the scarcity of this type of radiological protection studies, which are useful in the legalization phases of this equipment. Therefore, this work aims to offer a worthwhile tool for future users that allows the estimation of the dispersed dose in a simple way, as well as the evaluation of the facilities and the necessary shielding to maintain the safety of professionals and patients. To carry out this objective, a 3D model of the linear accelerator head was designed, and after meshing the geometry, detailed simulations were carried out using the Monte Carlo code MCNP, version 6. This study demonstrates the possibility of transferring the use of the Monte Carlo method to radiation protection studies in the field of radiotherapy. After calculating the dispersed dose around the *Mobetron* accelerator, results were compared with experimental data available in bibliography. Theoretical results agree with experimental values, demonstrating that this method is a convenient technique to study the dispersed dose in the vicinity of a linear accelerator.

KEYWORDS: Dispersed dose, intraoperative linear accelerator, Mobetron, intraoperative radiotherapy, IORT, MCNP6.

1. INTRODUCTION

The intraoperative radiotherapy (IORT) is a medical technique where the application of radiation is done in a single session, normally after surgical removal. Recently, there has been an increasing interest in the IORT technique because of the development of mobile electron beams accelerators. This type of machine can be settled directly into an operating area with no need for any special fixed shielding barriers. Furthermore, these mobile machines solve logistical difficulties, such as the need of moving the anesthetized patient, thereby decreasing the overall period of the procedure [1], [2].

In order to prevent too much radiation exposure in surrounding rooms, the maximum beam energy is limited to 10-12 MeV, and a beam stopper is placed under the operating table during the emission of the beam.

According to the principles of radiation protection, the medical use of electron linear accelerators involves a precise analysis of the radiation source and shielding to guarantee that radiation exposure limits stated by the European Community for radiation protection of workers and population are accomplished. The basic techniques of IORT, comprising radiation protection issues, acceptance testing and commissioning, and a recommended quality assurance program for mobile systems were published in an AAPM Radiation Therapy Committee Task Group No.72 report 5 [3]. Consistent with these recommendations, the peripheral dose around an accelerator head of a mobile unit is analyzed in this work.

According to this, the focus of this study is to simulate the head of the intraoperative linear accelerator using a Monte Carlo code. Based on this model, different theoretical studies are carried out to evaluate some relevant aspects of radiological protection related to the use of this equipment in the operating rooms [4].

Results obtained with Monte Carlo simulation offer accurate information about the dose around the linear accelerator, which is a useful tool in radiological protection protocols. For results validation, simulation doses are compared with data available in literature [5], which have been obtained from both experimental measures and computational calculations using other Monte Carlo codes.

2. METHODS AND MATERIALS

Mobetron (IntraOp Medical, Inc., Santa Clara, CA) [6] is a mobile linear accelerator specifically designed for the operating room. This accelerator produces high-energy electron beams (12, 9, 6, 4 MeV) that deliver a uniform dose to a surgically exposed target in a single fraction.

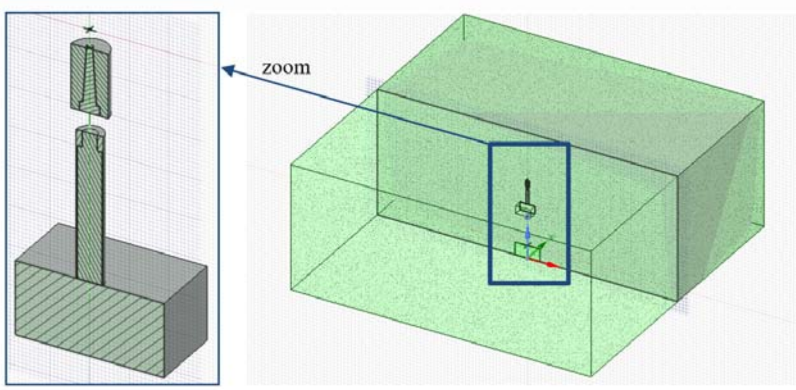
The Monte Carlo code, MCNP6.1.1 (Monte Carlo N-Particle) [7], has been used to build a model of the *Mobetron* treatment head with a 10 cm diameter applicator, emitting the maximum energy, 12 MeV electron treatment beam, and to calculate the amount of peripheral dose around the accelerator and the treatment field.

2.1. Accelerator modeling

The 3D geometry of the linear electron accelerator has been modeled with *SpaceClaim* software. The model is then exported in an appropriate format that can be read by *Abaqus/CAE* software for subsequent meshing. In this work, linear accelerator geometry has been simplified, avoiding components corresponding to the electron creation and acceleration system. Figure 1 shows the sketch of the intraoperative linear accelerator in a longitudinal cut.

The model includes the complete accelerator head (including scattering foils and collimators), the cylindrical IORT applicator, the solid water phantom and the beam stopper.

Figure 1: Detail view of the linear accelerator geometry model *Mobetron* placed at the operating room modeled with *SpaceClaim*.



Mobetron linear accelerator model has been created from geometric and compositional data collected from bibliography, with the aim of dosimetrically characterizing the emitting electron beams. Since the exact data of model is not an available information from manufacturing company, some geometric dimensions have been combined with data from other intraoperative linear accelerator models [5]. Potential dispersion generated by patients is modeled using a 15 cm thick solid water phantom. This phantom is composed by 15 plastic plates made of polystyrene (C_8H_8) with 2% TiO_6.

The flat-tipped applicator is placed over the phantom, and the modeled diameter is the maximum available, 10 cm. The last modeled structure is the beam-stopping, a cylindrical volume of lead placed under the patient's stretcher to ensure greater radiological protection.

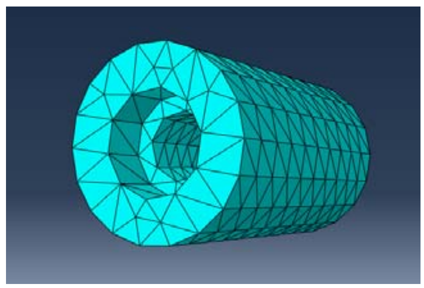
As shown in Figure 1 all the LinAc geometry is embedded in a treatment bunker, which simulates the space in which radiation therapy treatment is performed.

Once the geometry is completed, the *SpaceClaim* file is exported in *.step* format to perform the meshing process with the *Abaqus* software.

To generate the mesh geometry accepted by MCNP6 code, it is necessary to use *Abaqus/CAE* software. This program exports the mesh in *.inp* format, the one needed by MCNP6 to generate the simulation input [8]. Unstructured mesh has been selected in this work, which uses elements of different types or sizes, with the aim of modeling more complex geometries selecting the optimal number of elements. In addition, within the mesh types accepted by MCNP6 (tetrahedrons, pentahedral or first and second order hexahedron) we have used first order tetrahedrons (C3D4). In this case, the mesh is able to adjust the geometry calculations without such expensive as tetrahedrons of second order, where elements rather than straight faces and edges can be curvilinear. Finally, the file *.inp* is generated, where all the geometry information is stored.

Figure 2 shows the aluminum primary collimator of the *Mobetron* accelerator meshed with *Abaqus* software.

Figure 2: *Mobetron* aluminum collimator meshed by *Abaqus*.



2.2. Other Monte Carlo simulation specifications

Numerous input constraints are essential for an accurate simulation, therefore, a detailed physics transport for photons and electrons was defined in this work, including scattering and absorption behavior and medium materials characteristics and different variance reduction techniques.

To that, MODE P, E has been activated to follow tracks of photons and electrons. In all cases, at least 10^8 initial particles were simulated to obtain uncertainties lower than 3% in all evaluated points. Electrons with energies lower than 1 keV were not tracked, although their energy was considered to be deposited locally.

To speed up the calculations without compromising the accuracy of the results, the MCNP6 (version 6.1.1) code has been parallelized in SENUBIO ISIRYM research group´s cluster Quasar using the MPI protocol with 32 processors.

The Monte Carlo details are summarized in Table 1, following the recommendations of the AAPM TG-268 Report [9].

Table 1: Summary of the main characteristics of the Monte Carlo method used.

Item	Description
Code	MCNP6 Version 1.1compiled with Intel v12, with optimization flag O1
Validation	Previously validated
Timing	10 h (CPU time, total sum of 32 parallel processes). All simulations were performed with a number of histories 10^8
Source description	Electron beam model: Gaussian electron distribution (Mean:12 MeV, FWHM:10%)
Cross sections	MCNPLIB84 EL03
Transport parameters	Photon cutoff = 1 keV Electron cutoff = 10 keV

Variance reduction tools	(a) Source biasing (in direction) (b) Cell weighting (importances)
Scored quantities	Absorbed dose in air
Statistical uncertainties	≤3% (k = 2, maximum uncertainty of all calculated quantities)
Post-processing	None

2.2.1. *Beam modelling*

The MCNP simulation model includes the transport of the electron beam directed onto the front of the treatment head and applicators.

The main parameters that determine the electron source energy spectrum, considered as a Gaussian-distributed intensity profile, is the electron average energy (in this case 12 MeV) and the full width at half maximum (FMWH), (in this model, 10%). These parameters were extracted from manufacturer information.

2.2.2. *Results registration tallys*

The Monte Carlo code is based on an iterative statistical process to estimate random pathway transport and interactions of photon and electrons emitted by the sources in three dimensions, having the ability to record results of such interactions particles (*tallies*).

In this model, to obtain the particles flux per cm^2, a specific registration card, called FMESH4 *tally*, has been included in MCNP input. This *tally*, provides flux in units of $\#/cm^2 \cdot s$ (by emitted particle), but flux-to-dose conversion multiplier factors are introduced to get average doses results. I.e., Gy/s received in each point of the defined mesh is obtained. The multiplying coefficient are the mass energy linear absorption coefficients (μ_{en}/ρ) for photons (data obtained from NIST, National Institute of Standards and Technology) [10] and the stopping power for electrons. In addition, it is also implemented another factor that converts the MeV/g to J/kg ($1.60218 \cdot 10^{-10}$).

3. RESULTS

In order to validate the obtained results, an analysis of intraoperative linear accelerator dosimetry studies has been carried out. Given the particular characteristics that mobile LinAcs have compared with conventional accelerators, the information available is limited. For this reason, the information has been collected from the study of dosimetry of the linear accelerator model *Mobetron* 1000 carried out at the Catholic University of Murcia [5], given the similar methodology used.

After calibrating the accelerator by means of relative dosimetry from absorbed dose in water, absolute dosimetry measurements are carried out, using a ROOS flat-parallel ionization chamber and following the TRS-398 protocol of the IAEA [11]. This protocol recommends for mobile accelerators to perform a check under the most unfavorable conditions of maximum energy, 12 MeV, and maximum size of applicator diameter, 10 cm.

These measures are carried out facing the end of the applicator in contact with the water surface, [5]. According to this, values of the reference dose for a 12 MeV energy beam emitted by the *Mobetron* LinAc is 0.966 cGy/MU.

After performing the calibration, experimental measures of scattered radiation are taken around the accelerator. This experimental device measures the dose at discrete points within the operating room around the *Mobetron*, which will be used to compare and verify the results obtained in this work by MCNP6 simulation. Measurement points used are based on the coordinate system with reference in *Mobetron*, being the origin of the coordinate system (0, 0, 0) a point on the ground placed on the axis of the radiation beam, the other coordinates (x, y) of those points are those shown in Table 2.

Table 2: Coordinates (x, y) of discrete points measured around the linear accelerator.

POINT	COORDINATES X (cm)	COORDINATES Y (cm)
A	0	-100
B	100	-100
C	100	0
D	100	100

At each of these points selected, two experimental measures at different heights are made; one at z=107.5 cm, corresponding to height where the phantom is placed, simulating the patient, and the other at z=158.37 cm, corresponding to the height where the second scattering foil of the accelerator is placed.

As presented in the following section, the results obtained at discrete points are expressed in units of Sv/Gy. In order to compare the theoretical results obtained by MCNP6 code simulation it is necessary to divide the absolute doses by the reference dose 0.966 cGy/MU.

3.1. Equivalent dose at discrete points around the linear accelerator

This section shows the results of the equivalent doses, which include the electron and photon component, obtained with the Monte Carlo MCNP6 code at the points previously described (Table 2) around the *Mobetron* linear accelerator for a 12 MeV electron beam.

Table 3 shows calculated equivalent dose values at different points near the *Mobetron* expressed in µSv per Gy of absorbed dose at d_{max} for 12 MeV.

Table 3: Equivalent dose at discrete points around the *Mobetron*.

POINT	Z (cm)	SIMULATION DOSE (µSv/Gy)	EXPERIMENTAL DOSE (µSv/Gy)	ERROR (%)
A	107.5	118.6±0.08	102±20	13
	158.37	36.2±0.09	32±6	11
B	107.5	39.2±0.11	47±9	20
	158.37	14.4±0.12	20±4	42
C	107.5	118.2± 0.08	97±19	17
	158.37	35.5±0.12	30±6	14
D	107.5	42.5±0.08	44±9	4
	158.37	13.4±0.11	19±4	46

As shown in results presented in Table 3, when comparing Monte Carlo doses with those measured experimentally around *Mobetron*, values are very similar, remaining inside the uncertainty range of experimental doses.

Discrepancies observed may be due to the fact that modeling a linear accelerator requires both geometry and materials to be fully known, which it has not been not easily accessible in this work. In addition, there is a great variability between this type of equipment, which has a less standardized manufacturing, because the modeling of each accelerator has to be done individually. Therefore, the main uncertainties are associated with the unknowingness of the exact structure of the accelerator.

On the other hand, there is an uncertainty associated with the calibration factor of the detector with which experimental dose measurements have been performed and which are reflected in the literature [5]. Even if an attempting to minimize this error by controlling the experimental process with a prolonged stabilization process plus prior heating before the measurements is performed, nominal rates of 1000 MU/min, can be ranging from 960 to 1020 MU/min.

Despite these discrepancies, values of the theoretical simulated results using MCNP6 match the experimental measured ones, demonstrating the feasibility of this method to estimate the dispersed dose throughout the space around a linear accelerator without having to use area dosimeters.

3.2. Electrons and photon contribution to the dispersed dose

The use of 12 MeV with a 100 mm diameter applicator is expected to have higher x-ray contamination values because when electron beam energy and applicator diameter increase, the bremsstrahlung tail increases, so we used x-ray values measured with 100 mm applicator in radioprotection calculations to be more conservative.

The following figures (figure 3a and figure 3b) are generated with *Paraview* software from MCNP6 results, and show the electrons and photons dose dispersion around the linear accelerator. Figure 3 displays the operating room plant view where the intraoperative radiation therapy is carried out with the linear accelerator located in the center of the room, at 107 cm height (the upper plane of the water phantom).

Both images present isodose curves at 90%, 50%, 10%, 1% and 0.1% of the maximum doses. Table 4, on the other hand, shows the dose rate values at those isodose curves.

In the left figure (Figure 3a), the maximum dose rate produced by electrons is $7.345 \cdot 10^{-2}$ Gy/s which correspond to the center red colored area. Around it, the dose rate decreases, as indicated in Table 4, below 0.0001 Gy/s at the 0.1% isodose curve of the maximum dose rate. These results demonstrate that outside the irradiation bunker the dispersed dose produced by the electrons is null.

In the right figure (Figure 3b), the maximum dose rate produced by photons is $1.858 \cdot 10^{-5}$ Gy/s which correspond to the center red colored area. Around it, as showed in electron contribution figure, the dose rate decreases as indicated in Table 4, reaching the 0.1% curve of the maximum dose rate where values are below 5.10^{-8} Gy/s. These results demonstrate that outside the irradiation bunker the dispersed dose produced by the photons is also null.

Regarding the displayed planes perpendicular to electron beam direction, the horizontal plane XY, concentric radiation pattern is observed in both cases. As it can be seen, equivalent dose registered is relevant only near the accelerator, where nobody is allowed to stay during the irradiation.

Figure 3: Isodose curves around the *Mobetron* linear accelerator produced by electrons (a) and photons (b). (1. 90%, 2. 50%, 3. 10%, 4. 1%, 5. 0.1%).

(a) (b)

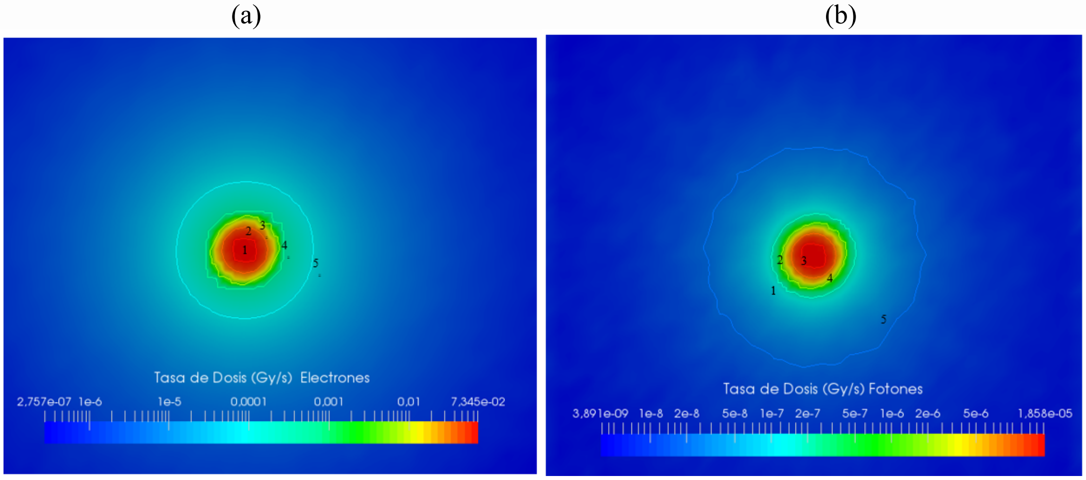


Table 4: Dose rate at different isodose curves.

ISODOSE CURVE	ELECTRON DOSE RATE (Gy/s)	PHOTON DOSE RATE (Gy/s)	TOTAL DOSE RATE (Gy/s)
90% maximum dose	6.61E-02	1.67E-05	6.61E-02
50% maximum dose	3.67E-02	9.29E-06	3.67E-02
10% maximum dose	7.35E-03	1.86E-06	7.35E-03
1% maximum dose	7.35E-04	1.86E-07	7.35E-04
0.1% maximum dose	7.35E-03	1.86E-08	7.35E-03

When comparing electron and photons doses, it can be concluded that the low-energy electron component contributes more to the dispersed dose around the linear accelerator than the photon component. This result has a great importance in the shielding of environmental radiation.

3.3. Maximum workload calculation

From equivalent dose values calculated with the MCNP6 simulation and taking into account the existing structural shielding in each installation, the maximum workload could be calculated to avoid exceeding dose limits for workers in areas adjacent to the operating room, which will allow the future user to evaluate a priori the suitability of facilities prior to commissioning of the equipment.

As the most unfavorable conservative criterion, the estimate of the maximum is done with data calculated for the maximum energy of 12 MeV and maximum applicator size of 10 cm diameter.

It has been assumed in the simulation the typical shields used and an occupancy factor of 1, estimating that personnel who have access to the intraoperative radiation room and contiguous areas is defined as non-professionally exposed personnel, whose the dose limit is 1 mSv/year.

Considering all these assumptions, the maximum workload allowed for the barriers considered in our simulation which are typical of a normal surgical area is calculated. The data is detailed in Table 5.

This is a useful tool to evaluate the IORT facilities before the linear accelerator is started. It should be noted that calculations have been performed under the most unfavorable conditions: maximum energy, 12 MeV, and maximum size of the applicator, 10 cm diameter.

The main purpose of this type of mobile accelerator is to deliver a high dose of radiation immediately after the number of clonogenic cells is reduced by surgical removal. The typical delivered dose during IORT ranges from 10 to 25 Gy. Therefore, it has been assumed an applied dose of 25 Gy per patient, as indicated in literature [5], which includes irradiation during quality control and heating of the equipment.

Taking these specifications into account, the maximum workload is calculated and presented in Table 5.

Table 5: Maximum workload

POINT	Z (cm)	SIMULATION DOSE (μSv/Gy)	Gy/year	PATIENTS BY YEAR	PATIENTS BY WEEK
Out of operation room	100	0.361 ± 0.08901	2770.08	110.80	2

With the assumptions considered, the maximum workload under the most unfavorable conditions is 2 patients/week considering 25 Gy/patient (including irradiation due to quality control and heating of the equipment). This information should be established in the safety study prepared during the installation phase of the equipment for the facility licensing.

4 CONCLUSIONS

This work presents the application of MCNP6 Monte Carlo code for photon and electron transport in intraoperative radiotherapy for radiological protection studies. It is demonstrated the feasibility of roughly modeling an IORT linear accelerator head using Monte Carlo simulation to obtain the dispersed radiation distribution produced around the equipment, which is an important issue in the field of commissioning IORT facilities.

Normally, the used methods to verify the scattered radiation in radiation therapy are analytical methods which are based on simplified equations which are associated with many concerns. Simple closed equations make designing quick, less error-prone, and insightful. Unfortunately, they also oversimplify radiation physics to the point where it is not unusual for radiation measurements to be significantly different from what the equations predicted. Moreover, recently, NCRP No. 144 has recommended the Monte Carlo simulation as a reliable method for the shielding calculations [12].

In this study, dose results obtained with Monte Carlo simulation have been validated with experimental data, so we can also conclude that the calculated dose distributions with this Monte Carlo simulation will be very useful for future users, which will be able to assess a priori the suitability of facilities before the start-up of the equipment and estimate the maximum number of treatments to be carried out while preserving the radiological protection of workers in the surgical area.

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