Romanian Reports in Physics, Vol. 66, No. 1, P. 61–74, 2014

MONTE CARLO SIMULATION OF THE ELECTRON BEAMS PRODUCED BY A LINEAR ACCELERATOR FOR INTRA-OPERATIVE RADIATION THERAPY*

D. MIHAILESCU, C. BORCIA

"Al. I. Cuza" University, Faculty of Physics, Carol I Blvd, No. 11, Iasi 700506,
E-mail: dmihail@uaic.ro, E-mail: cborcia@uaic.ro

Received April 23, 2013

Abstract. In this work, organized as a short review, we present some results obtained by applying the Monte Carlo techniques in the Intra-Operative Radiation Therapy (IORT). The electron beams generated by the NOVAC7 IORT accelerator have been simulated using BEAMnrc/EGSnrc Monte Carlo code, the dose distributions and water-to-air stopping power ratios being later calculated with the associated software (DOSXYZnrc and, respectively, SPRRZnrc). The influence of the cylindrical applicators on the energy spectra and angular distributions of the electrons at the phantom surface has also been investigated.

Key words: IORT, Monte Carlo method, BEAMnrc, DOSXYZnrc, BEAMDP, SPRRZnrc, depth dose distributions, transverse dose profiles, electron energy spectra, electron angular distributions, stopping power ratios.

1. INTRODUCTION

One of the main problems regarding the accelerators dedicated to Intra-Operative Radiation Therapy (IORT) arise from dosimetric characteristics of the electron beams, which are considerably different from those obtained with conventional accelerators. The mobile IORT accelerators are equipped with long cylindrical applicators. Due to their length, these applicators have a major contribution to the energy degradation, as well as to the spatial and angular distributions of the electrons at the phantom/patient surface [1]. Important parameters (as stopping power ratios and perturbation factors) used for absorbed dose determinations, depend on the electron beam characteristics. The correction factors and physical parameters recommended in the international dosimetry protocols are experimentally determined or calculated for conventional electron

* Paper presented at the 1st Annual Conference of the Romanian Society of Hadrontherapy (ICRSH 2013), February 21–24, 2013, Predeal, Romania.

beams. Applying those data to beams such as those used for IORT could be, in principle, not appropriate if the characteristics of IORT and conventional beams are too much dissimilar. In the calibration procedures of the electron beams the accuracy of the dosimetric measurements must be as good as possible and such errors should be avoided as much as possible. Taking into account that some dosimetric characteristics of the IORT beams are difficult or even impossible to be experimentally determined, the Monte Carlo simulations are usually employed.

2. INTRA-OPERATIVE RADIATION THERAPY

IORT (Intra-Operative Radiation Therapy) is a treatment modality of cancer which consists in the direct delivery of a high-level radiation dose (~ 20 Gy) to the residual tumour or to the tumour bed while the target area is exposed during surgery, after the removal of a neoplastic mass (Fig. 1). A well-collimated electron or X-ray beam can be used. IORT involves a single-fraction short time treatment (a few minutes), usually preceded or followed by an EBRT (the External Beam Radiation Therapy – *i.e.* the "classical" radiotherapy). IORT can be also used as a single radiation treatment in the case of the initial (small volume) neoplases and for palliative purposes in the case of unresectable tumours) [2, 3].

Fig. 1 – An IORT electron accelerator (NOVAC 7) during a typical irradiation procedure [3].

The main technical advantages of IORT consist in: (a) the direct visualisation of the target volume; (b) the possibility to protect the healthy tissues by moving them away from the radiation beam (are also used special shields to spare critical organs/tissues (Fig. 2)). Moreover, the use of electron beams allows the delivering of an homogeneous dose to a controlled layer of tissues (the dose falls off rapidly below the target site, therefore sparing underlying healthy tissue – “precision radiotherapy”). Potential side effects associated with conventional radiation therapy, such as irradiation of the skin or of subcutaneous lung and heart tissue, can be minimized or totally eliminated.

Very few disadvantages of IORT can be found in the literature: (a) higher risk of late effects, such as fibrosis, in late responding tissues [5], small risk of infections or the difficulty to monitor dosage to the radiation site [6, 7].

(a) **(b)**

Fig. 2 – An example of IORT procedure used in the treatment of the breast cancer: a) in order to minimize the irradiation of the thoracic wall, before the treatment delivery an aluminium-lead disk of proper diameter is placed between the deep face of the residual breast and the pectoralis muscle; b) close-up view of the proper placement of the electron applicator in the breast. To completely spare the skin from the radiation dose, the skin margins are stretched out of the radiation field using a metallic ring [4].

IORT has been used in the treatment of various malignancies. Local control has always been very high and the toxicity related to the methodology very low. Cancers of the stomach, pancreas, colon rectum and sarcomas, in which the local recurrence is the main cause of failure, have been the objects of numerous clinical studies. The long-term results confirm a positive impact on the local control that is generally associated with increased survival. New fields of application are the cancers of the breast, lung, bladder and uterine cervix [2].

3. ACCELERATORS FOR IORT

The Intra-Operative Radiation Therapy can be performed using electron beams produced by a conventional linear accelerator (a procedure that implies the patient transport, after the chirurgical intervention, to the radiotherapy bunker) or using electron beams produced by mobile *dedicated accelerators* which generate only electrons having 9–12 MeV maximum energy, directly co-located in the operating room. Avoiding the patient transport is the main advantage of the mobile IORT accelerators. Among the disadvantages of these IORT accelerators are mentioned the dosimetric problems which become more complex, compared to those associated with the conventional accelerators.

IORT may also be performed using low energy X-ray beams produced by small sources [8], but this technique is will not be discussed in this work.

The first mobile accelerators dedicated to IORT have been manufactured at the end of the 1990s: **Mobetron** (IntraOp Medical Incorporated, USA) (Fig. 3a) and **NOVAC 7** (Hitesys, Italy) (Fig. 3b). These accelerators, relatively smaller and lighter than the conventional linacs, can be easily transported into the operating room. These mobile IORT accelerators generate high energy electron beams (4, 6, 9, 12 MeV nominal energies (Mobetron) and, respectively, 3, 5, 7, 9 MeV (NOVAC7)), and are equipped with cylindrical applicators of various diameters (3, 4, 5, 6, 10 cm (Mobetron), and 4, 6, 8, 10 cm (NOVAC 7)). The field sizes are given by the applicators diameter. The field having 10 cm diameter is considered as *reference field*. The cylindrical applicators, usually made by polymethyl methacrylate (PMMA), can have an oblique distal part that is tilted with respect to the geometric axis of the beam, with angles ranging from 15° to 45° (base bevelled applicators) [2, 3].

(a) **(b)**

Fig. 3 – The first mobile electron accelerators dedicated to Intra-Operative Radiation Therapy: a) Mobetron (USA); b) NOVAC7 (Italy) [9, 10, 11].

In 2002, in Italy has started the commercialization of a new mobile accelerator dedicated to IORT. LIAC (**L**ight **I**ntraoperative **Ac**celerator) (Fig. 3) has been developed by Info & Tech (Rome), being the result of a collaboration between the firm technicians and a researcher group from ENEA (Ente per le Nuove Tecnologie, l'Energia e l'Ambiente) – Rome (Frascatti), the same that have been developed the NOVAC 7 accelerator. The LIAC accelerator was designed in two variants: (1) 4, 6, 8, and 10 MeV nominal energies and (2) 4, 6, 9, 12 MeV nominal energies. Both variants are equipped with 3, 4, 5, 6, 7, 8, 10 and 12 cm diameter PMMA cylindrical IORT applicators (flat and bevelled at 15°, 30° and 45°) [8].

The object of our investigations was the Italian IORT accelerator NOVAC7 (Figs. 1 and 3b). NOVAC7 generates pulsed electron beams (2–9 cGy/pulse) with four different nominal energies using long PMMA cylindrical applicators having four different diameters. It is equipped with a 3-D movable arm that can be pointed on the operating field (six degree of freedom are allowed). The IORT applicators (Fig. 4), made from a low atomic number material (PMMA) in order to reduce the bremsstrahlung radiation, have the wall thickness of 0.5 cm and lengths 69, 67, 67 and 87 cm, respectively, being available in two different shapes: flat-ended (0°) and bevel-ended (15°, 22.5° and 45°). The source-to-surface distance (SSD) is 80 cm, except for the applicator with the diameter of 10 cm for which the SSD is 100 cm.

Fig. 4 – IORT applicators used by NOVAC7 IORT system.

There are no scattering foils or flattening filters as the spatial uniformity of the treatment field is obtained by the scattering processes of electrons on the applicator wall.

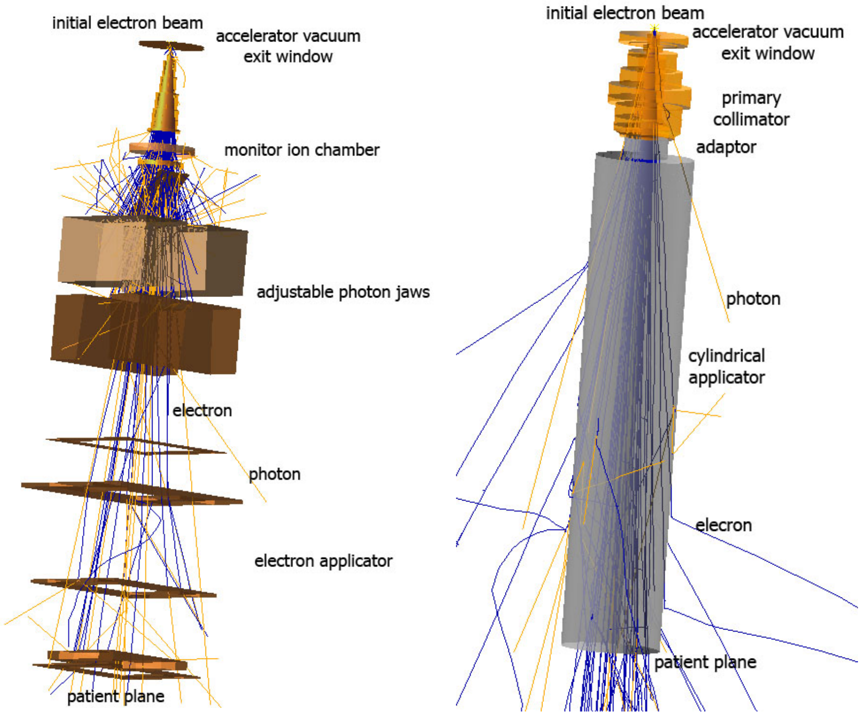
4. ACCELERATOR MODELLING AND SIMULATION OF THE ELECTRON BEAMS

Monte Carlo is considered today to be the most accurate and detailed calculation method in different fields of medical physics: medical imaging (X-ray, nuclear medicine, etc.), radiotherapy (accurate calculation of dose distributions, even in particular configurations; validation of the Treatment Planning System (TPS) and evaluation of their intrinsic limitations), radioprotection, etc.

In radiotherapy, the finality of the Monte Carlo calculation should be the improvement of the treatment planning for which there is always the necessity of a very accurate dosimetry. Dose delivered to the target (tumour) in radiotherapy has very high values (generally 10 times higher than the lethal dose – in the case in which this dose would be absorbed in the entire body). The treatment efficiency depends by the control of dose delivered to the target volume with an accuracy of 5% or even higher in certain situations [12]. Therefore, one of the most delicate task of the physicists in radiotherapy is to ensure the right dose (as high as possible and homogeneous) in the tumour and, in the same time, to plan the radiation treatment in order to minimize the dose in the healthy tissues. In this regard, the Monte Carlo simulation is a very useful tool to understand and compute radiation dose in radiotherapy. The method is also essential in output calibration, providing critical correction factors for international code of practice in clinical dosimetry. Additionally, the Monte Carlo treatment head simulation yields results of unprecedented accuracy and detail of the character of the realistic radiotherapy beams and has proven helpful in treatment head design [13].

This paragraph focuses on modelling and simulation of the realistic electron beams generated by NOVAC7 linear accelerator dedicated to IORT. It is recognized that a detailed and accurate modelling of the irradiation head is a precondition to obtain dose distributions in phantom/patient having the accuracy required in clinical dosimetry.

The electron beams generated by the NOVAC7 IORT accelerator have been simulated using BEAMnrc/EGSnrc Monte Carlo code [14, 15, 16], considered by many to be “the golden standard” in the medical physics community. BEAMnrc is a general purpose Monte Carlo simulation system for modelling radiotherapy sources and is based on the EGSnrc code system [17] for modelling coupled electron and photon transport. BEAMnrc was originally developed as part of the OMEGA project which was a collaboration between the National Research Council of Canada (NRC) and a research group at the University of Wisconsin (USA) [14]. The main goal of this project was to develop a 3-D treatment planning for radiotherapy.



(a) **(b)**

Fig. 5 – Models of irradiation heads of two linear accelerators obtained with BEAMnrc Monte Carlo code and illustrated using EGS_windows: (a) a conventional electron accelerator; (b) the NOVAC7 IORT accelerator.

The irradiation head of the NOVAC7 accelerator was modelled as a series of simple BEAMnrc component modules with cylindrical symmetry centred on the central axis of the beam – *z-axis* (see Fig. 1 from reference [1]). The simulation geometry includes the exit window, the primary collimator, the adaptor, and the applicators (Fig. 5). Shape, dimension, material of the various accelerator components were simulated according to the information provided by the manufacturer. For each simulated beam the complete information (energy, position, direction, charge, etc.) about any particle that crosses a given plane perpendicular to the beam axis (*scoring plane*) was stored in a data file (*phase-space file*).

Illustrations of the NOVAC7 and, for comparison, of a conventional linear accelerator in EGS_windows [18], a 3-D tool for interactively display the geometry of the accelerator and the track history for different particles (part of the OMEGA system), are shown in Fig. 5. In order to analyse the different components of the radiation beams, *i.e.* scattered and direct electrons, bremsstrahlung photons, etc. (see the reference [1]), the LATCH variable available into the BEAMnrc code have been used.

5. DOSE CALCULATION AND EXPERIMENTAL VALIDATION

The Monte Carlo method is an important auxiliary tool in radiotherapy but there is a problem to be solved: the initial beam parameters (spatial fluence distributions, angular divergences and energy spectra) are not exactly known and usually are fixed using a "tuning" procedure involving repeated and time consuming simulations. After a detailed Monte Carlo investigation of the influence of initial electron beam characteristics on the absorbed dose distributions obtained for particle beams generated by NOVAC7 IORT accelerator, we e have found [19] that only the transverse dose profiles for open beams (obtained without IORT applicator) are sensitive to divergence of the initial beam, the other geometrical parameters have no influence on any other dose distribution. On the other hand, the energy parameters (spectral shape, FWHM value, the weight of the low energy component from the energy spectra) influence only the depth-dose distributions curves, results which are quite similar with those obtained by other authors for electron beams generated by conventional accelerators adapted for IORT [20]. Taking into account the results, we proposed a relative short way to find the initial electron beam parameters. Firstly, the divergence of the beam is determined comparing only the calculated and measured transverse dose profiles for open beams. Then, once the geometrical parameters of the beam are known, the energy characteristics of the initial electron beam are determined by a similar procedure only for depth-dose distribution curves.

For comparison with the Monte Carlo calculations, relative absorbed dose measurements (percentage depth dose distributions – PDDs – and transverse dose profiles – TDPs) were performed by a PTW 23343 ion chamber into a PTW automatic water phantom.

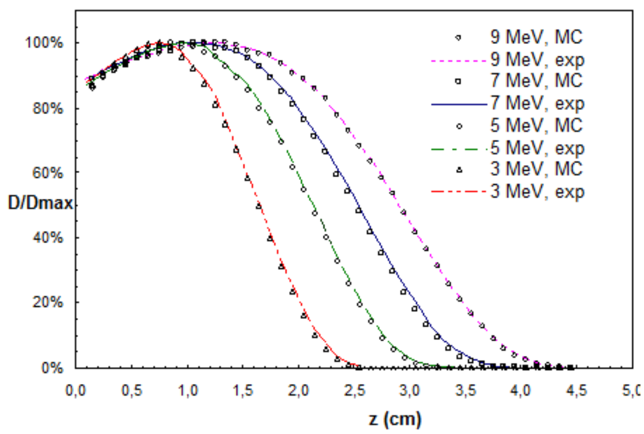
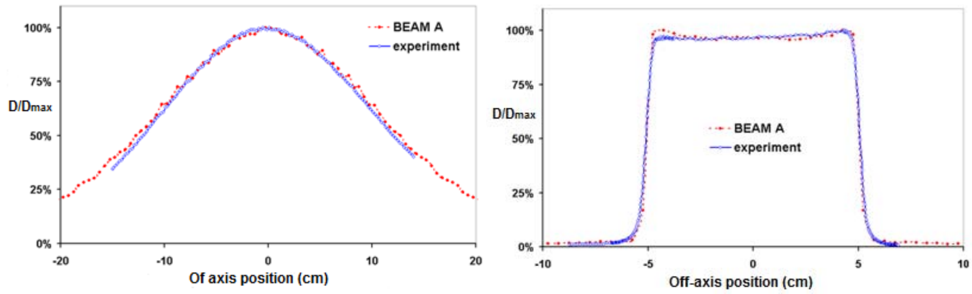


Fig. 6 – Comparison between experimental and Monte Carlo calculated depth dose distributions obtained using an IORT applicator with a diameter of 10 cm; the Source to Surface Distance (SSD) was 100 cm. The surface dose is about 6% greater, compared with conventional accelerators.

Similar dose distributions have been obtained using DOXYZnrc Monte Carlo code [21], an EGSnrc-based Monte Carlo simulation code for calculating dose distributions in a rectilinear voxel phantom, part of the OMEGA-BEAM system of codes developed at NRC.



(a) (b)

Fig. 7 – Comparison of experimental and Monte Carlo calculated transverse dose profiles: a) open 9 MeV beam; b) 9 MeV beam (d = 10 cm IORT applicator, SSD = 100 cm).

Density and material in every voxel may vary and is defined by the user. As input files in our simulations, the phase-space files obtained using BEAMnrc code have been employed. The simulations were determined in voxels of 1 cm × 1 cm × 0.1 cm with statistical uncertainties of about 0.8% (1σ).

After a tuning procedure similar with that described above, an agreement within ±2% or ±1 mm with the experimental dose distributions have been obtained (Figs. 6 and 7) both for open beams and clinical beams, for all nominal energies (3, 5, 7 and 9 MeV) and all applicators (4, 6, 8 and 10 cm).

These results can be considered an evidence that the accelerator was correctly modelled and we can continue our investigation of the dosimetric characteristics of particle beams produced by NOVAC7 IORT accelerator.

6. PHASE SPACE ANALYSIS AND EVALUATION OF DOSIMETRIC CHARACTERISTICS

The phase-space data files generated by BEAMnrc at the phantom surface for each simulated clinical beam have been analysed with BEAMDP (BEAM Data Procesor) [22] a computer program developed to analyse the parameters of the phase-space files generated by BEAMnrc and to derive the data required by a multiple-source model for representation and reconstruction of the electron beam for use in Monte Carlo radiotherapy treatment planning.

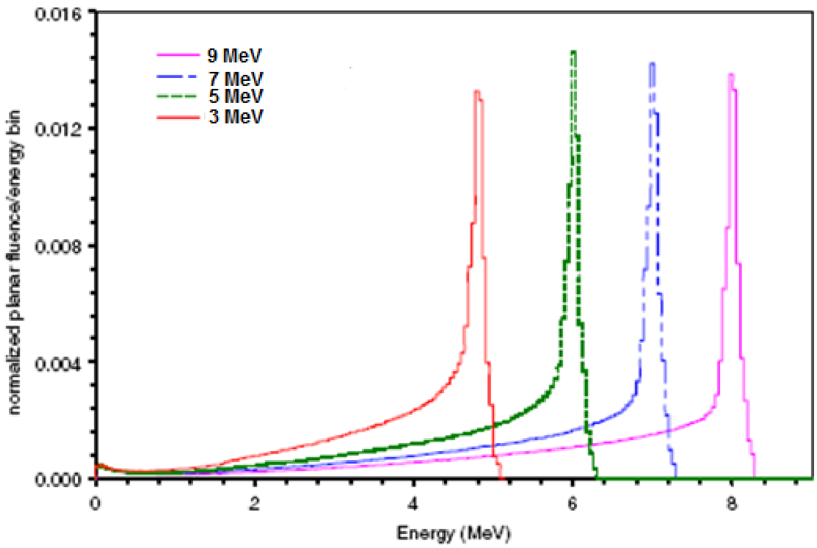


Fig. 8 – Electron energy spectra at the phantom surface for clinical beams generated by NOVAC7 using an IORT applicator with diameter $d = 10$ cm; (D, C, B, A → 9, 7, 5 and 3 MeV nominal energy)

In particular, we used BEAMDP to obtain the electron energy spectra (Fig. 8), electron fluence and the mean energy distributions of the electrons at the phantom surface and at various depths in the water phantom, as well as the angular distributions of the electrons at the phantom surface (Fig. 9).

Similar results as those shown in Fig. 8 have been obtained [1] for the other applicators (4, 6 and 8 cm). According to our data, the energy spectra are independent by the applicator's diameter. All energy spectra are characterized by a low energy continuum which includes about 60% by the total electron fluence. The presence of this "low energy tail" is the main cause of the substantial difference between the mean energy and the most probable energy of the NOVAC7 clinical beams.

In order to determine the beam uniformity at the phantom surface, the fluence and the mean energy of electrons in annular regions around the central axis of the beam were carefully calculated, including the contributions of direct and scattered electrons [1, 23]. We found that the fluence of scattered electrons increases from the beam axis towards the applicator wall. Compared to the direct component, the fluence of scattered electrons is much lower on the beam axis and approximately the same near the field edge. The sum of the two fluence components has, however, a good spatial uniformity. The mean energy of the direct electrons has approximately a constant value inside the field; the mean energy of the scattered electrons slightly increases from the beam axis to the field edge but its value is always much lower than the one for direct electrons. Due to the combined effect of the two components the electron mean energy remains constant within ±1% in a central region of the beam.

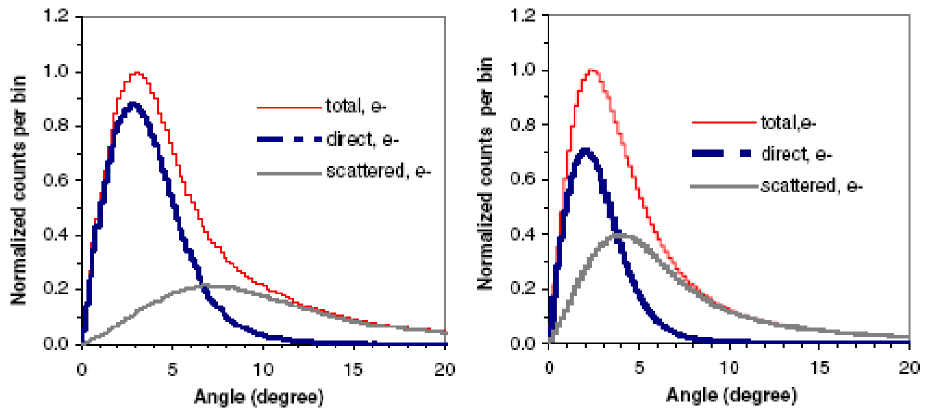


Fig. 9 – Electron angular distributions at the phantom surface for two beams having the same nominal energy (9 MeV) but different field sizes: (a) d = 10 cm; (b) d = 4 cm.

The angular distributions (Fig. 9) are characterized for all the IORT beams by a well-defined peak at small angles, the most probable values, θ_p, being between 2.3° and 4.3°, and a low component at large angles with a maximum angle in the range from 61° to 79°. The shape of the distribution depends on both the beam energy and the applicator size [1]. As expected, the distributions are wider for low energies and large applicators. It is important to note that the angular distributions of the NOVAC7 beams were found to be generally wider than those published in the literature for conventional beams but narrower than those relevant to the IORT beams studied by Björk *et al.* [23].

The same phase-space files were used as source inputs for the SPRRZnrc/EGSnrc code [22] to calculate the Spencer-Attix stopping power ratios of water-to-air, $s_{w,a}$, as a function of depth in water.

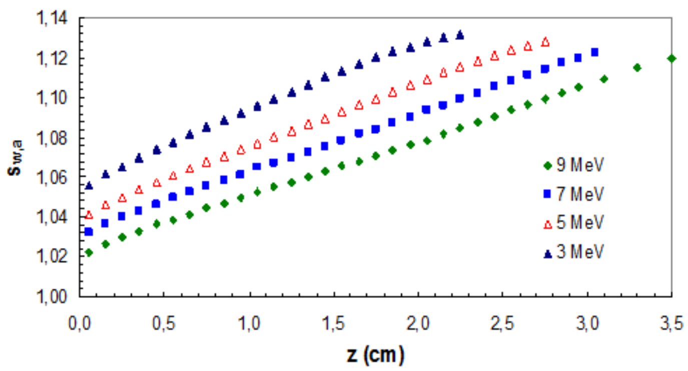


Fig. 10 – Spencer-Attix stopping power ratios of water-to-air calculated using SPRRZnrc Monte Carlo code (results obtained for d = 10 cm IORT).

In Fig. 10 are represented the $s_{w,a}$ values obtained for the thickest applicator (d = 10 cm) *i.e.* the reference beams. To avoid the systematic errors in the determination of the absorbed dose in water (D_w) under non-reference conditions, the variations of $s_{w,a}$ values should be also known for thinner applicators.

Our results [23] shown a small decrease of $s_{w,a}$ for smaller fields. Compared with the reference field, the variations are of maximum 0.1% for applicators with 8 and 6 cm. In the case of 4 cm applicator, the variations of the stopping powers ratios $s_{w,a}$ varies from 0.4% to 0.1% from the phantom surface to R_{50} (the depth at which the absorbed dose decrease to 50% of the maximum dose).

These stopping power ratios $s_{w,a}$ were compared with the corresponding $s_{w,a}$ values recommended by the TRS-381 and TRS-398 IAEA dosimetry protocols [24, 12] in order to estimate the deviations between a dosimetry based on generic parameters and a dosimetry based on parameters specifically obtained for the actual IORT beams. We have found significant deviations from these protocols, but not at the reference depths z_{max} and z_{ref} where values of about 0.4% have been found (Fig. 11). Greater differences (from +0.6% to –1.3%) were revealed along the depth dose curves, results recently confirmed by Righi *et al.* [24], despite of some small constructive differences between the two NOVAC7 accelerators. Thus, at the reference depth, their calculated values, using the same Monte Carlo codes included in the OMEGA-BEAM system, differ from the tabulated values in the TRS-398 dosimetry protocol by only 0.2%.

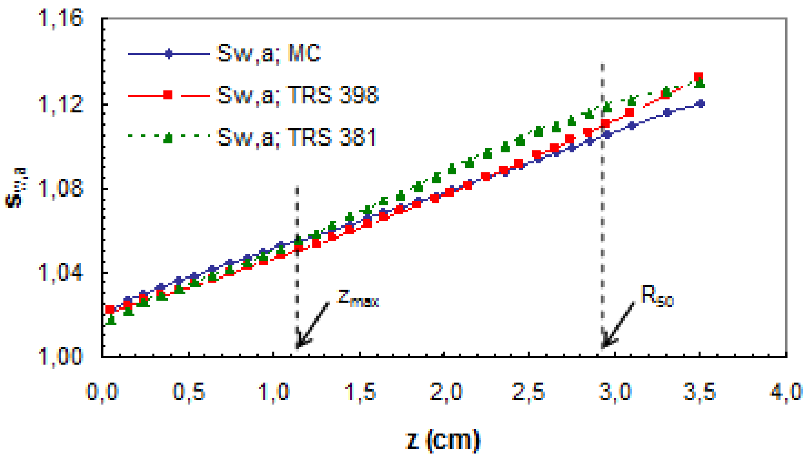


Fig. 11 – Spencer-Attix water-to-air stopping power ratios, $s_{w,a}$, as a function of depth in the water phantom calculated for the 9 MeV IORT beam generated using a 10 cm applicator. The data are compared to the IAEA TRS-381 and TRS-398 AIEA dosimetry protocols.

For LIAC, the other IORT accelerator investigated by these authors, the differences were of 0.4%, this relatively higher value being attributed to the presence in the irradiation head of a thin scattering foil of 85 microns of brass which also significantly increase the surface dose and the photon contamination, compared with NOVAC7.

7. CONCLUSIONS

The clinical beams generated by the NOVAC7 mobile electron accelerator dedicated to IORT have been simulated using BEAMnrc Monte Carlo code. The simulations have been validated comparing the calculated dose distributions in a water phantom with the experimental dose distributions. The main problem was to find the characteristics of the initial electron beams (which are basically unknown). A tuning procedure was used until an agreement of ±2% of the percentage depth dose has been obtained for every simulated beam. Dosimetric characteristics (such as energy spectra, angular distributions and stopping power ratios) of the clinical beams generated by the NOVAC7 IORT accelerator have been determined. Knowing of these characteristics, which generally can be investigated only by Monte Carlo methods and, sometimes, differ significantly from those of classical accelerators, is very important for an accurate clinical dosimetry of the IORT beams.

Acknowledgments. Most of the results presented in this work have been obtained by D. Mihailescu and his colleagues from INMRI in Rome, during a research stage. This author wants to thanks Dr. Maria Pimpinella, Dr. A. Guerra, Dr. C. Caporali and Dr. R.F. Laitano for their kindly support.

REFERENCES

1. M. Pimpinella, D. Mihailescu, A. S. Guerra, R. F. Laitano, *Dosimetric characteristics of electron beams produced by a mobile accelerator for IORT*, Phys. Med. Biol., **52**, *20*, 6197–6214 (2007).
2. A. Rosi, V. Viti, *Rapporti ISTISAN* 03/ *Guidelines for quality assurance in intra-operative radiation therapy*, 2003; http://www.iss.it/binary/publ/publi/0301EN.1109070522.pdf.
3. A. Sam Beddar *et al*, *Intra-operative radiation therapy using mobile electron linear accelerators:* Report of AAPM Radiation Therapy Committee Task Group No. 72, Med. Phys., **33**, *5*, 1476–1489 (2006); http://www.aapm.org/pubs/reports/RPT_92MPSynopsis.pdf.
4. U. Veronesi *et al*, *A preliminary report of intraoperative radiotherapy (IORT) in limited-stage breast cancers that are conservatively treated*, European Journal of Cancer, **37**, *17*, 2178–2183 (2001).
5. *** http://breastcancer.about.com/od/treatmentoptions/f/iort_basic.htm
6. *** http://www.breastcanceriort.org/iort10.html
7. *** http://meditec.zeiss.com/meditec/en_de/home.html
8. 17. G. Tosi, M. Ciocca, *IORT with mobile linacs: the Italian experience*, Oncología, **27**, *6*, 350–354 (2004).
9. *** http://www.intraopmedical.com/mobetron/
10. M. Fantini *et al.*, *IORT Novac7: a new linear accelerator for electron beam therapy*, Front. Radiat. Ther. Oncol., **31**, 54–59 (1997).
11. L. Picardi *et al.*, *The Italian IORT Project*, Proceedings of EPAC 2000, Vienna, Austria; http://epaper.kek.jp/e00/PAPERS/WEP5B04.pdf

12. International Atomic Energy Agency, *Absorbed Dose Determination in External Beam Radiotherapy: An International Code of Practice for Dosimetry Based on Standards of Absorbed Dose to Water*, IAEA TRS-398, IAEA Vienna, 2000.
13. C. Ma, B. Faddegon, B. Curran, *Monte Carlo for Radiotherapy I: Source Modelling and Beam Commissioning for Monte Carlo Treatment Planning*, AAPM 2005 Continuing Education Course; http://www.aapm.org/meetings/05am/pdf/18-4011-55406-780.pdf.
14. *** http://irs.inms.nrc.ca/software/beamnrc/.
15. D.W.O. Rogers, B. Walters and I. Kawrakow, *BEAMnrc Users Manual*, National Research Council of Canada, Report PIRS-0509 (A) revK, Ottawa, NRC, 2007; (http://irs.inms.nrc.ca/software/beamnrc/documentation/pirs0509.pdf)
16. D.W.O. Rogers, B.A. Faddegon, G. X. Ding, C-M Ma, J. We J and T.R. Mackie, *BEAM: a Monte Carlo code to simulate radiotherapy treatment units*, Med. Phys., **22**, 503–524 (1995).
17. I. Kawrakow and D.W.O. Rogers, *The EGSnrc Code System: Monte Carlo Simulation of Electron and Photon Transport*, National Research Council of Canada, Report PIRS-701, Ottawa, NRC, 2011; (http://irs.inms.nrc.ca/software/egsnrc/documentation/pirs701.pdf).
18. *** http://irs.inms.nrc.ca/software/egswindows/
19. M. Oprea, C. Constantin, D. Mihailescu, C. Borcia, *A Monte Carlo investigation of the influence of initial electron beam characteristics on the absorbed dose distributions obtained with a 9 MeV IORT accelerator*, U.P.B. Sci. Bull., Series A, **74**, *4*, 153–166 (2012).
20. P. Björk, P. Nilsson, T. Knöös, *Dosimetry characteristics of degraded electron beams investigated by Monte Carlo calculations in a setup for intraoperative radiation therapy*, Phys. Med. Biol., **47**, 239–256 (2002).
21. B. Walters, I. Kawrakow, D. W. O. Rogers, *DOSXYZnrc Users Manual*, National Research Council of Canada, Report PIRS-794 revB, Ottawa, NRC, 2004.
22. D.W.O. Rogers, I. Kawrakow, J.P. Seuntjens, B.R.B Walters and E. Mainegra-Hing, *NRC User Codes for EGSnrc*, National Research Council of Canada Report PIRS-702, revB, Ottawa, NRC, 2003.
23. M. Pimpinella, D. Mihailescu, A. S. Guerra and R. F. Laitano, *Rapporti dei poteri frenanti acqua-aria per fasci di elettroni prodotti da acceleratori del tipo NOVAC 7*, V Congresso Nazionale AIFM, 17–20 Settembre 2007, Castelvecchio Pascoli (LU), Italy.
24. S. Righi, E. Karaj, G. Felici, F. Di Martino, *Dosimetric characteristics of electron beams produced by two mobile accelerators, NOVAC7 and LIAC, for intraoperative radiation therapy through Monte Carlo simulation*, J. Appl. Clin. Med. Phys., **14**, 1, 6–18 (2013).