

*Research Paper***Carbon-Ion Therapy in the Geant4 Binary Light Ion Cascade Model**S N L SIRISHA¹ and SONALI BHATNAGAR^{2,*}¹*Srinivasa Institute of Engineering and Technology, Krapa Chintalapudi (Post), Katrenikona(M), Amalapuram*²*Department of Physics and Computer Science, Dayalbagh Educational Institute, Agra 282 005, India*

(Received on 14 March 2018; Revised on 23 August 2018; Accepted on 24 August 2018)

The precision results of Monte Carlo simulations are dependent on the consistency of the physical processes applied, particularly for study of nuclear interactions, requiring a validation of nucleus-nucleus versus experimental data for an accurate evaluation of fragmentation products. In this work the Geant4 *Binary Light Ion cascade* model has been used. Not many recent and accurate data are available in literature for carbon ion interactions on thin targets in hadrontherapy, especially for nuclear fragments production. The study of carbon ion interactions on thin targets hasn't been conducted in recent literature at the energy range of interest in hadron therapy (60-400 AMeV), especially for nuclear fragments production. The dosimetric parameters such as peak-plateau ratio, practical range, full width at half-maximum, distal dose fall-off are calculated and compared to the experimental data. Practical range, FWHM and distal fall-off show a very good agreement, considering that the highest discrepancy is of 30 μm (for the FWHM), which is within the position uncertainty of 50 μm . A discrepancy in the peak-plateau ratio of about 9% has been found. This value is larger than the uncertainties of the measured and simulated deposited energy which are respectively of 2% and 1%. The discrepancy is mainly due to the position uncertainty which affects the localization of the measuring point of the Markus chamber in the phantom.

Keywords: PACS-87.53.Ay; 87.53.Jw; 87.53.Kn; 87.55.D-; 87.55.dh; 87.55.dk; 87.55.dk; 87.55.Gh; 87.55.k-; 87.55.km; 87.55.ne

Introduction

Radiation therapy is a method used for cancer treatment in different forms such as external beam radiation therapy including proton therapy, 3-dimensional conformal radiation therapy, intensity modulated radiation therapy and stereotactic radiation therapy. The radiation therapy when implanted is called internal radiation therapy or brachytherapy. The external one damages the genetic part of the target tissue by depositing energy through atomic interactions which helps in destroying the cells so that their growth further discontinues in the area which is cured by radiation (CERN Courier, 2016). The foremost objective of Radio therapy is to exploit the dose within the tumor size and to spare the adjacent tissues. In other words a compromise between the local control of the tumor and the possible emergence of complications has to be found in order to have an

enhanced probability of success in the cure.

Hadrontherapy is one of the techniques in radiotherapy that uses "hadrons", i.e. collimated beams that comprises of composite particles made of quarks, for the decontamination of tumor cells. Several forms of radiation therapy are usually indicated, such as neutrons, protons, pions, antiprotons, light ions and heavy ions. The medical use of protons and carbon ions was originally proposed at the Berkley Cyclotron by Wilson, in order to measure depth dose profiles with a substantial growth in dose at the termination of the particle range, which gives rise to *Bragg peak* (Athar *et al.* 2010). Heavy ion therapy is a novel technique of high precision peripheral radiotherapy that yields a better outlook for cure of radio resistant tumor. The motivation of this transition relies on the fact that heavier ions combine the more favorable dose distribution of the protons to the benefits coming

*Author for Correspondence: E-mail: deisonali.bhatnagar@gmail.com

by an enhanced *Relative Biological Effectiveness* (RBE). Energies of about 400 AMeV (Carnicer *et al.* 2013) ($1A \text{ (a.m.u)} = 12$) are necessary to reach deep-seated tumors. The advantages of carbon ions, indeed, can be summarized as follows:

- Carbon ions deposit most of their energy towards the completion of the assortment, i.e., in the Bragg peak, where they yield stark loss to the cells while being cautious towards the healthy tissues as well as at deeper locations.
- They spare the patient with negligible scattering in lateral and longitudinal distribution, which are about 3 times lower than the protons. Moreover, carbon beams can be formed as narrow focused and scanning pencil beams of variable penetration depth, so that any part of a tumour can be accurately covered.
- Carbon ion beams undergo nuclear interactions along their path inside the tissue, so that positron emitting isotopes are yielded, allowing the use of innovative techniques such as the on-line PET (*Positron Emission Tomography*). The location where the dose is deposited can be determined with a high level of precision, therefore critical structures can be preserved.
- Beams of carbon ions show a favourable depth profile of the RBE. Indeed, in the entrance channel they mostly produce repairable damages, meanwhile in the last part of their path (where the target tumour is present) they cause more serious effects, showing enhanced RBE values in that region. Hence, for radio-resistant tumours they are more effective.

The loss in the intensity of initial beam is studied in the form of Bragg peak as explained below.

The main drawback of using these carbon ions depends on the nuclear fragmentation of carbon ion beams, with the consequent production of charged fragments. The reaction products are mostly produced by projectile fragmentation, so that they are characterized by an emission velocity close to that of the primary particles. Final result is the formation of a typical “tail” of dose just behind the Bragg peak, which is due to the contribution of fragments created along the path of primaries. The higher is the energy of the incident ion, the lower is the peak, because of a

great number of inelastic interactions, and the greater is the amount of extra-dose in the tail. Fragmentation reaction may occur already at the stage of acceleration of the beam, especially when passive scattering systems are used as beam delivery systems. Anyhow, the contribution of fragments production coming from the interaction of the beam along the beam transport line can be drastically reduced by means of relatively simple technical structures.

The incident particle interacts with the tissues and undergoes fragmentation giving rise to a Bragg peak tail. When critical structures are close to the distal part of the Bragg peak, where an undesired amount of dose is needed then it is to our disadvantage. However, consequences arising from carbon ion fragmentation have to be undertaken not only beyond the Bragg peak, but also in the whole radiation field, because nuclear reactions occur as early as initial entry into the patient-tissue. This fact means that, as the primary carbons traverse the matter a continuously growing number of secondary particles is produced “travelling” together with the primary ions and giving its contribute in terms of dose. In other words, a *mixed radiation field* is created along the depth, composed mostly of the primary but also by a not negligible number of nucleons and light ions. What complicates the description is the fact that each isotope is characterized by a unique RBE value. Biological effect, indeed, depends on various quantities, such as the atomic number and the local ionization density of the specific particle considered. The latter is quantitatively described by the *Linear Energy Transfer* (LET) (Dowdell *et al.* 2012). All these aspects get more importance in the case of using more than one irradiation field, creating even more complexity in mixed radiation fields.

Monte-Carlo Techniques to Simulate Hadron-therapy

The precision results of Monte Carlo simulations are dependent on the consistency of the physical processes applied, particularly for study of nuclear interactions. Hence, there should be validation of nucleus-nucleus versus experimental data for an accurate evaluation of fragmentation products. The treatment planning systems which are in use nowadays for the treatment of radiotherapy are optimized using Monte-Carlo techniques. There is a wide spread use

of Monte Carlo codes which are interfaced to real time commercial treatment planning systems in this area. Geant4 is an object-oriented tool kit developed for simulation-based studies of high energy physics experiments with their applications in medical science and astrophysics fields. It has wide variety characteristics for implementing along with geometrical description, primary particle generation, physical processes, visualization and analysis techniques which can fulfill the requirements in optimizing the experimental setup for any application. GATE (Geant4 Application for Tomography Emission) is developed to achieve an integrated, multipurpose, scripted simulation toolkit that can adapt the field of nuclear medicine for tomography applications. GAMOS (Geant4 based Application for Medicine Oriented Simulations) is another framework which is based on Geant4 along with other codes based on Geant4 such as MULASSIS (MUlti-LAYered Shielding SIMulation Software), GRAS (Geant4 Radiation Analysis Software), GATE, PTSIM (Particle Therapy Simulation) and TOPAS (Tool for Particle Simulation). GAMOS is used as an edge to reduce the complications that may arise in other simulation codes such as peculiar volume shape, a novel primary generator, and position distribution or in physics processes. All Monte Carlo codes have further sub systems such as random number generator, rules to sample probability distributions and sets of probability density functions. The accuracy of the code mostly depends on the (i) Particle interactions, (ii) The components of the detector, (iii) Validation of the code.

There are at least four ingredients which are vital in order to understand the basic strategy of Monte Carlo simulations. These are: (i) Random variables: characterized by a domain, contains all possible values that the random value may take. (ii) Probability Density Function (PDF), (iii) Moments of a Probability Density Function, (iv) Significant variance σ^2 . The probability distribution of particle transport can be implemented through different Monte-Carlo simulation algorithms. The first algorithm is Detailed MC Simulation. The electron scattering, back scattered electrons i.e., angular distribution and the reflection coefficient are described by Poisson stochastic process. This method is suitable for materials with low and intermediate atomic numbers. For large kinetic energies the average number of collisions is very large and the detailed simulation becomes very inefficient. High Energy

Simulation codes such as EGS4 (Electron and Gamma Transport Simulation Code), Geant4 (Geometry and Tracking 4) and ETRAN (Electron Transport) use condensed simulation algorithm, in which global effects of collisions are simulated at end of a track segment. The displacement and particle deflection are global effects which are computed from multiple scattering theories of Moliere, Gaudsmith Saunderson (MolierMSC_Bethe_Phys.REv.v89_pg1256-1266_1953) and Lewis|. The theories of Moliere and Gaudsmith Saunderson gives only the angular distribution after tracking of each particle while Lewis theory computes the moments of spatial distribution as well (Cirrone *et al.* 2011). The third kind is mixed simulation method which is also presently in use for Geant4 toolkit. It simulates hard collisions and uses a multiple scattering theory to treat the effects of soft collisions at the end of a given step. This algorithm reduces the dependence of angular distribution of particle tracking on step length which results in decrease of uncertainties in results.

In the present work, the Monte Carlo code *Geant4 Binary Ion Cascade* has been used to study the fragmentation of carbon ion beams at energy range of interest in hadrontherapy. A general description of its main features together with the physical models of interest, are discussed in section III. The accuracy of a Monte Carlo simulation is related to the reliability of the physical processes implemented, in particular the nuclear ones. Hence, for a realistic estimation of fragmentation products, nucleus-nucleus models inside the code have to be validated versus experimental data. The study of carbon ion interactions on thin targets hasn't been conducted in recent literature at the energy range of interest in hadrontherapy (60-400 AMeV), especially for nuclear fragments production. Hence, a preliminary validation of the nucleus-nucleus *Binary Light Ion cascade* model has been carried out. The results have been obtained using the *Hadrontherapy* application, which is an advanced example included in the public release of the Geant4 toolkit. It has been previously developed by the group for the simulation of the CATANA proton beam line, and here customized on purpose in order to reproduce the experimental beam line used for the carbon ion Bragg peak acquisition. An estimation of the dose and fluence contribution due to the charged fragments produced by thick PMMA targets have been carried out to study the

primary particles fragmentation. The method could also allow a better interpretation of biological responses obtained with human cell growth irradiations (discussed in section IV), whose results are still under analysis.

Simulation of Carbon-Ion Therapy

Carbon ions depth dose distributions inside the water phantom have been reproduced by simulating the LNS (Laboratori Nazionali del Sud) beam line used for the experiment. The *Hadrontherapy* Geant4 advanced example has been used at this aim. It is an open source application initially developed to simulate the passive proton beam line of the CATANA facility, at LNS, for the proton treatment of ocular tumours. Carbon beams at 62 AMeV are transported in the “CATANA room” (Fig. 1), where a passive beam line for delivering of the dose to the patient has been setup. The beam line is composed by a set of scattering systems and beam delivery monitors which have been completely simulated with Geant4 inside the *Hadrontherapy* application.

The *Hadrontherapy* application has been modified in order to reproduce the experimental setup used at the “0 degree” beam line, where the voxelised phantom is placed at the end of beam line. The human phantom has been partitioned into voxels which are defined in detector construction. The beam lines are blue tracks hitting the phantom within which is the tumor at a particular depth (as voxels). In the Detector Construction class the vacuum area with the kapton window has been implemented together with the

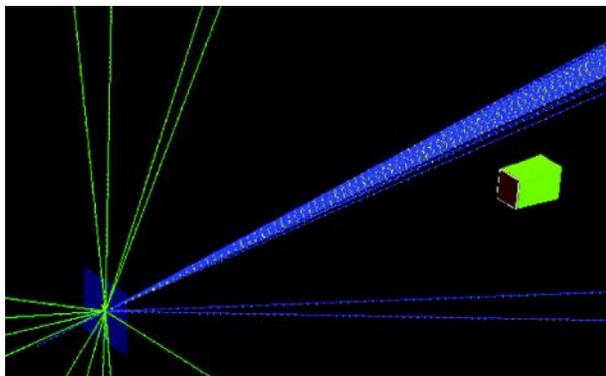


Fig. 1: Snapshot of the Geant4 simulation of the 0-degree beam line, the 2-d plane is the Kapton window and the ionization chamber. The positive charged particles are denoted in blue rays and green are considered as neutral particles

monitor chamber and the water phantom, placed at the same distances as in the experiment. The ionization chamber has been reproduced placing inside the simulated phantom a cubic “sensitive detector” 40 mm of side, divided in 4000 slabs of 10 μm of thickness orthogonal to the beam axis. The sensitive region represents part of the phantom where the dose deposition algorithm works.

This thickness has been chosen in order to have a good spatial resolution in the simulated results. From each slab it is possible to retrieve the total energy deposited by carbon ion beams as well as by the produced secondaries and to store it in ASCII or ROOT files. Each slab simulates the active volume of the ionization chamber. Hence, in a unique simulation run the total energy deposited along the penetration depth is calculated and the Bragg curve reproduced for further comparison with the data. In Fig. 2 a representation of the sensitive detector divided in slabs is showed.

Physical models have been activated by simply recalling the appropriate *Physics Lists* which is defined as precompiled models used as inputs in the Geant4 simulation. *Hadrontherapy* application, indeed, has been developed in order to allow the user to implement physical processes with all the possible methods available in Geant4: by hand, using the *Physics Lists* or using the packed Reference *Physics Lists*. In particular the *G4EmStandard-Physics_option3* physics list has been activated for the electromagnetic processes, which shows an enhanced level of accuracy on energy loss (this option3 version has been

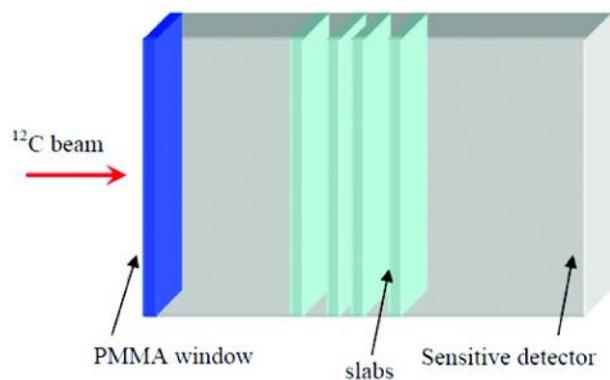


Fig. 2: Scheme of the sensitive detector implemented in the Geant4 simulation. Dose released by the carbon in beam is stored in the slabs composing the phantom

expressly developed by Geant4 collaboration to meet requirements of medical applications, where high level of accuracy is required). The nucleus-nucleus inelastic interactions have been simulated implementing the *Binary Light Ion Model*, together with the deexcitation models for the evaporation of nucleons and light fragments in the equilibrated stage. De-excitation stage has been activated implementing the same limits (step size definitions in simulation) used for the neutron production simulation and charged fragments simulation. Once the geometry set-up has been defined and the physical processes activated the final information *Hadrontherapy* application requires is the primary beams characteristics. A nominal energy of 744 MeV with a spread of 0.744 MeV (i.e., 0.1% of the nominal energy) has been considered in the simulation. This value of energy spread has been chosen after comparing the results of simulations run with different energy spread values with the experimental Bragg peak. The nominal spread value (i.e. 0.744 MeV) showed a better agreement to experimental data of LNS and, thus, it was finally adopted for the production of the results.

A production cut of 0.1 mm has been adopted, after having been verified not to affect final outcomes and as well as to reduce calculation time. A total of 10000 primary events have been shot, which give statistical uncertainties within 1%. The Geant4 version 9.3.p04 has been used. In Fig. 3 a snapshot from the Geant4 simulation is shown, where the carbon ion

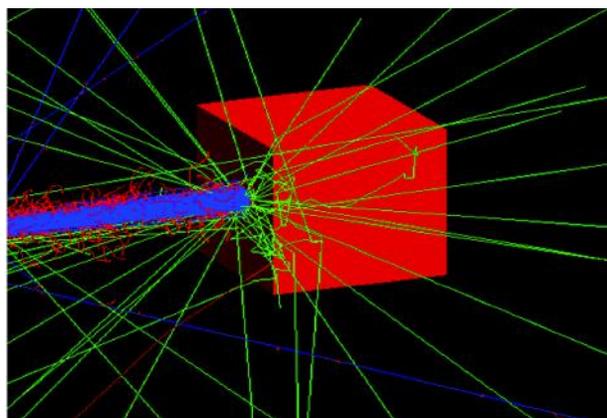


Fig. 3: Snapshot of the Geant simulation. ^{12}C particles that exists from the left side (blue rays) interacts with the water phantom (red), producing secondary particles. Red tracks represent negative charged particles and green ones neutral particles

beam (blue tracks) impinging the water phantom (red cube) is clearly visible and producing secondary particles (green and blue tracks) and electrons (red tracks).

The Bragg curve obtained with the Geant4 Monte Carlo simulation has been compared with that obtained with experimental data. Total dose distributions are, of course, the final result of different effects and processes which occur at each depth position, from the incident particle energy up to the zero value. Along the path of primary carbon ions, ionization and excitation processes are produced and secondary particles are emitted for nuclear interaction. Secondaries release their energy ionizing the material traversed and can, in turn, create other particles as electrons and gammas. Hence, different particles coincide and add to a total energy deposit which, therefore, is the integral effect of various physical processes suffered by primary particles and its products. However, depth dose distributions are of great importance in hadrontherapy because they give crucial information about some dosimetric quantities used also as input parameter for treatment planning. So a comparison between Bragg curves, even if it is somewhat of integral, indicates if the predicted distribution is accurate or not. In the specific case of the present study, it specifies how well the Monte Carlo simulations replicate the energy loss processes suffered by the incident and secondary particles. From a quantitative point of view, the agreement can be also evaluated by calculating some of the mentioned dosimetric quantities, such as:

- *Peak-plateau ratio*, defined as the proportion of the energy deposited at the position of the peak and its entrance.
- *Practical range*, defined as the distance from the entrance surface of the beam to the distal point where the 10% of the maximum dose is measured.
- *Full width at half maximum (FWHM)*, distinct as the size of the Bragg peak at the 50% positioned corresponds to its maximum dose value.
- *Distal dose fall-off*, defined as the space among the 80% and 20% positioning of absorbed dose along the beam axis of the Bragg peak.

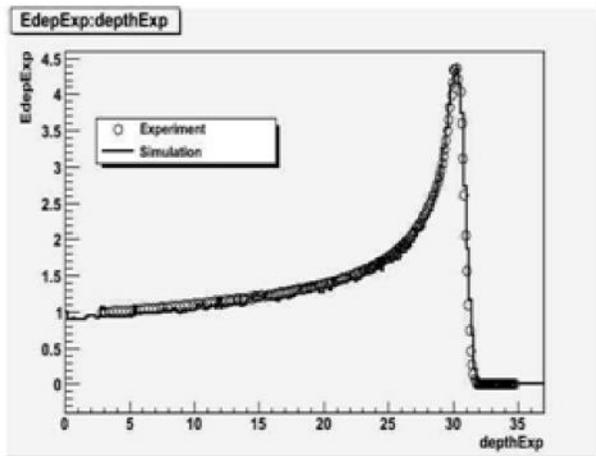


Fig. 4: Bragg curve of a 744 MeV of ^{12}C beam on water. The points give the experimental data acquired at LNS while the black line represents the Geant4 results

Comparison between experimental Bragg peak and that obtained with the Geant4 Hadrontherapy application is shown in Fig. 4, where the Bragg curves have been normalized to the area (beam line area was known and the sensitive region has been simulated accordingly). The depth dose distributions are in good agreement, as visible by the superimposition of the curves, both in the entrance channel (plateau) and at the peak, where the simulation slightly overestimates the released dose due to different detector placements.

In Table 1, the corresponding values of the quantities defined above both for the experimental data and the simulated peak are summarized. Practical range, FWHM and distal fall-off show a very good agreement, considering that the highest discrepancy is of 30 μm (for the FWHM), which is within the position uncertainty of 50 μm .

Anyway, a discrepancy in the peak-plateau ratio of about 9% has been found. This value is larger than the uncertainties of the measured and simulated deposited energy which are respectively 2% and 1%. The discrepancy is mainly due to the position

uncertainty which affects the localization of the measuring point of the Markus chamber in the phantom. Indeed, if we look at the Bragg peak and the gradient of dose just before and after that, it is evident that very small uncertainties in position can cause large difference in the Bragg curve values. This is a typical characteristic of carbon ions Bragg curve, especially for low energy incident particles, as in this case. In this configuration the peak is very steep, the gradient in the region of the peak is high and it can be difficult, from the experimental point of view, to put the detector at the peak position. This problem is not found when the incident energy of carbon ion beams is higher (200-400 AMeV) or, as an example, in case of proton beams. Proton beams accelerated at the same energies, indeed, exhibit distal fall-offs roughly one order of magnitude higher and peak-plateau ratios slight larger than half of the carbon ion case. Moreover, at these low energies, the fragment tail is not clearly visible because few fragmentation events occur in this case.

Biological Effects of Ion Irradiation

In addition to the advantages of the depth-dose profile as discussed earlier in the section III, heavy ions are effective in treating tumors due to the enhanced biological effect of high-LET particles. The LET (related to stopping power) depends quadratically on the projectile charge (Z_p^2) and results in large values for heavy ions. Typically, the large energy deposition in the center of ion tracks result in more severe DNA damage with respect to low-LET irradiation. Since the ionization density of light ions is larger for the low energetic high-LET particles as present in the tumor region relative to the swift ions in the entrance channel (often referred to as “plateau”), the biological effect in the target volume is more pronounced than in the surrounding normal tissue. The most common method adapted in radiation protection of estimating the biological response of ions relative to conventional radiation is the use of weighting factors, formerly known as quality factors. For an accurate estimate of the efficacy of ions, the concept of the RBE must be applied. The RBE is defined as the ratio of the dose of X-rays divided by the dose of ion irradiation that result in the same biological effect. It depends on many different parameters such as the biological end point, dose, particle type, and energy as well as the tissue under consideration. As a result, the RBE is

Table 1: Comparison between dosimetric parameters obtained with the experimental data and with Geant4

Parameters	Experimental data	Geant4 simulation
Peak to Plateau Ratio	6.44	7.04
Particle Range	10.59	10.58
FWHM	0.48	0.51
Distal Fall-off	0.10	0.09

different for every location in the treatment field. Therefore, the increased biological effectiveness must be thoroughly considered for heavy ion treatment planning and poses a big challenge for correct beam delivery. In the following section, we define the relative biological effectiveness and discuss its dependence on various parameters relevant for carbon-ion therapy. Subsequently, we present the pioneering research performed at LNS (Italy) that led to the first biologically optimized heavy-ion treatments (Cirrone *et al.* 2011). The description of the strategy of the worldwide second facility for carbon-ion therapy located at the HIMAC at the NIRS in Chiba, Japan, follows, which is based on an extensive data collection of cell experiments combined with their experience of radiation therapy with neutrons. This approach was also adapted at the HIBMC in Hyogo (www.hibmc.shingu.hyogo.jp/past/english/ionbeam_treatment.html). The most recent concept was developed at GSI (Weinrich, 2016) and uses the LET (light energy transfer) to determine the photon - equivalent dose. Additionally, we address current research topics in the framework of heavy-ion therapy such as the effect of different oxygen levels, cell transformation, and the induction of secondary cancer.

Radiation Damage by Photons and Heavy Ions

The most striking difference between photon and ion irradiation concerns the microscopic spatial energy distribution. In the case of photons the energy is transferred to the cell either by photo effect or by Compton effect depending on the energy of the penetrating photon. Since the cross sections for these processes are rather low, the number of ionization events per incident photon within the volume of a cell is also small. Typically, only a few electrons are ejected from (Moyers and Vatsnitsky, 2012) target molecules possibly ionizing further molecules if they have received enough energy during their primary interaction. Due to this low number of events, many photons are required to deposit a relevant dose. Since these photons are randomly distributed, the resulting ionization density can be assumed to be homogenous over the entire cell volume.

Microscopic Track Structure of Ion Beams

The spatial distribution of energy is entirely different for heavy ions than for photons (Moyers and Vatsnitsky, 2012). It is this localized energy distribution associated

with ion beams that results in a typically larger biological effect induced by particles. The radial dose distribution around ions is governed by two steps. First, electrons (often named “secondary or electrons”) are emitted in ion-atom or ion molecule interactions by means of Coulomb interaction of the projectile and the target. Second, the liberated electrons are scattered by frequent interactions with the medium (Newhauser, 2015).

The dominant ionization process can be described by the binary-encounter approximation assuming that the projectile collides with a quasi-free electron. Another source of energetic electrons originates from the Auger effect, which is the expulsion of outer electrons in the process of filling inner-shell vacancies created by direct Coulomb collisions. A third important interaction mechanism produces electrons. These electrons are either lost or picked up into unbound states of the projectile thus being sharply emitted into forward direction. Those primary electrons induced by interaction of the projectile and the target are subsequently transported through the medium by elastic and inelastic collisions. The last process merely leads to change in direction whereas in the Auger effect process energy is transferred to the medium by either ionization or excitation. Most of the induced electrons receive either only a small energy transfer or they are scattered in the forward direction, depositing most of the dose in the center of ion tracks. However, the electrons that are fast enough to leave the track core typically undergo a large number of interactions. Due to those frequent scattering processes, the initial preference of electrons in the forward direction diminishes, resulting in a broad angular distribution. All existing models, analytical or Monte Carlo simulations, as well as experimental studies show a steep radially symmetric dose distribution with a negative gradient for an increasing distance r , approximately following $1/r^2$ dependence.

DNA Damage and Cell Inactivation

The distributions of photons and ions interacting with DNA molecule at different energies can qualitatively understand the larger radiation damage done by ions. The typical extension of the track center with the highest “local” dose is on the order of nanometers, thus resulting in a large probability of correlated nearby DNA damages like single or double strand breaks or

base damages. In contrast, the approximately homogeneous dose distribution of photons generates much larger distances between neighboring damage sites. Since the cell's repair capability is reduced for more complex DNA damage, the radiation damage of heavy ions is larger than that of photons. Typically, one distinguishes DNA damage induced by *direct* hits of the ion or its surrounding secondary electrons and *indirect* DNA damage generated by radiation induced radicals. The contribution of indirect damage (about 70%) is larger than the DNA damage by direct hits (about 30%) for low-LET radiation. For high-LET carbon ions, the contribution of direct hits is slightly increased.

A common way to analyze the different effects of photons and heavy ions is by means of cell survival curves. These experiments are relatively easy to perform and they have a high significance for radiation therapy because they give insight into the potential of radiation to kill tumor cells. In the standard experimental protocol, cell proliferation is analyzed about 1-2 weeks after irradiation and cells are counted as survivors, if they have formed a colony with more than 50 daughter cells (Newhauser *et al.* 2014). The surviving fraction is given by normalization to the number of seeded cells. The most common way to parameterize the cell survival S uses the linear-quadratic (LQ) model (Newhauser *et al.* 2014), $S(D) = \exp(-\alpha D - \beta D^2)$, where D is the absorbed dose and α and β are experimentally determined parameters. The ratio α/β determines the shoulder of the survival curve and represents an important quantity in conventional radiotherapy.

Conclusion

Monte Carlo methods represent one of the most effective tools for verification of dose computation in radiotherapy with carbon ion beams. The physical

processes involved in this novel radiation therapy technique, as well as the involved biological aspects give light to the description and level of complexity which analytical methods and experiment do not always cover at all. Monte Carlo simulations are able to achieve a more accurate description of the physical processes, taking into account the effects due to the primary particles as well as to the secondary ones produced along the path in the matter. The tracking and the consequently interactions of each particle produced are possible for all the energy range of interest, and realistic geometrical configurations can be implemented in detail, taking into account the presence of different material compositions. If reliable results have to be expected, the physical models implemented inside Monte Carlo codes have to be validated versus experimental data. In particular, the accuracy of nucleus-nucleus interaction models is crucial in carbon ion therapy, in order to have a reliable prediction of the produced nuclear fragments.

In this work, the Monte Carlo toolkit Geant4 has been used. An extended validation of the Geant4 nucleus-nucleus models at the energy range of interest in hadrontherapy has not been fully achieved yet, especially for nuclear fragments production predictions on thin targets. The work discussed in this paper represents a contribution in this direction. Indeed, the results carried out and discussed have allowed, at first, to compare the Binary Light Ion Cascade model implemented in the Geant4 toolkit with experimental data found in literature. Afterwards, the code has been used in order to estimate the "radiation quality" of carbon ion beams used in radiotherapy, by computing the dose due to the charged fragments produced and calculating the depth dose distribution of Linear Energy Transfer (LET) characterizing both primaries and the different isotopes produced.

References

- Athar B S, Bednarz B, Seco J, Hancox C and Paganetti H (2010) Comparison of out-of-field photon doses in 6 MV IMRT and neutron doses in proton therapy for adult and pediatric patients *Phys Med Biol* **55** 2879-91
- Carnicer A *et al.* (2013) Quantification of dose perturbations induced by external and internal accessories in ocular proton

therapy and evaluation of their dosimetric impact *Med Phys* **40** 061708

- Cirrone *et al.*, Progress in Nuclear Science and Technology, Vol. 2, pp. 207-212 (2011)

- Dowdell S J *et al.* 2012 Monte Carlo study of the potential reduction in out-of-field dose using a patient specific aperture in pencil beam scanning proton therapy *Phys Med Biol* **57** 2829-42

MolierMSC_Bethe_Phys.REv.v89_pg1256-1266_1953

Moyers M F and Vatnitsky S M (2012) *Practical Implementation of Light Ion Beam Treatments* (Madison: WI: Medical Physics Publishing)

Newhauser W D, Giebeler A, Guo B and Zhang R (2014a) Uncertainty in dose permonitor unit estimates for passively scattered proton therapy: II. The role of compensator and patient scatter in thoracic cases in preparation *Phys Med* **2** 26pp

Newhauser and Wayne D, The physics of proton therapy (2015) *Phys Med Biol* **60** R155-R209

Proton therapy enters precision phase, CERN Courier, October issue 2016

Weinrich U, GSI, Darmstadt, Germany, Proceedings of EPAC 2006, Edinburgh, Scotland, www.hibmc.shingu.hyogo.jp/past/english/ionbeam_treatment.html.