A Didactic and Interactive Simulator in Radiotherapy

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Abstract

Radiotherapy is one of the main therapeutic modalities in oncological treatment, making the understanding of the physical interactions of radiation with matter fundamental for its clinical application. This work presents the development of an interactive and didactic web simulator designed to illustrate and compare the relative dose deposition profiles of heavy charged particles (protons and alpha particles) and photon beams in different biological tissues. The computational core was developed in JavaScript, implementing phenomenological and analytical physical models. To prioritize didactic simplicity, the photon module acts as an interactive qualitative illustrator operating within a continuous clinical range of 6 to 15 MV, employing the pure exponential Beer-Lambert law based on tabulated mass attenuation coefficients (μ/ρ) [1]. For protons and alpha particles, an iterative numerical integration of the Bethe-Bloch equation with relativistic corrections was implemented to calculate energy deposition in real-time and model the Bragg Peak. The results graphically demonstrate the fundamental differences in particle behavior, highlighting the ballistic superiority of heavy ions, which spare posterior healthy tissues, in contrast to the continuous exponential dispersion and exit dose inherent to photons. It is concluded that the simulator demonstrates significant potential in achieving its educational role, facilitating the visualization and communication of complex Medical Physics concepts. Future perspectives involve quantitative validation via Monte Carlo simulations to refine the inaccuracies of the adopted simplified models.

1. Introduction

Cancer comprises a group of diseases characterized by the uncontrolled growth and proliferation of abnormal cells in the body [2]. Under normal physiological conditions, human cells grow and multiply through cell division to form new cells as the body needs them [2]. When they age or become damaged, a programmed cell death mechanism (apoptosis) is activated [2]. However, due to the vast amount of cell divisions that occur continuously, failures in DNA repair and verification mechanisms can happen, allowing damaged cells to ignore stop signals and multiply anomalously to form tumors [2].

Currently, there are more than 100 types of cancer, which are classically categorized according to the organ or tissue of origin, as well as the type of cell or genetic mutation involved [2]. Such genetic alterations can be inherited (internal factors) or triggered by environmental exposures and lifestyle habits (external factors), such as ultraviolet radiation and chemicals present in tobacco smoke, which cause direct damage to the DNA [2]. To illustrate the critical impact of these external factors, it is estimated that approximately 87% of lung cancer cases in men and 84% in women in the United States are attributable to smoking [3].

Given the complexity of the disease, modern clinical management involves diverse therapeutic modalities, often combined. Chemotherapy, for example, acts systemically—usually via intravenous injection—using drugs to attack rapidly proliferating cells. Another important branch is immunotherapy. Cancer cells are often able to "hide" from or deceive the body's defenses; therefore, immunotherapy consists of treatments designed to assist or restore the ability of the patient's own immune system to detect, attack, and eliminate tumor cells [2].

The central treatment addressed in this article, however, is radiotherapy. This technique uses high-energy radiation to damage the genetic material of tumor cells, preventing their replication. Radiotherapy can be applied with curative intent (active), focused on destroying the target tumor, or in an adjuvant manner (preventive), seeking to eliminate residual cells after surgical interventions and prevent local recurrences [4, 5].

The relevance of the study and the continuous improvement of oncological therapies is evidenced by alarming epidemiological data. An extensive statistical survey of the United States population demonstrated that the probability of developing some type of invasive cancer over a lifetime is about 39.2% for men (approximately 1 in 3) and 38.7% for women [3]. On a global scale, the estimates are equally impactful, pointing out that about 1 in 5 people in the world will develop cancer during their lifetime [6].

Considering the vital importance of this area and the need to improve the teaching of its associated technologies, the objective of this study is to present an interactive simulator developed to demonstrate, in a simplified, didactic, and intuitive way, the operating principles of oncological treatment by radiotherapy. The simulator covers the relative dose deposition of distinct radiations: photons, protons, and alpha particles (Helium nuclei). Due to their fundamental physical natures, these particles interact with matter in very different ways, thereby determining distinct clinical indications and applications [7, 5].

The design of the simulator aims to broaden the knowledge of the area and facilitate the transmission of medical physics

concepts to future generations. By translating the physics of radiation into an accessible visual interface, the tool not only helps laypeople understand the process but also presents great utility for the hospital environment. Physicians and physicists can use it in clinical practice to transparently explain the treatment to patients, contributing to the demystification of cancer and radiotherapy, and consequently, reducing the fear and anxiety frequently associated with the therapy.

2. Development

The computational core of the simulator relies on specific mathematical models to describe how photons, protons, and alpha particles interact with biological tissues. Because these particles possess fundamentally different physical properties, characterizing their dose deposition profiles requires distinct computational approaches. These physical differences ultimately dictate their specific clinical applications in oncology, as detailed in the following subsections.

2.1. Photons

Photons, widely employed in conventional radiotherapy in the form of megavoltage X-rays, are clinically generated in Linear Accelerators (LINACs). The process occurs through the acceleration of electrons to high energies which, upon colliding with a high atomic number metallic target (usually tungsten), produce electromagnetic radiation by deceleration (*Bremsstrahlung*).

Unlike heavy particles, photons have no rest mass or electrical charge. Biologically, they are characterized by a Low Linear Energy Transfer (low LET). Upon penetrating the tumor tissue, they create sparse ionization events, inducing simple genomic lesions, such as oxidative damage, abasic sites, and DNA single-strand breaks (SSBs), in addition to rare double-strand breaks (DSBs) [7]. Tumor cells attempt to correct these injuries using common repair pathways, such as Base Excision Repair (BER) and Non-Homologous End Joining (NHEJ). Therapeutic success occurs when the continuous accumulation of damage overloads these repair mechanisms, leading the cell to programmed death [4].

Physically, the absence of electrical charge means that photons do not undergo continuous Coulomb interactions, leading instead to a probabilistic exponential attenuation as they penetrate matter [8]. In the clinical context, this attenuation is primarily governed by three interaction mechanisms: the Photoelectric effect, Compton scattering, and Pair Production. To translate this behavior to the computational environment of the simulator while maintaining didactic simplicity, the photon module was designed to function as a qualitative interactive illustrator rather than a highly complex iterative numerical simulator.

Therefore, the photon dose deposition curve is strictly modeled by the exponential Beer-Lambert Law:

$$D(x) = D_0 \cdot e^{-\mu x}$$

Where D_0 represents the incident intensity (or dose) of the beam prior to attenuation, and μ is the total linear attenuation coefficient of the material. To ensure physical accuracy without unnecessary computational overhead, the coefficient μ is derived directly from the tabulated total mass attenuation coefficients (μ/ρ) extracted from National Institute of Standards and Technology (NIST) databases [1], multiplied by the respective biological tissue's volumetric density (ρ).

Although the clinical dose distribution of a megavoltage beam presents an initial *build-up* region followed by an approximately exponential decay, the present simulator represents only the latter region. This simplification was adopted to emphasize the didactic comparison between the continuous attenuation of photons and the ballistic behavior of charged particles.

The simulator enables dynamic exploration of photon beams across a continuous clinical range of 6 to 15 MV. This interval was chosen because it corresponds to the energies most widely used in external radiotherapy. Energies below 6 MV generally do not exhibit sufficient penetration for deep-seated tumors, while energies above 15 MV significantly increase the production of secondary neutrons in the accelerator head, making them less common in clinical practice. This physically justifies the implemented range. As the user slides through the 6 to 15 MV range, the algorithm performs real-time linear interpolation of the μ/ρ values, providing an immediate graphical illustration of the beam's continuous attenuation and exit dose.

However, continuous dose deposition remains an intrinsic physical limitation of this uncharged modality. This unpreventable exit dose has driven the development of therapies based on heavy charged particles that pause definitively within the target tissue.

2.2. Heavy Charged Particles: Protons and Alpha Particles

Although the clinical use of charged particles has been expanding, therapies based on protons and heavier ions emerge as modalities that seek to bypass the ballistic limitations of photons. In contrast to conventional radiation, protons and alpha particles, having mass and positive electrical charge, interact in a distinct and more biologically controllable manner.

Upon penetrating the tissue, they initially lose little energy; however, as they decelerate, their ionization density (LET) rises sharply, depositing most of their dose at a well-defined depth, followed by a steep decay immediately after. This phenomenon forms the Bragg Peak [7] and significantly reduces the exit dose. This high ionization density in tumor regions deposits dose in the form of clustered lesions, generating the so-called Complex DNA Damage (CDD).

CDDs consist of multiple lesions formed in close proximity on the same DNA helix [7]. The therapeutic advantage lies in the inhibition of cellular repair: this level of complex damage presents a high difficulty in being restored by tumor enzymatic mechanisms, favoring the Homologous Recombination (HR) pathway and directing the cell towards death [7]. Additionally, these particles induce higher levels of Reactive Oxygen Species (ROS), contributing to cellular apoptosis [7].

To computationally model this ballistic phenomenon that characterizes the Bragg Peak in real-time, the simulator calcu-

lates the Stopping Power, defined as the amount of energy the particle loses per unit distance ($-dE/dx$). Since oncological ions reach significant fractions of the speed of light, the numerical core utilizes the complete Bethe-Bloch Equation with relativistic corrections [9]:

$$-\left\langle \frac{dE}{dx} \right\rangle = Kz^2 \frac{Z}{A} \frac{1}{\beta^2} \left[\frac{1}{2} \ln \left(\frac{2m_e c^2 \beta^2 \gamma^2 T_{max}}{I^2} \right) - \beta^2 \right]$$

In this expression adapted for the numerical code, the constant and explicit terms are defined as:

- $K \approx 0.307075$ MeV·cm^2/mol: binding constant that encompasses fundamental physical parameters;
- M: rest mass of the incident particle (used indirectly in the kinematic calculation of γ);
- $m_e c^2 \approx 0.510998$ MeV: electron rest energy;
- z: electrical charge of the incident particle;
- Z/A: ratio between the atomic number and the atomic mass number of the target tissue;
- T_{max}: maximum kinetic energy transferable in a single collision;
- I: mean excitation potential of the biological tissue.

To highlight the ballistic superiority and differences in interaction, the simulator was expanded to include Helium nuclei (alpha particles). The inclusion of the alpha particle alters the Bethe-Bloch calculation in two fundamental parameters. First, its rest mass ($M \approx 3727.38$ MeV/c^2) is roughly four times that of a proton ($M \approx 938.27$ MeV/c^2), which directly modifies the relativistic velocity kinematics (β and γ) for a given kinetic energy. Second, and most importantly, the alpha particle possesses a double electrical charge ($z = 2$). Because the stopping power is proportional to the square of the charge ($z^2 = 4$), an alpha particle loses energy approximately four times faster than a proton traveling at the same speed. This results in a dramatically denser ionization track and a correspondingly shorter range in tissue.

To validate the numerical accuracy of the iterative method implemented for charged particles, a comparative analysis between the theoretically calculated values from the algorithm and the reference data obtained from the NIST databases (PSTAR for protons and ASTAR for alpha particles) [1] was performed. Tables 1 and 2 detail the comparative values for Stopping Power (SP) and Continuous Slowing Down Approximation (CSDA) Range.

This quantitative foundation ensures that the algorithm implemented in the simulator calculates the relative dose and the Bragg Peak with an adequate correspondence to the qualitative clinical behavior. The stopping power distribution is strictly employed as an approximation of the dose deposition in the tissue. It is highlighted that the model does not include the density effect correction ($\delta(\beta\gamma)$), which is relevant only at very high energies, nor the shell correction—both of which are outside the practical and didactic scope intended by this simulator.

3. Clinical Applications

The understanding of the physical principles governing the interaction of radiation with matter is what underpins its clinical applications. Intuitively, proton therapy presents itself as a favorable modality for deep-seated tumors adjacent to organs at risk (OARs). Thanks to the dose deposition in the Bragg Peak and the absence of a prolonged exit dose, protons allow for a more precise dose conformation, sparing neighboring tissues [5]. In contrast, photons prove to be viable and widely used solutions when peripheral tissues show a higher tolerance to certain doses of radiation.

Clinical practice demonstrates that the theoretical physical basis requires practical considerations to determine therapeutic superiority. A clinical example of this dichotomy occurs in the treatment of prostate cancer. Comparative studies have recorded that, in certain scenarios, patients treated with protons showed a higher incidence of gastrointestinal toxicity (such as rectal bleeding) when compared to those treated with photons [5]. This phenomenon stems from range uncertainties and the constant internal motion of pelvic organs. The ballistic precision of the proton beam makes it sensitive to daily anatomical changes; a displacement of the prostate can direct the Bragg Peak towards the intestinal wall. In these cases, the more diffuse dose distribution of photons (via IMRT) dilutes the impact of positioning uncertainties [5].

On the other hand, in pediatric tumors, especially in the Central Nervous System (CNS), the physical benefits of protons tend to align with clinical expectations. Due to the radiosensitivity of developing tissues, the dose provided by the scattering and exit of photons can lead to long-term side effects [4]. Clinical evaluations revealed that children subjected to craniospinal irradiation with photons presented a greater decline in Intelligence Quotient (IQ) compared to those treated with protons [5]. The reduction of the exit dose in proton therapy favors the preservation of the patients' neurocognitive development [10].

From the perspective of intra-body interactions, photons possess useful characteristics due to their progressive attenuation. While they interact with the cellular medium predominantly via the Photoelectric Effect and Compton Scattering, they pass through the patient's body entirely, depositing residual dose in healthy tissues along the exit path [4].

The impact of these differences in dose deposition is related to discussions regarding the emergence of radio-induced secondary malignancies. It is known that cancers caused by the dose received by healthy tissues can take 15 to 20 years to manifest. Although dosimetric studies predict a reduction in these risks with the use of protons [10], the adoption of this technology still requires monitoring. Long-term epidemiological data is still being consolidated to definitively attest to this clinical advantage [10].

It becomes evident, therefore, that the choice of therapeutic modality requires a careful clinical analysis of each case. This decision-making complexity reinforces the utility of visual tools that assist both in the education of new professionals and in communication with patients, a premise that bases the design of the interactive platform presented below.

Table 1: Comparative analysis between the values calculated by the Simulator and the PSTAR reference data for Protons.

Energy	SP Calc (MeV·cm^2/g)	SP PSTAR (MeV·cm^2/g)	Error SP	CSDA Calc (g/cm^2)	CSDA PSTAR (g/cm^2)	Error CSDA
Water (Liquid)						
50 MeV	12.4574	12.4500	0.06%	2.2217	2.2270	0.24%
60 MeV	10.7854	10.7800	0.05%	3.0873	3.0930	0.19%
70 MeV	9.5628	9.5590	0.04%	4.0742	4.0800	0.14%
80 MeV	8.6280	8.6250	0.04%	5.1770	5.1840	0.13%
90 MeV	7.8894	7.8880	0.02%	6.3907	6.3980	0.11%
100 MeV	7.2904	7.2890	0.02%	7.7107	7.7180	0.09%
125 MeV	6.1925	6.1920	0.01%	11.4489	11.4600	0.10%
150 MeV	5.4451	5.4450	0.00%	15.7674	15.7700	0.02%
175 MeV	4.9031	4.9030	0.00%	20.6162	20.6200	0.02%
200 MeV	4.4921	4.4920	0.00%	25.9516	25.9600	0.03%
Adipose Tissue						
50 MeV	12.7693	12.8300	0.47%	2.1631	2.1560	0.33%
60 MeV	11.0496	11.1000	0.45%	3.0077	2.9970	0.36%
70 MeV	9.7928	9.8380	0.46%	3.9713	3.9560	0.39%
80 MeV	8.8325	8.8740	0.47%	5.0484	5.0280	0.41%
90 MeV	8.0739	8.1120	0.47%	6.2342	6.2090	0.41%
100 MeV	7.4589	7.4950	0.48%	7.5242	7.4920	0.43%
125 MeV	6.3323	6.3630	0.48%	11.1789	11.1300	0.44%
150 MeV	5.5658	5.5930	0.49%	15.4030	15.3300	0.48%
175 MeV	5.0101	5.0350	0.49%	20.1475	20.0500	0.49%
200 MeV	4.5889	4.6110	0.48%	25.3696	25.2500	0.47%
Cortical Bone						
50 MeV	10.6766	11.1100	3.90%	2.6040	2.5080	3.83%
60 MeV	9.2545	9.6350	3.95%	3.6133	3.4780	3.89%
70 MeV	8.2131	8.5530	3.97%	4.7629	4.5820	3.95%
80 MeV	7.4161	7.7240	3.99%	6.0465	5.8140	4.00%
90 MeV	6.7856	7.0690	4.01%	7.4580	7.1690	4.03%
100 MeV	6.2740	6.5360	4.01%	8.9923	8.6420	4.05%
125 MeV	5.3353	5.5600	4.04%	13.3333	12.8100	4.09%
150 MeV	4.6955	4.8930	4.04%	18.3433	17.6200	4.10%
175 MeV	4.2311	4.4100	4.06%	23.9641	23.0100	4.15%
200 MeV	3.8787	4.0430	4.06%	30.1450	28.9400	4.16%
Lung (Calculation Data Only)						
50 MeV	12.3285	-	-	2.2451	-	-
60 MeV	10.6740	-	-	3.1196	-	-
70 MeV	9.4640	-	-	4.1169	-	-
80 MeV	8.5390	-	-	5.2312	-	-
90 MeV	7.8080	-	-	6.4576	-	-
100 MeV	7.2153	-	-	7.7913	-	-
125 MeV	6.1288	-	-	11.5683	-	-
150 MeV	5.3891	-	-	15.9317	-	-
175 MeV	4.8527	-	-	20.8309	-	-
200 MeV	4.4459	-	-	26.2217	-	-

Table 2: Comparative analysis between the values calculated by the Simulator and the ASTAR reference data for Alpha Particles.

Energy	SP Calc (MeV·cm^2/g)	SP ASTAR (MeV·cm^2/g)	Error SP	CSDA Calc (g/cm^2)	CSDA ASTAR (g/cm^2)	Error CSDA
Water (Liquid)						
50 MeV	152.4923	151.7000	0.52%	0.1822	0.1845	1.25%
60 MeV	131.4529	130.9000	0.42%	0.2531	0.2557	1.03%
70 MeV	115.9262	115.5000	0.37%	0.3343	0.3372	0.86%
90 MeV	94.4443	94.2100	0.25%	0.5266	0.5300	0.64%
100 MeV	86.6774	86.4900	0.22%	0.6373	0.6409	0.56%
125 MeV	72.3064	72.1900	0.16%	0.9547	0.9588	0.43%
150 MeV	62.3990	62.3200	0.13%	1.3281	1.3330	0.37%
175 MeV	55.1318	55.0800	0.09%	1.7554	1.7600	0.26%
200 MeV	49.5610	49.5200	0.08%	2.2345	2.2400	0.25%
250 MeV	41.5608	41.5400	0.05%	3.3416	3.3480	0.19%
Adipose Tissue						
50 MeV	157.1284	157.3000	0.11%	0.1762	0.1772	0.55%
60 MeV	135.3365	135.6000	0.19%	0.2450	0.2459	0.35%
70 MeV	119.2717	119.5000	0.19%	0.3240	0.3246	0.20%
90 MeV	97.0714	97.3600	0.30%	0.5110	0.5111	0.02%
100 MeV	89.0527	89.3400	0.32%	0.6187	0.6185	0.03%
125 MeV	74.2280	74.5000	0.37%	0.9277	0.9264	0.14%
150 MeV	64.0179	64.2700	0.39%	1.2916	1.2890	0.20%
175 MeV	56.5341	56.7700	0.42%	1.7082	1.7040	0.25%
200 MeV	50.8009	51.0200	0.43%	2.1755	2.1690	0.30%
250 MeV	42.5730	42.7600	0.44%	3.2560	3.2450	0.34%
Cortical Bone						
50 MeV	129.1911	133.6000	3.30%	0.2167	0.2110	2.70%
60 MeV	111.5739	115.5000	3.40%	0.3003	0.2918	2.91%
70 MeV	98.5412	102.2000	3.58%	0.3959	0.3840	3.10%
90 MeV	80.4619	83.5200	3.66%	0.6219	0.6018	3.34%
100 MeV	73.9104	76.7500	3.70%	0.7517	0.7268	3.43%
125 MeV	61.7661	64.2000	3.79%	1.1236	1.0850	3.55%
150 MeV	53.3755	55.5100	3.85%	1.5604	1.5050	3.68%
175 MeV	47.2106	49.1200	3.89%	2.0596	1.9850	3.76%
200 MeV	42.4784	44.2100	3.92%	2.6189	2.5220	3.84%
250 MeV	35.6722	37.1400	3.95%	3.9095	3.7620	3.92%
Lung (Calculation Data Only)						
50 MeV	150.8954	-	-	0.1841	-	-
60 MeV	130.0789	-	-	0.2558	-	-
70 MeV	114.7163	-	-	0.3378	-	-
90 MeV	93.4609	-	-	0.5322	-	-
100 MeV	85.7756	-	-	0.6440	-	-
125 MeV	71.5555	-	-	0.9647	-	-
150 MeV	61.7520	-	-	1.3421	-	-
175 MeV	54.5607	-	-	1.7738	-	-
200 MeV	49.0481	-	-	2.2580	-	-
250 MeV	41.1314	-	-	3.3766	-	-

4. Methodology

Based on the structured physical models, the adopted methodology consisted of the development of a *client-side* web application, utilizing HTML5 and CSS3 markup and styling languages for the visual structuring and responsiveness of the interface. The computational core, responsible for interactivity and modeling of particle transport in real-time, was programmed entirely in JavaScript (*Vanilla JS*). For the continuous and dynamic rendering of graphs, the *Chart.js* library was implemented, while the *MathJax* library was used for the vector formatting of the differential equations displayed on the platform's information panel.

The physical simulation of the heavy charged beams (protons and alpha particles) required the target tissue to be spatially discretized in continuous microscopic steps ($dx = 0.05$ cm). Since the linear energy transfer (dE/dx) dictated by the Bethe-Bloch Equation [8] is dependent on the variable velocity of the particle as it crosses the medium, the Euler Numerical Method was used to solve the iterative integration of the trajectory. At each iteration of the algorithm, the relativistic kinematic parameters (β and γ) were recalculated, and the corresponding energy loss was subtracted from the incident energy until the particle reached a cutoff energy of 0.5 MeV for protons or 1.0 MeV for alpha particles. This method allowed the modeling of the initial plateau and the subsequent rise in energy deposition, culminating in the Bragg Peak. At the end of the integration, a mathematical smoothing function (*Gaussian blur*) was applied to simulate the statistical dispersion of energy (*energy straggling*) that occurs in radiotherapeutic practice.

In the case of photon beams, the attenuation curve was programmed into the simulator functioning as an illustrator, restricted to a continuous clinical energy range of 6 to 15 MV. The dose relative decay algorithm calculates the pure exponential attenuation using tabulated total mass attenuation coefficients (μ/ρ). For any user-defined energy within the 6 to 15 MV slider, the JavaScript interpolates the corresponding μ/ρ values derived from standard physics tables. The physical behavior is then scaled continuously across the depth axis, reflecting simple physical principles without complex real-time numeric integration algorithms.

Finally, to ensure the consistency of the approximations executed in the code, the physicochemical properties associated with the simulated biological tissues (skin, lung, cortical bone, and adipose tissue) were standardized. The values of volumetric density (ρ), mean excitation potential (I), and the effective ratio of atomic number to mass ($\langle Z/A \rangle$) — specifically 0.555 for water, 0.550 for lung, 0.500 for cortical bone, and 0.554 for adipose tissue — were extracted from the standards of the *International Commission on Radiation Units and Measurements* [11] and the NIST databases [1].

5. Results

The development of the project resulted in an interactive simulation platform with a primarily didactic purpose. In addition to fostering the dissemination of knowledge in the area of Medical Physics, the simulator presents potential for application in academic and clinical educational scenarios, serving as a visual tool to illustrate the principles of dose deposition to patients in a comprehensible manner.

The user interface was structured around three fundamental input parameters: the type of incident particle (protons, alpha particles, or photons), the initial energy of the beam (limited to values corresponding to the reality of treatments), and the target biological tissue (skin, adipose tissue, cortical bone, and lung). The manipulation of these variables generates, in real-time, specific relative dose curves, providing various scenarios of radiation-matter interaction that facilitate learning through experimentation.

In parallel with the plotting of graphs, the platform displays a visual representation of the patient's anatomy and the irradiation equipment. A didactic visual simplification was adopted: the use of a standard Linear Accelerator (LINAC) model to illustrate the emission of all beams, aiming to maintain the usability of the interface, even though real hadrontherapy facilities differ in size.

The analysis of the graphical results generated by the simulator exhibits behavior consistent with physical expectations. For proton and alpha beams, the formation of the Bragg Peak is observed. As predicted by theory, it is verified that increasing the initial energy shifts the peak to greater depths, while raising the tissue density pulls it back. In simulations with photon beams, the exponential drop caused by the continuous attenuation through the medium is perfectly visualized, representing the continuous attenuation of the beam in the medium without explicitly reproducing the *build-up* effect observed in clinical beams.

The dose distribution visualized on the patient's silhouette reflects the differences between the curves. When simulating protons or helium nuclei, the interface demonstrates focused dose deposition on the target, with less irradiation in anterior tissues and reduced or zero deposition in posterior tissues. In contrast, the photon simulation exhibits a more continuous dispersion pattern along the path, reflecting dose deposition beyond the tumor volume. This graphical visualization supports the conceptual understanding of the theoretical advantages of dose conformation in each modality.

The application features an information panel summarizing the analytical models, physical parameters, and the scientific literature used, providing transparency regarding the dosimetric foundations of the tool. The presence of features for dynamic controls contributes to the rapid development of comparative scenarios.

6. Conclusion

It is concluded that the developed simulator demonstrates significant potential in achieving its objective of acting as an interactive didactic tool that illustrates in an accessible way the physical principles of radiotherapy and the concepts of dose deposition in Medical Physics. The elaborated interface facilitates

technical communication, assisting the understanding of the relationship between radiation and biological tissues by different user profiles.

It should be noted that the tool has a strictly didactic nature, being built upon simplified physical models and real-time numeric integration, and does not substitute for clinical Treatment Planning Systems (TPS). These commercial systems calculate radiation transport based on advanced algorithms and real tomography scans, whereas the approximations adopted in the project are aimed exclusively at a simplified phenomenological understanding of the physical process.

As a future perspective, it is intended to perform a quantitative validation of the implemented models through comparison with Monte Carlo simulations (e.g., Geant4 or TOPAS). This step will allow for a more precise quantification of the inaccuracies inherent to the approximations adopted in the project—particularly in the purely exponential photon model acting as an illustrator—and will expand the tool's application potential in teaching and research activities. Furthermore, it is suggested to implement the behavior of the electron beam, in order to broaden the repertoire of simulated oncological treatment modalities and refine the educational models.

By integrating concepts of physics, mathematical modeling, and software development, it is expected that the project will stimulate interest in the area of technological applications in health, thereby contributing to the demystification of the use of radiation in medical treatments.

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