A simulation-based assessment of proton beam therapy for oesophageal cancer

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Abstract. Proton beam therapy is a relatively new and alternative modality for cancer treatment. Its major advantage is the highly localised energy deposition of protons at a specific depth, enabling controlled damage to malignant tissues while sparing surrounding healthy cells. However, the successful application of this technique to a specific cancer site needs accurate modelling of several qualitative and quantitative parameters. These parameters are related to both the beam and the target anatomy. In this work, we investigate the suitability of proton beam therapy for the treatment of oesophageal cancer. We use a simulation-based workflow implemented with SRIM. The SRIM code and its TRIM module, enables detailed evaluation of proton energy loss profiles, penetration ranges, stopping powers, and Bragg peak localisation in multilayer tissue equivalent targets, which provides critical insight into energy deposition characteristics relevant for therapeutic optimisation.

1.1 Introduction

Oesophageal cancer is one of the major global health issues and continues to be among the leading causes of cancer related mortality. Over the last few decades, the number of cases has been increasing worldwide. considerable regional and histological variations that have a direct impact on healthcare and disease epidemiology [1, 2]. In general, there are two main subtypes of oesophageal cancer: oesophageal squamous cell carcinoma (OSCC) and oesophageal adenocarcinoma (OAC). These subtypes differ in their biological behaviours, epidemiological distribution, and aetiological factors [2].
Although radiotherapy plays a central role in the management of locally advanced and inoperable oesophageal cancer, traditional photon-based techniques often result in substantial irradiation of surrounding healthy tissues, including the heart, spinal cord and lungs. Due to their associated risks, there is a requirement for new techniques to mitigate the drawbacks of these methods. Proton beam therapy (PBT) is an effective external beam radiotherapy technique that utilizes the unique interaction properties of charged particles in matter. Protons

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gradually lose kinetic energy along their path as they interact with biological tissues. The energy deposition is primarily through inelastic Coulomb interactions with atomic electrons and elastic interactions with atomic nuclei [1, 2]. Due to their comparatively large rest mass, there occurs minimal lateral scattering for protons and the proton trajectory maintains an approximately straight path. As the proton slows down, the interaction probability increases, and as a result energy deposition becomes maximum near the end of its range, known as the Bragg peak [1]. This sharp rise in energy deposition enables precise tumour targeting while significantly reducing irradiation of adjacent normal tissues.

Proper implementation of proton beam therapy requires a detailed understanding of proton–tissue interactions, such as penetration depth, energy loss, and the impact of tissue thickness and composition. Monte Carlo simulation technique provides a robust and quantitative framework for simulating these interactions and for optimising proton beam energy for clinical applications [3, 4]. In the present study, we have used the Monte Carlo code SRIM-2013 (Stopping and Range of Ions in Matter) [5] to analyse proton energy loss and transport through tissue layers relevant to oesophageal cancer. This simulation-based assessment aims to determine whether proton beam therapy is appropriate for oesophageal cancer and to facilitate the optimization of proton beam parameters for better treatment. In this context, the present study aims to offer a physics driven feasibility analysis of proton beam therapy for oesophageal cancer using a multilayer tissue model. The novelty of this study lies in: (i) the development of a multilayer oesophageal model incorporating realistic layer thicknesses and elemental compositions, (ii) a systematic optimisation of proton beam energy guided by depth dependent energy loss and the positioning of the Bragg peak, and (iii) a comprehensive layer-by-layer evaluation of both differential and cumulative energy deposition. Overall, this work provides a simplified yet informative link between fundamental ion–matter interaction studies and future efforts toward patient-specific treatment planning.

2. Methods

2.1 Monte Carlo Simulation Framework

For our purpose, we have utilized the SRIM software. SRIM, a powerful software collection, facilitates the study of ion transport phenomena in various materials. Key applications include the calculation of ion stopping and range in various targets, determining stopping powers and modelling range and straggling distributions across wide range of ions and energies. Furthermore, it supports applications like ion implantation, sputtering, ion transmission and ion beam therapy simulation. In case of target having multilayer composition TRIM (integrated part of SRIM) is used, which provides several outputs such as ion ranges, backscattered/transmitted ions, etc. We can easily calculate the energy loss of various layers such as energy loss due to ionization and energy loss due to recoils and find out position of Bragg peak and lateral distribution of protons in the target. For this scenario, the quick calculation damage option was chosen as it computes the final distribution of ions within the target, energy transferred to recoil atoms, ionization energy loss by the ion, and the number of backscattered and transmitted ions, simulating the interaction between the beam and a target. The simulation of the interaction of the beam with any target, i.e., the energy loss profile of the beam, requires specific quantitative and qualitative information about the target and beam. The following subsection discusses in detail the requirements for the simulation inputs. Although SRIM/TRIM offers a reliable and computationally efficient approach for simulating ion transport and stopping power, it is not tailored for clinical dosimetry applications. In particular, its framework does not comprehensively model complex nuclear reactions, the generation of secondary particles, or the detailed dose calculations required in medical physics. For high precision radiotherapy studies, more sophisticated Monte Carlo platforms such as Geant4, and FLUKA can be employed.

Incorporating these advanced tools will be an important step in future work to achieve greater dosimetric accuracy.

2.2 Anatomical Model

The oesophagus is a muscular tubular organ extending from the pharynx to the stomach and is situated within the mediastinum. It has no protection from a rigid bony framework and lies in close proximity to several critical structures, including the lungs, heart, and spinal cord. The oesophageal wall is composed of different layers such as mucosa, submucosa, muscularis propria, and adventitia. Carcinomas originating in these layers may locally invade adjacent mediastinal tissues and metastasise to regional lymph nodes as well as to distant organs such as the liver, lungs and brain. Therefore, it is essential to choose the incident proton beam energy carefully to both minimize radiation exposure to surrounding organs at risk and ensure the best possible tumour management.
Accurate Monte Carlo-based simulations using SRIM require detailed characterisation of the target materials. This involves atomic number, atomic weight and stoichiometric ratios of constituent elements. In addition, the elemental composition, mass density, and thickness of each tissue layer are essential parameters to calculate reliable stopping power and energy-loss. In the context of proton beam therapy (PBT), these target properties must be specific for the cancer-affected organ and the beam propagation path.
The physical properties and the elemental composition of individual tissue layers are summarised in Table 1 and Table 2 respectively. Light elements such as hydrogen, carbon, nitrogen, and oxygen dominate the composition of human tissues and play a decisive role in determining proton stopping power, penetration depth, and energy deposition within the oesophageal region.
The anatomical representation used in this study relies on a simplified multilayer slab model with consistent thickness and material properties. Although this approach allows for a controlled analysis of proton transport across distinct tissue layers, it falls short of reflecting the intricate anatomical structure and individual variability present in the oesophageal region. In real clinical settings, treatment planning depends on CT derived geometries to achieve higher accuracy. Extending the model to incorporate such patient-specific anatomical details is therefore a logical direction for future work.

Table 1. Properties of layers through which the proton beam propagates to reach the oesophageal cancer site [6, 7, 8].

Layer name	Layer width (cm)	Layer density (g/cm3)
Skin	0.14	1.09
Adipose tissue	0.60	0.95
Skeletal muscles	0.20	1.05
Oesophagus	0.80	1.03

Table 2. Composition (mass fraction in %) of various layers [5, 8, 9].

Layer	H	C	N	O	Na	P	S	Cl	K
Skin	10.0	19.9	4.2	65.0	0.2	0.1	0.2	0.3	0.1
Adipose tissue	11.4	58.8	0.80	28.7	0.10		0.10	0.12	
Skeletal muscles	10.2	14.2	3.4	71.1	0.1	0.2	0.3	0.1	0.4
Oesophagus	10.4	21.3	2.9	64.4	0.1	0.2	0.3	0.2	0.2

2.3 Proton Beam Parameters

Accurate simulation of proton propagation requires specification of the fundamental properties of the particle, including mass, atomic number, atomic weight, and incident energy. In addition, the number of incident protons can be adjusted to achieve the desired statistical precision in the simulations. In proton beam therapy (PBT), protons in the mega–electron-volt (MeV) energy range are employed; however, the optimal beam energy depends on the anatomical location, depth, and material composition of the cancer-affected organ. To obtain an initial estimate of the required proton beam energy for oesophageal irradiation, a water phantom layer was first considered with a cumulative thickness equivalent to that of the tissue layers extending from the skin to the oesophagus. The use of a water phantom for preliminary energy selection is justified by the high-water content of the human body (approximately 60–70%) and the close similarity between the mass density and stopping characteristics of water and soft tissues. Consequently, water serves as an effective tissue-equivalent material for range estimation. Proton ranges were then evaluated for a series of incident energies, as illustrated in Fig. 1. From this analysis, it is observed that proton energies in the range of 32–43 MeV yield penetration depths sufficient for the beam to reach the oesophageal region when propagating through a water-equivalent medium. The energy window identified from the water phantom study is subsequently employed as the input for detailed multilayer tissue simulations using TRIM, where the energy loss and depth–dose characteristics of the proton beam are analysed across individual anatomical layers from the skin to the oesophagus.

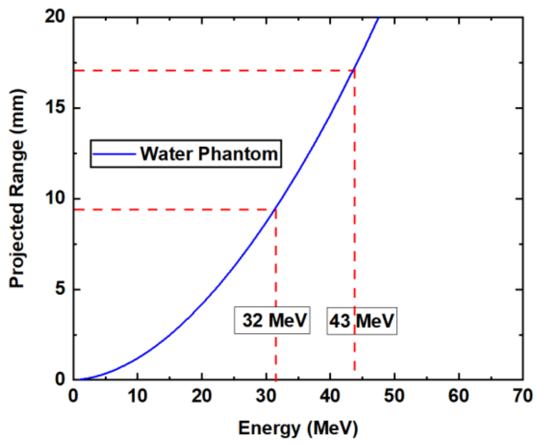

Figure 1. Estimation of required incident proton beam energy, considering water phantom layer.

3. Result And Discussion

In this section, we present and analyse the simulation results, with particular emphasis on proton energy loss, range, and beam spreading characteristics along the beam path. To reach oesophageal cancer cells, the proton beam traverses several tissue layers of approximately 0.9 cm.

3.1 Depth dependent energy loss and cumulative energy deposition

The energy loss of the proton beam through different layers has been simulated with TRIM considering different initial beam energy. The range of the beam energy has been judiciously chosen as per the phantom layer estimation (Fig. 1) as discussed in Sec. 2.3. In Fig. 2 the depth dependent differential energy loss (dE/dx) of proton beams with incident energies of 30, 32, 34, 36, 38, 40, 42, and 44 MeV are shown, as they traverse successive tissue layers and the oesophagus.

The differential energy loss across the various tissue layers strongly depends on the charge-to-mass ratio of the constituent elements and, consequently, on the elemental composition of each layer. Layers enriched with hydrogen or other low-mass elements therefore exhibit higher energy loss, as evident in Fig. 2. Owing to its high hydrogen content, the adipose tissue layer shows a significantly larger differential energy loss compared with the other layers. Finally, a sharp increase in energy loss is observed near the end of the proton range, corresponding to the formation of the Bragg peak. The position of this peak shifts progressively to greater depths with increasing incident proton energy, illustrating the strong depth–energy correlation characteristic of proton transport in matter.

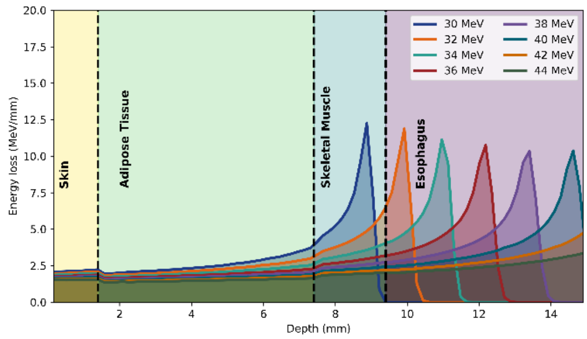

Figure 2 Depth dependent differential energy loss of proton beam having different energies across different layers.

For lower energies (< 32 MeV), the Bragg peak is formed inside or just before the skeletal muscle layer, suggesting that the proton beam does not penetrate sufficiently to reach the oesophageal region. On the other hand, proton energies around 38 MeV place the Bragg peak within the oesophageal layer, enabling maximal energy deposition at the intended target site. For higher energies (~42 MeV), the Bragg peak shifts beyond the oesophagus, suggesting more dose delivery to deeper tissues and a potential rise in unwanted irradiation in healthy organs. These results clearly indicate

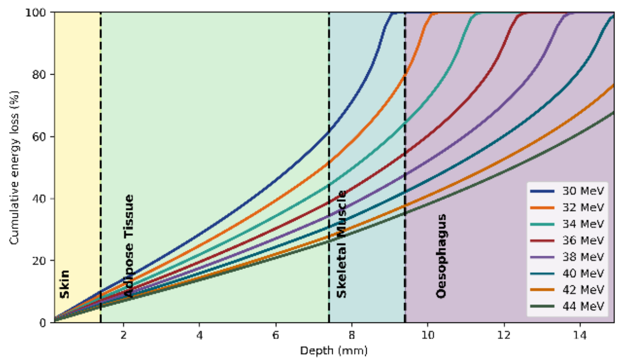

Figure 3 Depth dependent cumulative energy loss of proton beam having different energies across different layers.

that fine energy tuning is essential to confine the high-dose region to the tumour volume while sparing adjacent organs at risk.

Fig. 3 presents the cumulative percentage of energy deposition as a function of depth through the different layers. The monotonic increase reflects the progressive loss of proton kinetic energy through inelastic collisions, with a steeper rise near the Bragg peak region. For beams in the 32–40 MeV range exhibit a pronounced energy deposition within the oesophageal layer, consistent with the Bragg peak positioning observed in Fig. 2. At higher energies, although the cumulative energy deposition continues to rise within the oesophagus, a substantial fraction of the total energy is deposited beyond the target layer.

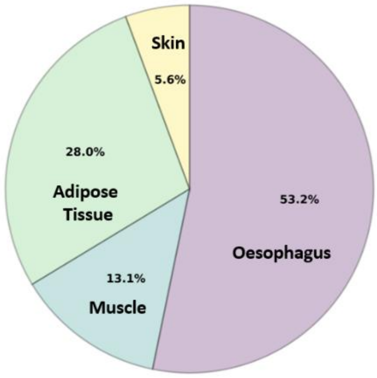

Figure 4 Percentage of total energy loss in each layer for 38MeV incident beam.

From a radiobiological perspective, carefully choosing the beam energy allows the radiation dose to be concentrated sharply in the oesophageal region. This means that more energy is delivered directly to the tumour. As a result, the chances of controlling or destroying the tumour increase as well as less radiation reaches the surrounding healthy tissues. This helps to reduce the risk of damage and complications in normal tissues. Therefore, the combined analysis of differential and cumulative energy loss provides a robust framework for identifying the optimal proton energy window for oesophageal cancer treatment. Overall, these figures shows that proton beams with incident energies around 38 MeV offer the most optimal balance between sufficient penetration and highly localised dose deposition for the anatomical configuration considered in this study. It is important to note that the current study focuses mainly on energy deposition profiles and does not directly consider clinically meaningful dosimetric metrics such as absorbed dose (Gy) or relative biological effectiveness (RBE). To determine these quantities accurately, a voxel-based anatomical representations along with specialised dose-calculation algorithms is required. The analysis of energy loss and Bragg peak behaviour presented here offers valuable first-order insight into the optimisation of proton beams.

3.2 Relative Ionisation Energy Loss in Individual Layers

Fig. 4 displays the layer-wise energy deposition percentage of a 38 MeV proton beam in skin, adipose tissue, skeletal muscle, and the oesophagus. A major fraction of the total energy (~53%) is deposited within the oesophageal layer. This indicates that the Bragg peak for this incident energy lies inside the target region. Adipose tissue is responsible for approximately 28% of the total energy loss, reflecting its relatively hydrogen-rich composition compared to other tissues. Skeletal muscle contributes about 13.1% of the energy deposition, while only a small fraction (~6%) is deposited in the skin layer. The significant localisation of energy deposition in the oesophagus is a key advantage for proton beam therapy, as it enhances dose conformity to the tumour volume.

3.3 Beam Broadening in Different Layers

Fig. 5 shows the depth-wise broadening of proton trajectories along the *Y*-axis (upper panel), *Z*-axis (middle panel), and in the transverse plane (lower panel) as the beam penetrates through successive tissue layers and reaches the oesophagus. Each line corresponds to an individual proton track. The vertical axis shows lateral displacement, while the horizontal axis shows penetration depth. The vertical lines represent the interfaces between multiple anatomical layers.

In the superficial skin layer, the proton trajectories remain well collimated and show very little lateral spread. As the beam enters the adipose tissue layer, the angular dispersion gradually increases. This reflects the increasing multiple Coulomb scattering arising from repeated interactions with electrons and atomic nuclei. This effect becomes more pronounced within the skeletal muscle layer, where the accumulation of small-angle deflections leads to a broader spatial distribution of proton pathways.

Within the oesophageal region, the proton beam exhibits a further increase in lateral spread. This trend is expected, bea the gradual loss of proton momentum near the Bragg peak enhances interaction cross-sections and, consequently, multiple scattering. Nevertheless, the overall beam broadening in this region remains relatively small, of the order of ~1–1.5 mm, indicating good lateral confinement of the proton beam.

Although direct comparison with experimental or clinical data is beyond the scope of the present work, this agreement supports the physical validity of the results. Future studies will include cross validation using other Monte Carlo frameworks and available experimental datasets.

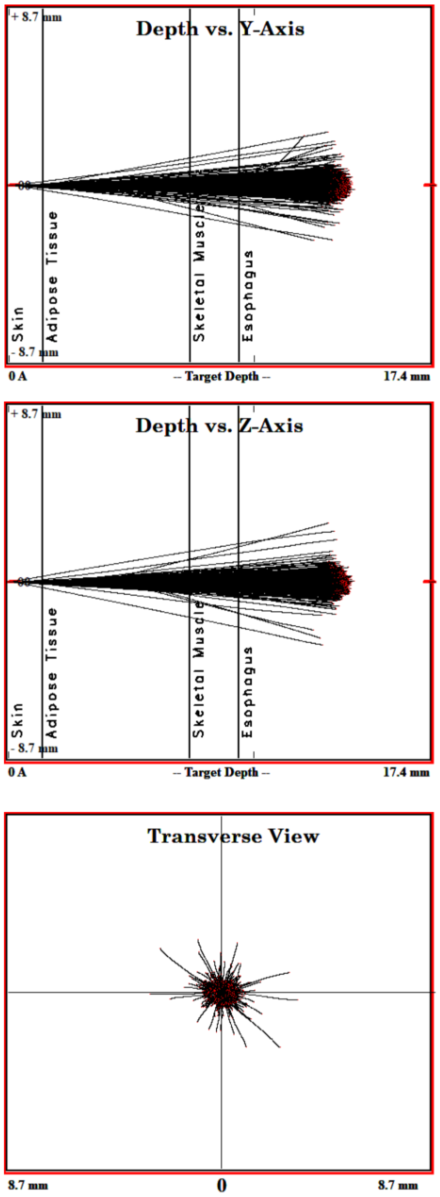

Figure 5 Beam broadening of 38 MeV proton beam across different layers. The top most image represents lateral confinement along Y axis, the middle one along Z axis and the last one is along X axis, i.e., the transverse view.

4. Summary and Future Endeavour

Overall, in this study we have provided a macroscopic demonstration of applicability of proton beam therapy to treat oesophagus cancer. The SRIM-TRIM simulation confirms that depending on the tumour's position in the oesophagus, a proton beam with energies in the range of 32–40 MeV can provide highly localised energy deposition while avoiding large dose delivery to surrounding healthy tissues. Moreover, proton beams within this energy window exhibit good lateral confinement, providing a narrow beam to target specific tumour sites. Based on this macroscopic assessment, our future work will focus on microscopic aspects, including detailed interaction mechanisms and dosimetric characteristics. Future work will also focus on extending the present model to patient-specific CT-based geometries to improve clinical applicability and enable more realistic dose estimation. Additionally, for the sake of model simplicity, this study does not incorporate physiological effects such as respiratory motion, organ deformation, and setup uncertainties, all of which can have a significant impact on proton range and dose distribution in thoracic cancer treatments.

Acknowledgements

One of the authors (A.G.) acknowledges the IEDC Laboratory, Department of Basic Science and Humanities, IEM Kolkata (Newtown Sector), for providing computational facilities. The authors extend their gratitude to Dr. Santanu Nandi, Associate Professor at Calcutta National Medical College and Hospital, for providing the essential information.

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