Applied Radiation and Isotopes 142 (2018) 173–180

Contents lists available at ScienceDirect

Applied Radiation and Isotopes

journal homepage: www.elsevier.com/locate/apradiso

Accuracy of using high-energy prompt gamma to verify proton beam range with a Compton camera: A Monte Carlo simulation study

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HIGHLIGHTS

- The accurate estimation of the BP position can be achieved using high-energy PG rays.
- The proposed position estimation method can provide a mean accuracy of < 2 mm.
- Irradiated tissue and event selection should be carefully taken into account.

ARTICLE INFO

Keywords:
Proton radiotherapy
Compton camera
Prompt gamma imaging

ABSTRACT

Prompt gamma (PG) rays emitted during proton therapy has been used for proton range verification. Because high-energy PG emission is well correlated to the Bragg peak (BP), high-energy PG rays are well-suited for proton range verification. However, the low production and detection of high-energy PG rays often lead to inaccurate BP position estimates. The aim of this study is to improve the BP position estimates obtained from high-energy PG rays. We propose a BP position estimation method based on the local maximum closest to the distal fall-off region. We present the results of Monte Carlo simulations in which a water phantom was irradiated with a proton beam. Our results show that the BP position estimated from the 6.13 MeV PG rays can be improved using the proposed position estimation method. Moreover, the 6.92 and 7.12 MeV PG rays can be used for predicting the BP position. However, the accuracy of the BP position estimation decreases with decreasing tissue oxygen levels. We also found that the subtraction of the PG images of 6.13 MeV from those of 6.92 and 7.12 MeV can be used to predict the BP position with a mean accuracy of < 2 mm. The accurate estimation of the BP position can be achieved using different high-energy PG rays, but factors including position estimation, irradiated tissue and event selection should be carefully taken into account.

1. Introduction

Due to the unique physical properties of protons, proton therapy can deliver higher doses to the tumor while reducing doses to the surrounding normal tissues. However, the unique properties make proton therapy sensitive to internal organ motion, patient positioning errors and anatomic changes (Knopf and Lomax, 2013). A reliable method for in-vivo dose and range verification is required. During proton therapy, a large number of gamma rays are produced by proton-induced nuclear reactions with the patient's tissue. These rapidly emitted gamma rays are known as prompt gamma (PG) rays. Several studies showed that there is a strong correlation between the distributions of the proton dose and the PGs (Min et al., 2006; Testa et al., 2010; Zarifi et al., 2017). This indicates that PG imaging can be a useful tool for both proton dose and range verification. Like other imaging modalities, PG imaging is a non-invasive imaging technique that has recently been shown to be a promising tool for proton range verification (Krimmer et al., 2018). In addition, PG rays are emitted almost instantaneously from the decay of the excited nuclei. As a result, PG imaging has no biological washout effect (Moteabbed et al., 2011) and is potentially suitable for real-time tracking of the BP during beam delivery.

Over the last ten years, PG imaging has undergone a rapid development which is still continuing. In addition to continuous improvements in hardware and software, several different imaging

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https://doi.org/10.1016/j.apradiso.2018.10.003
Received 21 June 2018; Received in revised form 30 August 2018; Accepted 6 October 2018
Available online 10 October 2018
0969-8043/

methodologies were proposed (Krimmer et al., 2018). Two different collimated cameras, a multi-parallel-slit camera and a knife-edge slit camera, were designed to produce a one-dimensional (1D) projection of the beam path (Krimmer et al., 2015; Roellinghoff et al., 2014; Smeets et al., 2012). Compton cameras that consist of two parallel detectors were used to produce three-dimensional (3D) distribution of the PG rays (Singh, 1983; Todd et al., 1974). Furthermore, new proton range verification techniques, including PG spectroscopy (Verburg and Seco, 2014), PG timing measurements (Golnik et al., 2014), and the PG peak integrals (Krimmer et al., 2017), were proposed. Although there is currently no commercial PG imaging modality available, some PG imaging prototypes are being developed and tested under clinical conditions (Polf et al., 2015; Richter et al., 2016).

Several recent studies reported that high-energy PG rays are well correlated to the Bragg peak (BP) position of the proton beam (Hilaire et al., 2016; Zarifi et al., 2017). In particular, the 6.13 MeV PG emission line which originates from the de-excitation of $^{16}O^{*}$ nuclei exhibits a distinctively close correlation with the BP (Zarifi et al., 2017). Their simulation results showed that the 6.13 MeV PG rays generated from the 100–200 MeV proton pencil-beam irradiation of a homogeneous (or heterogeneous) phantom could accurately predict the BP position and provide an accuracy to 1 mm under ideal conditions (Zarifi et al., 2017). Recently, our simulation results showed that the reconstruction of the 6.13 MeV PG rays obtained by a two-stage Compton camera yielded poor image quality and induced large errors in BP position estimation (~ 17 mm) (Jan et al., 2018). The main reason is that the yield and detection efficiency of high-energy PG rays are extremely low (Jan et al., 2018; Zarifi et al., 2017). In addition, the performance of a Compton camera is limited by the finite energy and spatial resolutions of the detector system, the neutron-induced noise, incomplete absorption of PGs and Doppler broadening (Mackin et al., 2013).

In this study, we aim to improve the BP position estimation using the 6.13 MeV PG image with different position estimation methods. We also investigate the feasibility of using the 6.92 and 7.12 MeV PG emission lines to predict the BP position. These two PG emission lines exhibit a strong correlation with the BP, but have not been studied. We present the results of Monte Carlo simulations in which a water phantom was irradiated with a proton beam. Because the three characteristic PG rays (i.e. 6.13, 6.92 and 7.12 MeV) originate from the de-excitation of $^{16}O^{*}$ nuclei, we conduct Monte Carlo simulations that evaluate the effect of tissue oxygen contents on the accuracy of the BP position estimation. Finally, we investigate whether the accuracy of the BP position estimation can be improved by combining signals obtained from the characteristic PG emissions of 6.13, 6.92 and 7.12 MeV.

2. Materials and methods

2.1. The principle of Compton camera

As shown in Fig. 1(a), a Compton camera generally consists of two parallel plane detectors, often called scatterer and absorber, working in the coincidence mode (Singh, 1983; Todd et al., 1974). In the scatterer detector, an incident gamma-ray photon undergoing a Compton scattering is deflected through a scattering angle (θ) with respect to its original direction. Then, the scattered photon is absorbed in the absorber detector. According to the Compton formula, the scattering angle of the scattered photon can be determined using the energies deposited in the scatterer and absorber detectors. Using the scattering angle, the possible source location can be restricted to the surface of a cone where the axis of the cone is defined by the two interactions measured in the scatterer and absorber detectors. Finally, the 3D distribution of gamma-ray sources can be reconstructed simply by back-projecting all coincidence events into an imaging volume through cone surfaces.

2.2. Simulations of PG emission, detection and reconstruction

In this work, we used Geant4's (v10.02) (Agostinelli et al., 2003; Allison et al., 2016) QGSP_BIC_HP_EMY physics list to simulate the PG emission. We next used Gate version 7.2 (Jan et al., 2004) to simulate the detection of PG emission. The electromagnetic (EM) standard physics option 3, which models the processes of ionization, bremsstrahlung, multiple scattering, Compton scattering photoelectric effect, pair production and annihilation for EM interactions of photons and charged particles, was selected. The maximum step size and range cuts for gammas, electrons, positrons, and protons were set to 1 mm (Robertson et al., 2011). Fig. 1(b) illustrates the simulation setup. A 120 MeV Gaussian pencil beam (σ = 0.5 cm) was used to irradiate a 16 cm long water cylinder (radius = 8 cm). Each simulation was repeated ten times and performed with 10^9 primary protons (Smeets et al., 2012). To decrease simulation time, we used the accelerated PG estimation proposed by Huisman et al. (2016). The accelerated method generates PG production probabilities which are calculated once and stored in a file. Next, a voxelized image of PG production yield, normalized to a single primary, is computed from the stored file, as a function of the current energy of the primary, the material in each voxel and the step length. Finally, the intermediate PG image is used as a source to generate and propagate the number of PGs throughout the detector system.

As also illustrated in Fig. 1(b), we modeled a two-stage Compton camera consisting of a scatterer made of 10 × 10 × 1.5 cm^3 lutetium-yttrium orthosilicate ($Lu_{1.8}Y_{0.2}SiO_5$:Ce, LYSO) scintillation detectors and an absorber made of 15 × 15 × 1.5 cm^3 LYSO detectors. Our previous stduy showed that compared to the thick absorber, the thin absorber had higher incompletely absorbed PGs, but it had better depth resolution (Jan et al., 2017). In this work, we didn't simulate the internal radioactivity of LYSO (i.e. ^{176}Lu) because it has small effect on the coincidence detection mode (Huber et al., 2002). The scatterer and the absorber were sampled into 50 × 50 and 75 × 75, respectively, discrete square detector elements. Each detector element has the size of 0.2 × 0.2 cm^2. Fig. 1(c) is a flow chart of the entire simulation process. First, we searched coincidence events involving interactions in both the scatterer and the absorber. The searched coincidence events may include incompletely absorbed PGs. Then, the position and deposited energy of each interaction were extracted. In practice, a PG may undergo several interactions in the detector. If multiple interactions occur, the event position was calculated using the centroid of all interaction points, each weighted by the ratio of its individual energy to the total energy deposited in the crystal (i.e. the energy-weighted centroid). In the image reconstruction process, we assumed that there was no depth-of-interaction information. The energy resolution of the LYSO detectors was modeled by adding a random mismeasurement ($\Delta E = (2\sqrt{2\ln 2})^{-1} \times 0.056 \times E + 0.038\sqrt{E} \times N[0,1]$) to the total energy (E) deposited in the detector (Hueso-González et al., 2016). N[0, 1] is a sample of random numbers from a normal distribution with a mean of 0 and a standard deviation (SD) of 1.

In this study, coincidence events having total energy depositions in the 5.93–6.33 MeV energy window (i.e. 6.13 ± 0.2 MeV) were used to reconstruct PG images. It should be noted that the 6.92 and 7.12 MeV PG rays produced from an excited oxygen nucleus (Kozlovsky et al., 2002) also exhibit a close correspondence with the BP. As a result, the 6.92 and 7.12 MeV PG rays may have the potential to verify proton beam range. However, the yield of the 6.92 and 7.12 MeV PG rays is relatively low. We thus evaluated the PG image reconstructed from coincidence events having total energy depositions in the 6.72–7.32 MeV energy window (i.e. sum of 6.92 ± 0.2 MeV and 7.12 ± 0.2 MeV PG rays). All PG images were reconstructed using the proposed list-mode ordered subset expectation maximization algorithm with a shift-variant point spread function model (8 subset and 10 iterations) (Jan et al., 2018). More details about image reconstruction can be found in our recent study (Jan et al., 2018).