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First demonstration of real-time gamma imaging by using a handheld Compton camera for particle therapy

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ABSTRACT

The use of real-time gamma imaging for cancer treatment in particle therapy is expected to improve the accuracy of the treatment beam delivery. In this study, we demonstrated the imaging of gamma rays generated by the nuclear interactions during proton irradiation, using a handheld Compton camera (14 cm × 15 cm × 16 cm, 2.5 kg) based on scintillation detectors. The angular resolution of this Compton camera is ~8° at full width at half maximum (FWHM) for a ^{137}Cs source. We measured the energy spectra of the gamma rays using a $LaBr_3(Ce)$ scintillator and photomultiplier tube, and using the handheld Compton camera, performed image reconstruction when using a 70 MeV proton beam to irradiate a water, $Ca(OH)_2$, and polymethyl methacrylate (PMMA) phantom. In the energy spectra of all three phantoms, we found an obvious peak at 511 keV, which was derived from annihilation gamma rays, and in the energy spectrum of the PMMA phantom, we found another peak at 718 keV, which contains some of the prompt gamma rays produced from ^{10}B. Therefore, we evaluated the peak positions of the projection from the reconstructed images of the PMMA phantom. The differences between the peak positions and the Bragg peak position calculated using simulation are 7 mm ± 2 mm and 3 mm ± 8 mm, respectively. Although we could quickly acquire online gamma imaging of both of the energy ranges during proton irradiation, we cannot arrive at a clear conclusion that prompt gamma rays sufficiently trace the Bragg peak from these results because of the uncertainty given by the spatial resolution of the Compton camera. We will develop a high-resolution Compton camera in the near future for further study.

1. Introduction

Particle therapy is a type of cancer therapy, which utilizes the dosage based on the distribution of energetic charged particles, called the Bragg peak. This Bragg peak enables the particle beams to damage tumors efficiently. During this process, it is important to ensure that the energy deposits from the charged particles are only on the tumors and that the surrounding normal tissues are unaffected. One common methods for verifying this is the imaging of the gamma rays emitted by nuclear interactions during the therapy. In order to trigger nuclear reactions, charged particles need to have sufficient energy to overcome the Coulomb barrier of the nucleus, defined as the threshold energy. A natural consequence of this is that peak of the gamma ray emission may not coincide with the position of the Bragg peak. In general, the Bragg peak results in maximum ionization (dE/dx) energy deposit, occurring at depths where protons have lost most of their initial kinetic energy. In contrast, nuclear reactions that yield excited nuclides and generate high-energy gamma radiation occur at shallow depths, where protons retain most of their energy. In fact, the peak position of 511 keV gamma-ray distribution differs typically about several mm to more than 10 mm, depending on the initial energy of charged particle, from the Bragg peak position [1]. Nevertheless, current proton therapy generally uses 511 keV gamma-ray imaging in order to verify the range before, during, and after proton therapy [2].

There are two types of gamma rays emitted during charged particle irradiation, namely annihilation gamma rays and prompt gamma rays. Positron emission tomography (PET) is the most common medical imaging method of detecting tumors. A PET detector has a full-ringed structure in order to detect the annihilation gamma rays produced from biologically active compounds

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Table 1
The half-lives of the main positron emitters.

Positron emitter	Half-lives
^{11}C	20.39 min
^{15}O	2.04 min
^{13}N	9.97 min
^{10}C	19.3 s

labeled with a positron emitting isotope in the human body [3]. During the particle therapy, positron emitters such as ^{11}C, ^{15}O, ^{13}N, and ^{10}C are generated by the interactions between the incident particles and the nuclei of the tissues. Table 1 shows the half-lives of the main positron emitters produced during the therapy. Since PET scanners can also detect the annihilation gamma rays created in this process, PET imaging is currently the most popular method of verification. However, a PET scanner cannot be installed in the particle therapy room due to its full-ringed detector structure, which interferes with charged particle irradiation. Thus PET imaging can only acquire delayed (offline) images [4] and cannot obtain all contributions from the distribution of positron emitters, as the patients need to move from particle therapy room to the PET scanning room after the therapy (approximately 10~15 min). In order to acquire online PET images, an alternative PET detector structure is necessary: an example would be a structure with two opposite detectors [1,5], or a ring structure with a small opening for the incoming particle irradiation beam [6].

Since prompt gamma rays are emitted concurrent with the nuclear interactions, or are emitted from the de-excitation of the radioactive nuclides produced by the nuclear interactions, which have short half-lives, we can retrieve real time information on the nuclear interactions. Moreover, prompt gamma rays are single-emission photons; this makes them detectable by gamma cameras, and their distribution, *particularly gamma rays with an energy of approximately* 4 MeV, *traces the Bragg peak well owing to their low threshold energy of nuclear interactions, such as an inelastic scattering between a proton and* ^{12}C, as shown in previous reports [7,8]. Therefore, the acquisition of online images using prompt gamma rays is expected to be useful for range verification in particle therapy. A Compton camera can be used as a detector for online monitoring in particle therapy because it is capable of conducting measurements across a wide energy range, and can measure not only the annihilation gamma rays (511 keV) but also a part of the prompt gamma rays.

Although we should try to acquire the high energy gamma-ray image such as around 4 MeV, we first demonstrate whether we can acquire the online gamma-ray images which have sufficient statistics for image reconstruction (see Section 3.1). In this paper, we report the experimental results of the measurement of the energy spectra of gamma rays during proton irradiation and of the online gamma imaging using our Compton camera.

2. Materials and methods

2.1. Principle of a Compton camera

A Compton camera is an imaging device which can constrain the incoming direction of gamma rays based on the Compton kinematics, by measuring the energy deposits and interaction positions of the scatterer and the absorber (Fig. 1). The angular resolution measure (ARM) is the difference between the angle θ_e calculated from the Compton scattering equation (Eq. (1)), and the angle θ_g calculated from the interaction positions and source

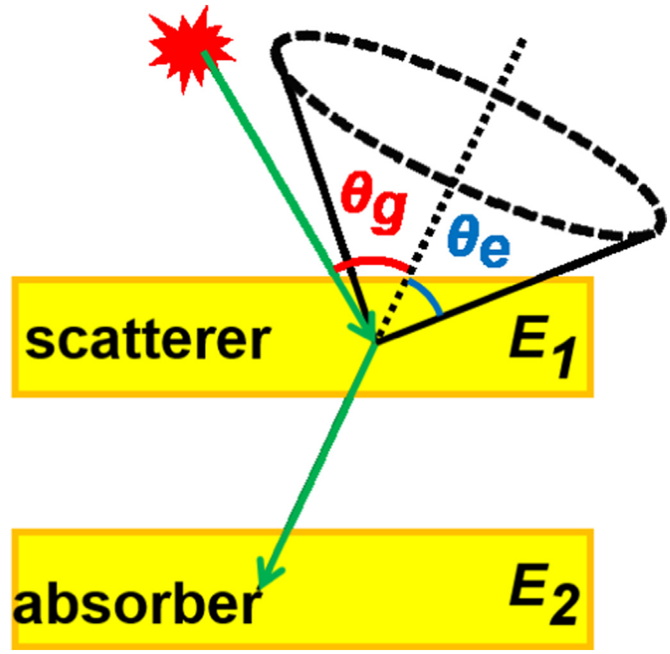

Fig. 1. The principle of operation of a Compton camera.

direction. The ARM is an indicator of the spatial resolution. To get good ARM, we need to improve the position and energy resolution of the detectors:

$$\cos\theta_e = 1 - \frac{m_e c^2}{E_2} + \frac{m_e c^2}{E_1 + E_2} \tag{1}$$

$$\mathrm{ARM} \equiv \theta_e - \theta_g \tag{2}$$

2.2. Our Compton camera

Fig. 2 shows the configuration of our Compton camera. It was developed for the measurement of experimental radiation levels and is based on scintillation detectors which consist of multi-pixel photon counters (MPPC) and Ce:GAGG scintillators [10]. Its ARM measured at full width at half maximum (FWHM) is about 8° at 662 keV [11].

Our Compton camera is configured to attain high efficiency for the imaging of ^{137}Cs. The camera has four scintillator blocks and one block has two scatterers and one absorber, which have an array configuration of 11 pixels × 11 pixels × 1 pixel of 2.0 mm × 2.0 mm × 4.0 mm Ce:GAGG scintillators, and 11 pixels × 11 pixels × 10 pixels of 2.0 mm × 2.0 mm × 2.0 mm Ce:GAGG scintillators, respectively. The distance between the second scatterer and absorber is 12.5 mm [12].

2.3. Experimental setup

We performed experiments using proton beams at the National Institute of Radiological Sciences, in Japan. Fig. 3 shows the experimental setup and a schematic diagram of the setup. The proton energy and beam intensity were set to 70 MeV and 3 pA, respectively. The irradiation targets were water, calcium hydroxide, and polymethyl methacrylate (PMMA) phantom, respectively. The size of the water and calcium hydroxide phantoms were approximately 10 cm × 10 cm × 30 cm, while the size of the PMMA phantom was 3 cm × 3 cm × 10 cm. The irradiation time was set at 10 min for the water and calcium hydroxide phantoms and 20 min for the PMMA phantom in order to obtain sufficient statistics with image reconstruction (see 3.2). We measured the energy spectra