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Construction and performance of a dose-verification scintillation-fiber detector for proton therapy

Suhyun Lee, Byungsik Hong, Kyong Sei Lee*, Benard Mulilo, Sung Keun Park

Korea Detector Laboratory and Department of Physics Korea University, Seoul 136-713, Republic of Korea

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ABSTRACT

A multilayer scintillation-fiber detector has been developed for precision measurement of time-dependent dose verification in proton therapy. In order to achieve the time and position sensitivity required for the precision dose measurements, a prototype detector was constructed with double-clad 1-mm-thick scintillation fibers and 128-channel silicon photodiodes. The hole charges induced in each channel of the silicon photodiodes were amplified and processed with a charge-integration mode. The detector was tested with 45-MeV proton beams provided by the MC50 cyclotron at the Korea Institute of Radiological and Medical Science (KIRAMS). The detector response for a 45-MeV proton beam was agreed fairly well with the predicted by GEANT4 simulations. Furthermore, the quantitative accuracy appearing in the spatial distribution of the detector response measured for 20 s is in the order of 1%, whose accuracy is satisfactory to verify beam-induced dose in proton therapy. We anticipate that the detector composed of scintillation fibers and operating in the charge-integration mode allows us to perform quality measurement of dynamic therapeutic beams.

1. Introduction

Protons have a great advantage over X-rays in cancer therapy in virtue of well-controlled delivery of the dose to the planned target volume (PTV) [1,2]. The well-defined depth of interaction and the low lateral dispersion of the protons minimize unnecessary delivery of dose to normal tissue or sensitive organs surrounding the cancer tumor. As the technology of particle therapy becomes more sophisticated, the dose measurements and the simulations [3] for predictions and confirmations are required to be more accurate and swifter.

In carbon therapy [4,5], parallel-plate ionization chambers (PPIC) in an ionization-chamber-mode operation are typically used for precision measurement of dose responses [6]. Dose distributions in the orthogonal plane to the beam direction were measured as functions of the depth of interaction of the beam in a water phantom.

In spite of the well-established technology, a major drawback in the use of thick PPIC chambers is the complicated calibrations required to compensate for the difference in densities between the detector material and the actual human tissue, especially when a multilayer detector is required to measure the three-dimensional dose response over the expected range of the therapeutic beam in the PTV.

It is hypothesized that simultaneous measurement of the doses over a wide range allows us to reduce the scan time and labor required for the dose verification process. Furthermore, tracking the movement of pencil- [7] or spot-scan beams [8] in each depth of the PTV can be achieved by a data acquisition speed of a few hundred Hz.

In virtue of the uniform response to radiation and the high detector granularity, scintillation fibers have been widely applied for the measurement of doses in proton therapy [9], for the measurement of thermal-neutron fluxes in reactor fuel rods [10], and for the visualization of transmitted X-rays [11].

In the present research, we developed a dose verification detector using polystyrene-based scintillation fibers because of the following reasons:

(1) The thin and uniform thickness of the scintillation fibers is fairly advantageous to establish a multi-layered structure of the detector.
(2) The density of the typical polystyrene-based scintillation fiber is 1.032 g cm^{-3}. Therefore, the mean density of the detector can be closely adjusted to be water- (or tissue [12]) equivalent ($\rho = 1.00$ g cm^{-3}) within an accuracy less than 1% when the detector is constructed with active layers and absorbers.

In the previous researches [13], the feasibility of a detector based on scintillation fibers and silicon photodiodes operating in a charge-integration-mode signal process was confirmed and the imperative points to be noted are:

* Corresponding author. Tel.: +82 2 3290 4277
E-mail addresses: Kyong.Sei.Lee@cern.ch, kslee0421@korea.ac.kr (K.S. Lee).

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(1) The lateral spread of a 45-MeV proton beam was measured with a position resolution of 1 mm.
(2) The channel variation of the dose distributions measured at different depths of the beam is about 2% (standard deviation).

The number of the scintillation photons produced in a typical 1-mm thick scintillation fiber by a proton at its Bragg peak region is about 5000. In consideration of attenuation loss of the scintillation light and inefficiency in photocathodes of silicon photodiodes, the expected number of the photoelectrons produced by the proton is about 1000. The intensity of the hole current induced by a 1-nA proton beam at the distal edge, ~1 μA, is therefore in an optimal range of the signal process when typical multichannel silicon photodiodes are used for the photon sensors.

In spite of the advantages of high granularity and quantitative accuracy, the scintillation fibers have a drawback of non-uniform scintillation response to the energy loss of heavy charged particles. The reduction in the scintillation response (so called quenching) obviously becomes more severe as the particles reach the Bragg-peak region. An accurate measurement of the scintillation response of the scintillation fibers is, therefore, essential to ensure accuracy of the dose measurement for the proton therapy.

In this paper, the construction of the prototype detector comprising of multilayer scintillation fibers equipped with 128-channel silicon-photodiodes and the details of the design of the signal processing electronics operating in the current-sensitive mode are described in Sections 2 and 3, respectively. The dose-distribution data measured for 45-MeV proton beams provided by the MC50 proton cyclotron at the Korea Institute of Radiological and Medical Science (KIRAMS) are reported in Section 4. The first-order parameter of the Birks model, *kB*, for the scintillation fibers used for the present detector R&D was estimated by comparisons of the scintillation responses to the specific energy losses predicted by GEANT4 simulations. The result is reported in Section 5. Finally, in Section 6, we make conclusions for the current R&D performed with the scintillation-fiber-based detector and for the relevant application to the particle therapy.

2. Detector

The prototype detector constructed for the present R&D was designed for therapeutic proton beams with a maximum energy of 180 MeV, whose expected range in the detector was 20 g cm^{-2}. Fig. 1 shows the prototype detector composed of five orthogonal pairs of scintillation fibers and absorbers made of polymethyl methacrylate (PMMA). The mean density of the detector that a proton beam experiences was closely adjusted to 1.00 g cm^{-3} by adjusting the thicknesses of the absorbers. The variation of the density in the sensitive volume of the detector (standard deviation) was expected to be 5×10^{-3} g cm^{-3}.

As shown in Fig. 1, an orthogonal pair of the fiber layers measures dose–response distributions in the orthogonal plane to the incident beam direction at the given depth. The gap lying between two adjacent detector units, each composed of an orthogonal pair, to be filled with a series of PMMA absorber plates is 40 mm.

The detector responses at the positions lying in the 40-mm gaps between two adjacent detector units can be measured by attaching an additional PMMA absorber to the detector window. By increasing the thickness of the additional absorber from 0 to 3.8 g cm^{-2} with 0.2–g cm^{-2} steps, we reconstruct dose–response data for a 180-MeV therapeutic beam with totally 100 depth steps. The amount of time required to complete a single scan was expected to be 200 s with a 10-s measurement for each step.

Each fiber layer consists of 104 1-mm-thick scintillation fibers (Bicron model BCF-60) with a square cross-section of 1×1 mm^2. The scintillation fibers were double clad to enhance the efficiency of total reflection for the scintillation light. The refraction indices for the core polystyrene and the doubly clad acrylic of the scintillation fibers are 1.60 and 1.49, respectively. The light yield of the BCF-60 fibers is maximum at a wavelength of 530 nm.

The scintillation fibers were carefully mounted on a 1-mm thick black PMMA plate as shown in Fig. 2. In order to maximize the collection of scintillation light, a 100–μm-thick light-reflection film with reflectivity larger than 90% was attached on the PMMA plate with epoxy glue. Then, the scintillation fibers were placed and properly fixed on the PMMA plate by applying the same epoxy at both ends of the scintillation fibers. The surface of the fiber layer on the other side of the PMMA plate was also covered with the same light-reflection film and, finally, with 110–μm-thick black tape for a tight light shield. The size of the active detection area defined by a pair of orthogonal-fiber layers is 104×104 mm^2.

Both ends of the fiber layers of the detector were carefully polished to maximize the efficiency of the transmission of light. The lower end of the fiber layer directly contacts the photocathode array of a 128-channel silicon photodiode board mounted on the signal processing board. An air gap of 0.5 mm between the end of the fiber layer and the photocathode array was required to ensure safety of the sensitive photocathodes from physical damage. The upper end of the fiber layer was also covered with a layer of the

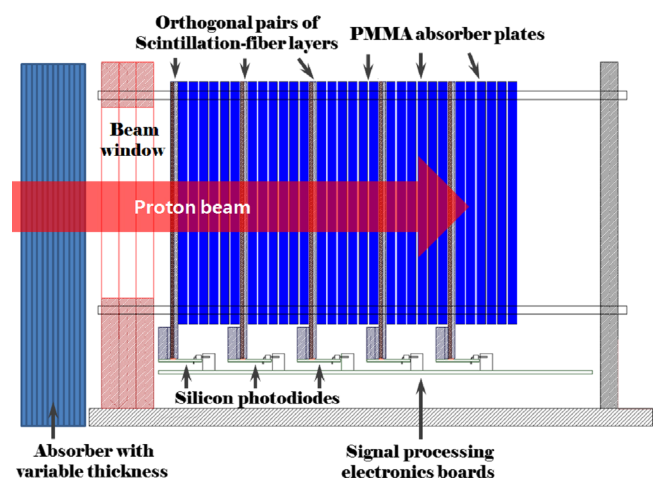

Fig. 1. Schematic diagram of the prototype detector composed of five orthogonal-pairs of scintillation-fiber layers and absorbers made of PMMA.

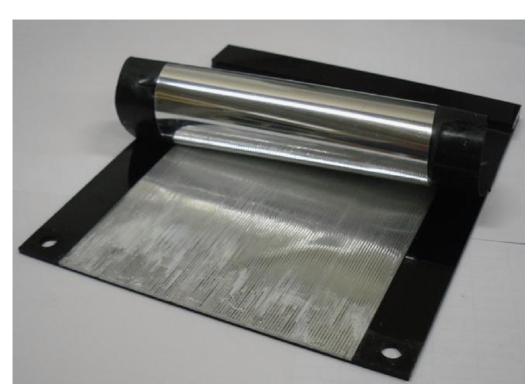

Fig. 2. A single detector layer composed of 104 scintillation fibers mounted on an 1-mm thick PMMA plate.