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# Development of a Compton Camera for Medical Applications based on Silicon Strip and Scintillation Detectors

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## Abstract

A Compton camera is being developed for the purpose of ion-range monitoring during hadrontherapy via the detection of prompt-gamma rays. The system consists of a scintillating fiber beam tagging hodoscope, a stack of double sided silicon strip detectors ( $90 \times 90 \times 2$  mm<sup>3</sup>,  $2 \times 64$  strips) as scatter detectors, as well as bismuth germanate (BGO) scintillation detectors ( $38 \times 35 \times 30$  mm<sup>3</sup>, 100 blocks) as absorbers. The individual components will be described, together with the status of their characterization.

*Keywords:* Compton camera, silicon strip detectors, prompt gamma, hadrontherapy

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## 1. Introduction

Hadrontherapy, i.e. the treatment of tumors via a beam of light ions - mainly proton and carbon, is an emerging technology that takes advantage of the effect, that ions deposit a large fraction of their energy close to the end of their range, in the Bragg peak region, while travelling along almost straight trajectories. In comparison to conventional radiotherapy with X-rays, this property allows a better concentration of the applied dose to the tumor volume whereas surrounding healthy tissues are widely spared. Hence, tumors close to organs at risk, are for instance particularly indicated for this type of treatment.

A critical issue in the quality control of hadrontherapy is the surveillance of the Bragg peak location and its conformation to the tumor volume. A mismatch could lead to an under-dosage in the target volume and an over-dosage in healthy tissues.

Methods for monitoring the ion range are based on the detection of secondary radiation. One modality, which has already proven its clinical applicability, is the registration of the emission point of 511 keV annihilation radiation following a  $\beta^+$ -decay, by using positron emission tomography (PET) [1, 2, 3]. In the case of carbon ion treatment, tracking of emitted light charged fragments can be used for a reconstruction of the primary interaction vertex [4, 5, 6, 7]. Inelastic nuclear reactions of the incident ions lead also to the generation of prompt-gamma rays which are emitted almost instantaneously after the interaction. It has been shown that the production rates of the prompt-gammas (energies up to approximately 10 MeV) are highly correlated to the range of the primary ions [8, 9, 10, 11]. Camera systems for the detection of prompt-gamma rays, based on passive collimation, can either be of the knife-edge type [12, 13] or can include a parallel multi-slit collimator [14, 15, 16]. An alternative approach to passive collimation relies on the Compton camera concept, which has the potential advantage of an expected improved efficiency compared to passive collimation. In the field of nuclear medicine, Compton cameras can replace classical single photon emission computed tomography (SPECT) systems with passive collimation and open the door to new radiotracers with gamma energies on the order of 1 MeV.

Several groups worldwide are studying Compton cameras (see e.g. [17, 18, 19, 20, 21]). The present article is focused on the development of a time-of-flight Compton camera of clinical size.

37 **2. Compton camera**

38 *2.1. Principle*

39 The principle of the Compton camera is shown in Figure 1 [22]. Incident  
40 ions are passing a beam tagging hodoscope, which provides information about  
41 the transverse position and may also serve as a time reference for time-of-  
flight (TOF) measurements. Photons produced via nuclear interactions of

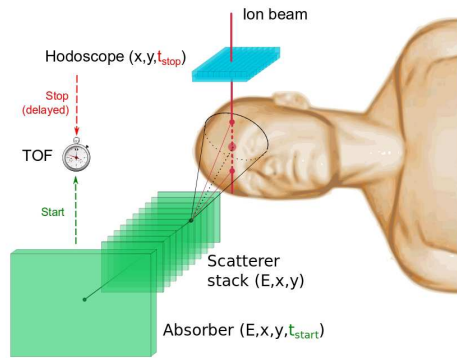


Figure 1: Principle of the Compton camera with its basic components: the beam tagging hodoscope, the scatter detectors and the absorber. The vertex of photon production is reconstructed via the intersection of the Compton cone with the line of incident ions.

42  
43 the incident ions interact first in a stack of low-Z element scatterers before  
44 the scattered photons hit the high-Z element absorber. Using energy- and  
45 position-sensitive detectors, the Compton cone defined by the apex in the  
46 scatterer and the scattering angle is determined. The vertex of the photon  
47 creation is then given by the intersection of this cone with the incident ion  
48 trajectory. One of the two intersection points generally obtained can be  
49 considered as background.

50 *2.2. Components*

51 The basic components of the Compton camera comprising the hodoscope,  
52 the scatter detectors and the absorber are displayed in Figure 2.  
53 The beam tagging hodoscope consists of an array of scintillating fibers (BCF 12,  
54  $1 \times 1 \text{mm}^2$  square section)<sup>1</sup>, which are coupled to multianode photomultipliers  
55 (PMs) (H-8500) via optical fibers<sup>2</sup>. Two prototypes have been built with  
56  $2 \times 32$  and  $2 \times 128$  fibers, respectively. Test measurements have shown that  
57 a timing resolution better than 1 ns full width at half maximum (FWHM)

<sup>1</sup><http://www.crystals.saint-gobain.com/Scintillating-Fiber.aspx>

<sup>2</sup><http://www.foretec.fr/fibre-optique.html>

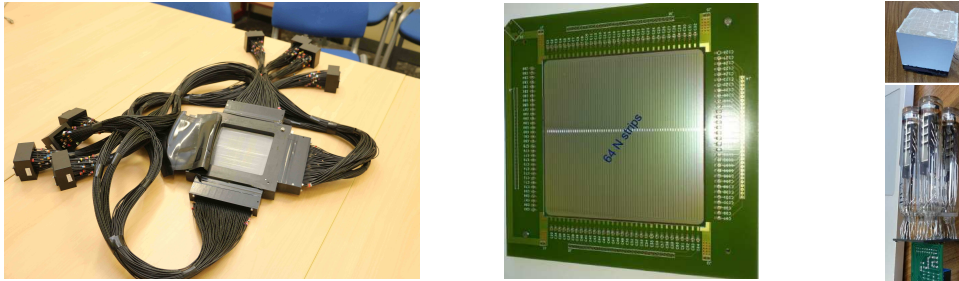


Figure 2: Individual components of the Compton camera with the hodoscope (left), a silicon strip detector (middle) and a BGO detector (right).

58 is possible and that count rates above 10 MHz can be reached. For the full  
 59 size prototype, the scintillating fibers are read from both ends and the signals  
 60 from neighboring fibers are connected to different PMs for an improved count  
 61 rate capability. Dedicated front-end electronics is being developed. The first  
 62 version of an application specific integrated circuit (ASIC) [23] contains a  
 63 current comparator for each channel to provide digital information on the  
 64 fibers that have been hit, as well as the possibility to measure the charge  
 65 produced by single fibers by using a charge sensitive amplifier (CSA) in or-  
 66 der to monitor aging of the fibers. The second version of the ASIC, which is  
 67 currently under test, includes timing capabilities by using a 160 MHz clock in  
 68 combination with a delay locked loop (DLL). The ASICs have been designed  
 69 for count rates up to  $10^8$  Hz.

70

71 Double sided silicon strip detectors ( $90 \times 90 \times 2$  mm<sup>3</sup>,  $2 \times 64$  strips) will be  
 72 used as scatterers (see Fig. 2 for a detector mounted on a printed circuit  
 73 board). Tests with a small-size prototype ( $14 \times 14 \times 2$  mm<sup>3</sup>,  $2 \times 8$  strips) ex-  
 74 hibit a leakage current below 1 nA per strip for temperatures below 0 °C  
 75 and an energy resolution of 8 keV FWHM for the gamma lines of a <sup>133</sup>Ba  
 76 source (81 and 356 keV). The corresponding front-end ASIC [24] comprises  
 77 a fast (15 ns) and a slow shaper (1 μs) for the time and energy information,  
 78 respectively. The second version of the ASIC, which is designed for low noise  
 79 (120 electrons root mean square) and count rates of  $10^5$  Hz, is currently being  
 80 characterized.

81

82 The absorber detector will be composed of 100 BGO blocks ( $38 \times 35 \times 30$  mm<sup>3</sup>  
 83 for each block). Simulation studies for a comparison of different absorber ma-  
 84 terials including LaBr<sub>3</sub>:Ce with its excellent energy and time resolution, have

85 been performed. These simulations showed that cerium doped lutetium yt-  
86 trium orthosilicate (LYSO:Ce) and BGO provide the best performance with  
87 respect to position resolution and efficiency [25]. This is due to a larger  
88 photoabsorption probability in comparison to LaBr<sub>3</sub>:Ce or NaI(Tl). Further-  
89 more, an absorber with dimensions 400×400×30 mm<sup>3</sup> made from LaBr<sub>3</sub>:Ce  
90 would be cost-intensive. Moreover, as compared to LYSO:Ce, BGO avoids  
91 the inconvenience of intrinsic radioactivity, although LYSO:Ce, as well as  
92 LaBr<sub>3</sub> would be faster. The BGO crystals coming from an ancient PET  
93 system are streaked, providing 8×8 pseudo pixels. Each scintillator block is  
94 read out via four PMs (Figure 2 (right)) which allows a reconstruction of  
95 the impact position via a centroid calculation. Results from test measure-  
96 ments with prompt-gamma rays induced by 95 MeV/u <sup>12</sup>C-ions at the Grand  
97 Accélérateur National d’Ions Lourds (GANIL, Caen, France)<sup>3</sup> are given in  
98 Figure 3. For the measurement of the timing resolution (2 ns FWHM), the  
99 HF-signal of the accelerator has been used as reference. The beam time struc-  
100 ture consists of 1 ns pulses every 80 ns. The reconstruction of the impact  
101 position via centroid calculation of the signals from the four PMs reveals the  
102 pixel structure of the scintillator blocks (Fig. 3 (down)).  
103 The data flux of the clinical size prototype will be handled by a Micro  
104 Telecommunications Computing Architecture (μ-TCA) data acquisition sys-  
105 tem [26].

### 106 2.3. Simulations and clinical applicability

107 Geant4 [27] (version 9.6.p02) simulations have been performed for an  
108 optimization of the Compton camera arrangement. These simulations have  
109 also been used to explore the applicability of the present setup under clinical  
110 conditions. Typical parameters for a treatment with a proton beam are:  
111 intensities of  $\sim 10^{10}$  protons/s with beam bunches of 2 ns every 10 ns. This  
112 results in  $\sim 200$  protons per bunch. In the simulations the timing resolutions  
113 of the BGO (3 ns) and silicon detectors (15 ns) have been applied. The  
114 reconstructed vertices of  $10^8$  incident protons, which corresponds to a typical  
115 distal spot in pencil beam scanning [13], are given in Figure 4. In the upper  
116 part of the figure, at clinical intensities, the distribution of *true gamma* events  
117 (i.e. good reconstructible Compton events) is dominated by *other* (random)  
118 coincidences. In the lower part of the figure the beam intensity has been

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<sup>3</sup><http://www.ganil-spiral2.eu/leganil>

119 reduced to one proton per bunch. Here, the fall-off after the Bragg peak (at  
120 position 0) is revealed.

### 121 **3. Conclusion**

122 The status of the development of a clinical-size Compton camera has  
123 been presented. The individual detector components and their corresponding  
124 front-end electronics are under characterization. Simulation studies have  
125 shown that for a usage of the Compton camera to monitor the ion range  
126 during hadrontherapy, the intensity needs to be reduced to one ion per bunch.  
127 In the case of a proton beam with pencil-beam scanning, the duration for  
128 a single spot increases to about 1 s, which can be tolerated for selected  
129 individual spots.

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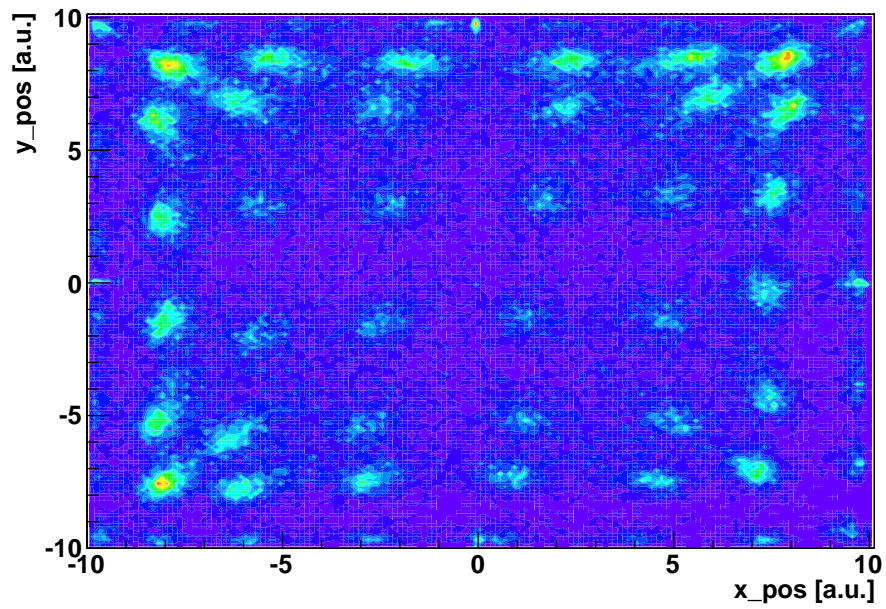
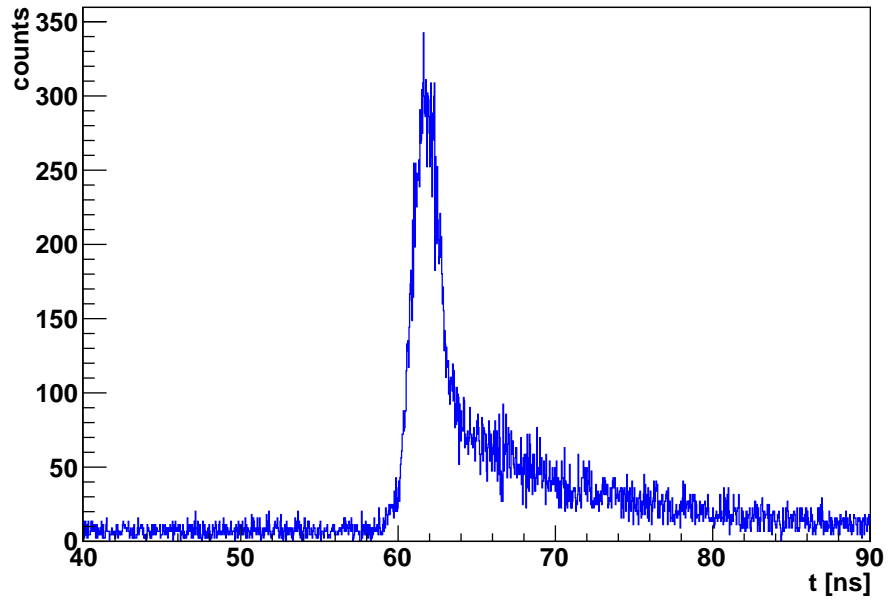


Figure 3: Results from tests with a BGO detector. Up: Measurement of the timing resolution, down: Reconstructed impact positions revealing the pixel structure of the scintillator blocks.

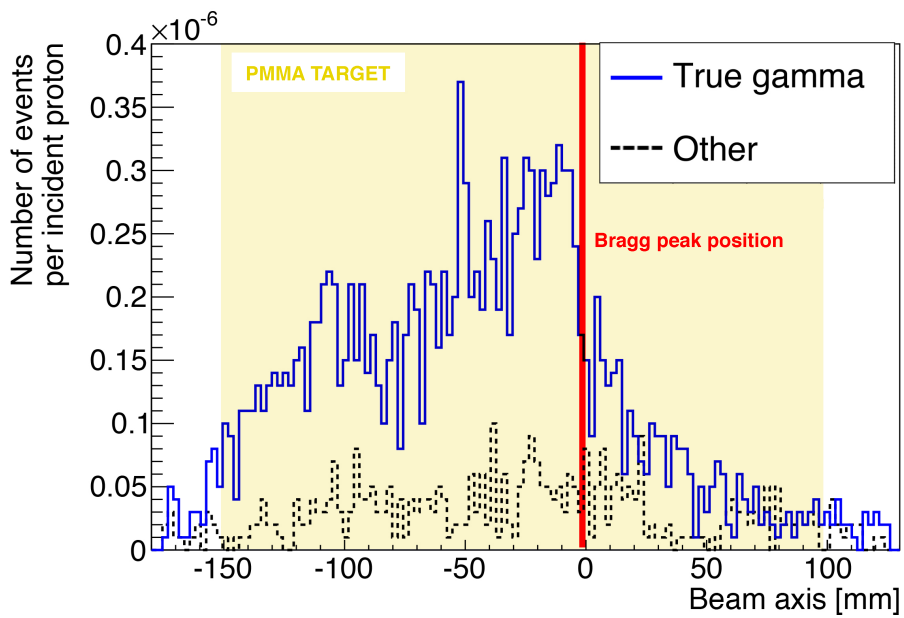
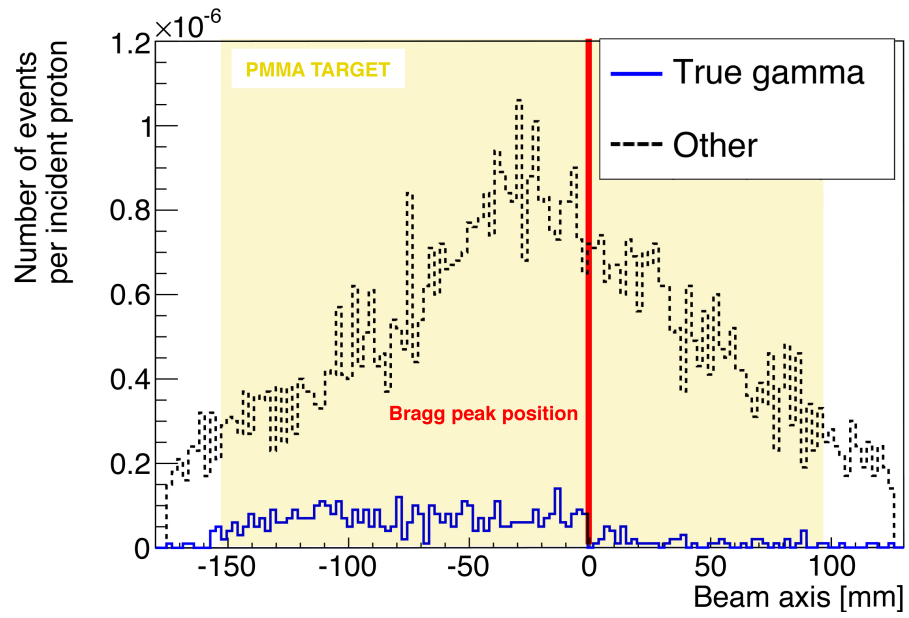


Figure 4: Reconstructed vertices of  $10^8$  incident protons (160 MeV) at clinical (up) and reduced (down) intensities with 200 and 1 protons per bunch, respectively. The location of the Bragg peak is indicated at position 0.