Development of Electron tracking Compton Camera using Micro Pixel Gas Chamber for Medical Imaging

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Abstract

We have developed the Electron Tracking Compton Camera (ETCC) with reconstructing the 3-D tracks of the scattered electron in Compton process for both sub-MeV and MeV gamma rays. By measuring both the directions and energies of not only the recoil gamma ray but also the scattered electron, the direction of the incident gamma ray is determined for each individual photon. Furthermore, a residual measured angle between the recoil electron and scattered gamma ray is quite powerful for the kinematical background-rejection. For the 3-D tracking of the electrons, the Micro Time Projection Chamber (μ-TPC) was developed using a new type of the micro pattern gas detector. The ETCC consists of this μ-TPC (10x10x8cm^3) and the 6x6x13mm^3 GSO crystal pixel arrays with a flat panel photo-multiplier surrounding the μ-TPC for detecting recoil gamma rays. The ETCC provided the angular resolution of 6.6 degrees (FWHM) at 364keV of ^{131}Iodine. A mobile ETCC for medical imaging, which is fabricated in a 1m cubic box, has been operated since October 2005. Here we present the imaging results for the line sources and the phantom of human thyroid gland using 364keV gamma rays of ^{131}Iodine.

Keyword: Compton camera, nuclear medicine imaging, micro pattern gaseous detector, TPC

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1. Introduction

We have realized both full ray tracing and the background rejection for MeV and sub-MeV gamma ray imaging by detecting the direction of the scatted electron in Compton process [1-3]. Although the track of the scattered electron was proposed to be useful for higher energy (>2 MeV) gamma ray in the Compton camera based on silicon strip detectors [4-8], multiple scatterings in dense matter intrinsically prevent from obtaining a sufficient angular resolution of it in those energy regions. Hence, a gas-tracking device looks a unique useful detector to catch such a fine track. By measuring both the directions and energies of a recoil gamma ray and a recoil electron in the gas detector as shown in Figs.1a, the direction of the incident gamma ray can be definitely determined for each Compton scattering. Furthermore, a residual measured angle between the scattered electron and the recoil gamma ray (hereafter, we say α angle) is used for the kinematical background-rejection. This α angle is also calculated from the measured hit positions and energy deposits, and used for the kinematical constraint. Thus, a full ray-tracing method surely provides gamma-ray images of higher quality with less radiation dose. As well known, the detection efficiency of the gas detector is low, but it provides both a large detection volume and an easy handling of signals due to the high gain of it. A simulation study indicates that a 30x30x15cm^3 gas detector with Xe or CF_4 gas pressured by several times atmosphere provides the detection efficiency of a few % for 500keV gamma rays, which is better than the typical efficiency of SPECT. Also such a large volume detector gives a flat efficiency and position resolution as shown in Fig 2. On the other hand, those of PET (Positron Emission Tomography) are quite good at the center, but linearly worse from the center. Compton camera generally have a wide field of view of 2~4str, which covers a large area by ~4 times of the detector size. For the 30 x 30cm gas detector, its detection area covers the whole body. Thus, a large-area ETCC will surely provide us to new benefits even for the use of radio pharmaceutical which are now used for SPECT and PET.

2.Instrument and Imaging Performance

In order to realize such an ETCC, we developed the Micro Time Projection Chamber (μ-TPC) for the 3-D tracking of the recoil electrons [1-3]. Typical reconstructed tracks of the low energy electrons in Compton scattering are shown in Fig. 1b. The μ-TPC consists of a new type of the gaseous proportional two-dimensional wireless position-sensitive detector, or a Micro Pixel Gas Chamber (μ-PIC) [9,10] as shown in Fig1c, and a drift volume. The prototype camera consisted of the 10 x10x8cm^3 μ-TPC with a argon-ethane gas mixture (9:1) and the Anger camera with a large 30cm x30cm x1.5cm^3 NaI(Tl) single scintillator and a 6x6 2-inch photo-multiplier array for detecting the scattered gamma rays [11]. For this prototype, the angular resolutions of 12 degree and 34 degree (FWHM) for AMP (Angular Resolution Measure) and SPD (Scatter Plane Deviation) were obtained for 662keV gamma rays when the energy of the incident gamma ray was used as a known parameter [11]. In 2005 we improved this detector as follows; the uses of GEM developed by Sauli et al. [12] as an intermediate electron multiplier above the μ–PIC and 6x6x13mm GSO crystal pixel arrays with a flat panel photo-multiplier surrounding the base and side of the μ–TPC instead of the Anger camera [13-15]. By these improvements we achieved a stable operation of the μ–TPC with a high gain of >20,000 (μ-PIC:2000, GEM:10) during one year, and obtained the energy resolution of the μ-TPC with ~20% at 22.2keV. In addition, good energy and position resolutions of GSO scintillation array were obtained to be ~3mm and 9% at 662keV (FWHM), respectively. Then, combined energy resolution of the ETCC for gamma rays was obtained 15% at 662kev (FWHM). Using this improved ETCC we obtained an ARM resolution of ~7 degree at 662keV (FWHM) with measuring the total energy of the incident gamma ray, of which details are reported in [15]. Based on this improved ETCC, we have developed a mobile Compton camera fabricated within a 1m cubic box as shown in Fig.3 [15]. The angular resolutions of ARM and SPD at 364keV gamma rays of ^{131}Iodine were improved from 12degree and 100 degree to 6.6 degree and 77 degree due to the improvement of the electron reconstruction. We have measured simultaneously the images of the two different energy gamma-ray sources (^{133}Ba: 356keV & ^{137}Cs 662keV) as shown in Fig.4, where two images are obviously distinguished by the energy spectrum. This is a unique ability expected mainly for the Compton camera in medical use. In the reference [15], similar images using the point sources of ^{137}Cs: 662keV & ^{133}Mn: 853keV were already presented. In this time we have obtained the image in the energy region used in nuclear medicine. From this figure, we infer that an image of gamma rays at 511keV would be separated from that of ^{131}Iodine or ^{67}Ga. For medical imaging, the recognition of the extended distribution of isotopes in the body is quite inevitable. Using this mobile ETCC, we tried to get the image of the phantom of human thyroid gland for 364keV gamma rays of ^{131}Iodine. Before measuring the images, the acceptance of ETCC was measured by moving the ^{131}Iodine point source on the plane at the 20cm front from the center of the ETCC, and obtained its acceptance plotted in Fig.5a. One should note the ETCC covers about four times the size of the μ–TPC (the region having a better acceptance than a half of the maximum acceptance). Using this acceptance, the images of a 30cm x 1cmϕ line source filled with ^{131}Iodine were measured by changing its position on vertical and horizontal as shown in Fig.6. Also the image of two line sources filled with ^{131}Iodine is presented in Fig.7 after applying the acceptance correction. Obviously two lines are distinguished although its shape is distorted. In both Figs.6 and 7, the acceptance of the detector was crucial for getting the shape of the images. The distortion of the line sources in the under part may indicate the necessity of the more accurate measurement for the acceptance. Based on such a study, the image of the phantom of the human thyroid grand filled with ^{131}Iodine was measured. The phantom was set on the 20cm front from the center of the ETCC. Figures 5b and 5c shows the images before and after applying the acceptance corrections, where you recognize the shape of the phantom although the image is still blurred.

3.Perspective and Discussion

As mentioned in Introduction, Compton camera, if its imaging ability were similar to that of PET, would have excellent features for nuclear medicine and molecular imaging, which are not be realized by SPECT and PET. We have developed the ETCC and obtained several images for the extended sources. For reconstructing the extended images, an acceptance correction is found quite significant. While any type of Compton camera is expected to have a wide field of view, our results shows that a half of the target size is at lease necessary as a detection area to reconstruct the image of the extended target. In the observation for the human organs, more 10x10cm^2 detection area looks needed. A Compton camera based on the gas detector easily provides such a large detection area and a flat acceptance. At present obtained angular resolution is not enough for the imaging of the human organs, however simulation shows that the angular resolution of the 5degree would provide a similar quality of the image to that of PET. In general a Compton camera provides better images as energy increasing. A simple extrapolation from the angular resolution at 364kev says that the ETCC will have an angular resolution of ~5 degree at 511keV. In 2006 July, we have observed phantom images using ^{18}F (511keV)-ion and ^{18}F- FGD for a rat, of which results will appear soon. In addition a large μ–TPC (30x30x15cm^3) for the ECC imaging a human body is now being tuned.

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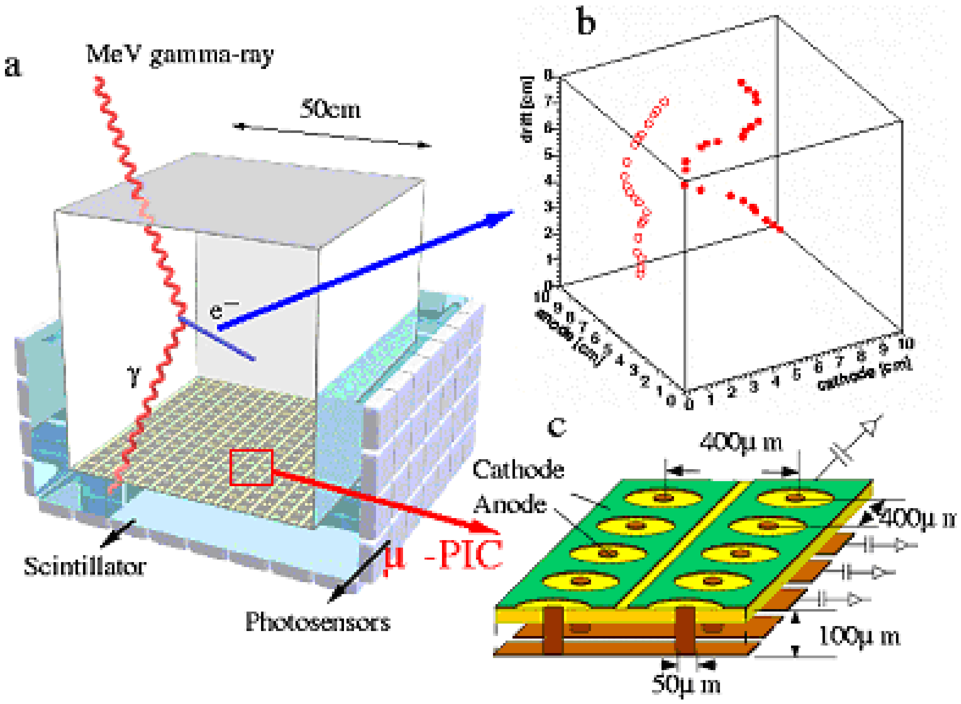

Fig.1: (a) Conceptual structure of Electron Tracking Compton Camera (ETCC). (b) Typical tracks of electrons from β-decay detected by the μ-TPC. (c) Schematic structure of μ−PIC.

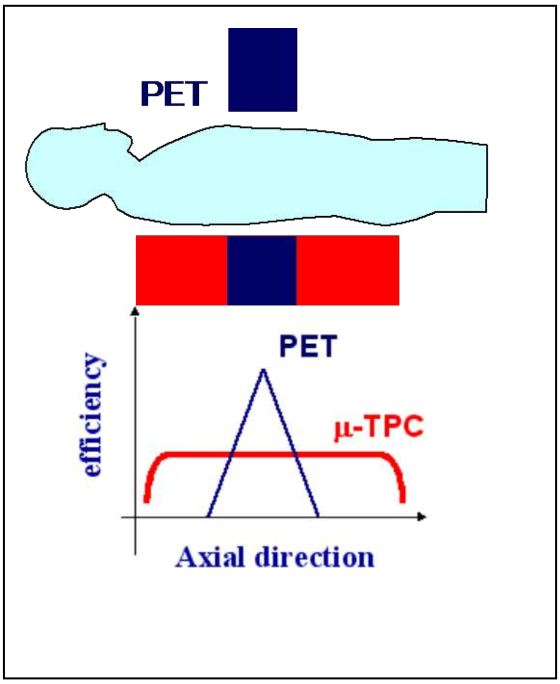

Fig.2: Schematic comparison of the efficiencies of PET and ETCC along the Axial direction of a human body

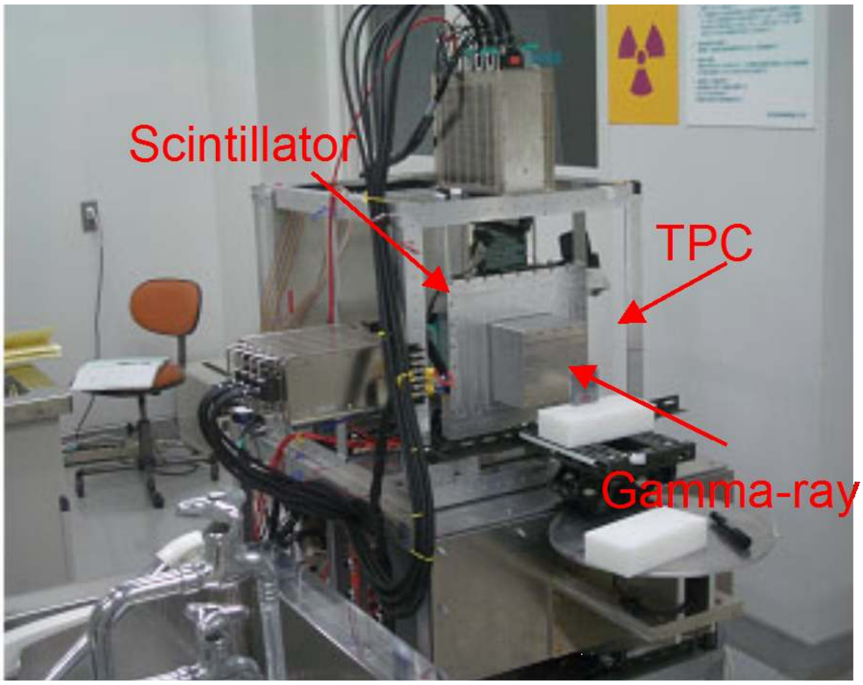

Fig.3: Photograph of the mobile ETCC. All detectors, electronics, and computers are installed in the wagon.

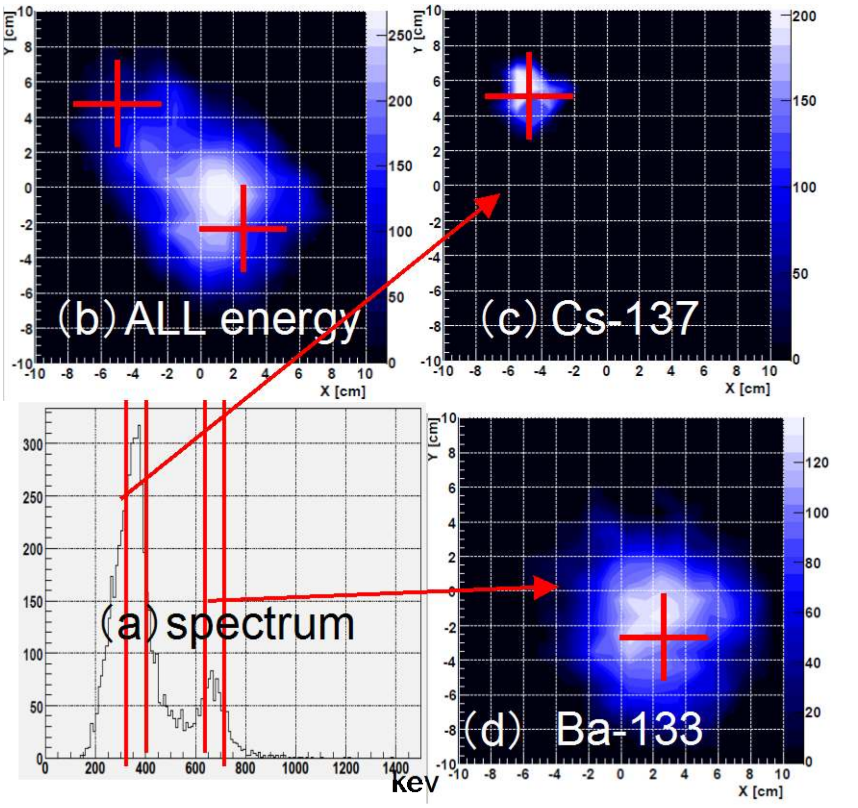

Fig.4: Images obtained by the simultaneous irradiation of ^{133}Ba: 356keV & ^{137}Cs 662keV.

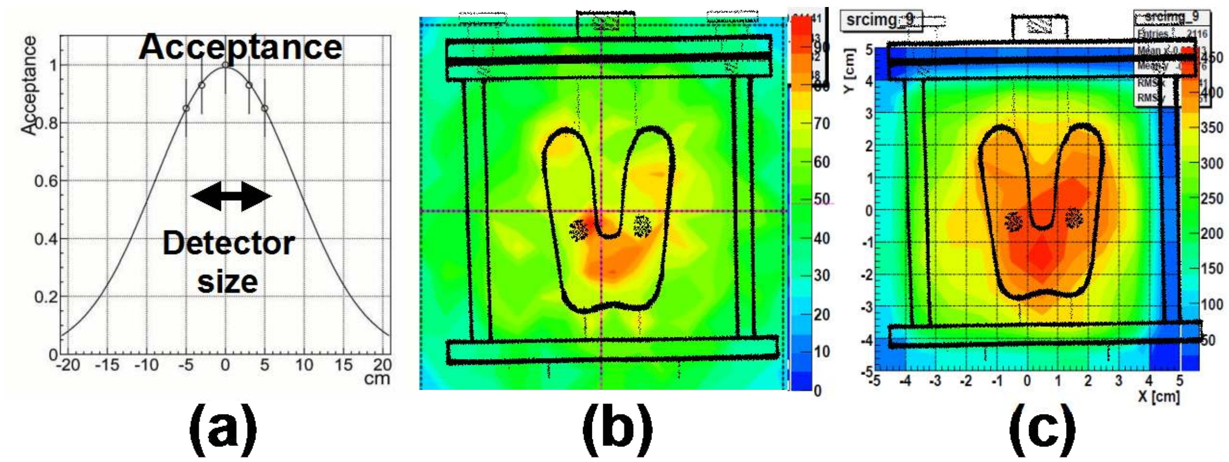

Fig.5: (a) Acceptance of the detector, in which horizontal axis is an arbitrary unit. (b) and (C) are the images of the phantom of the human thyroid grand filled with ^{131}Iodine (364keV) before and after applying the acceptance correction, respectively. The phantom was set on the 20cm front from the center of the ETCC.

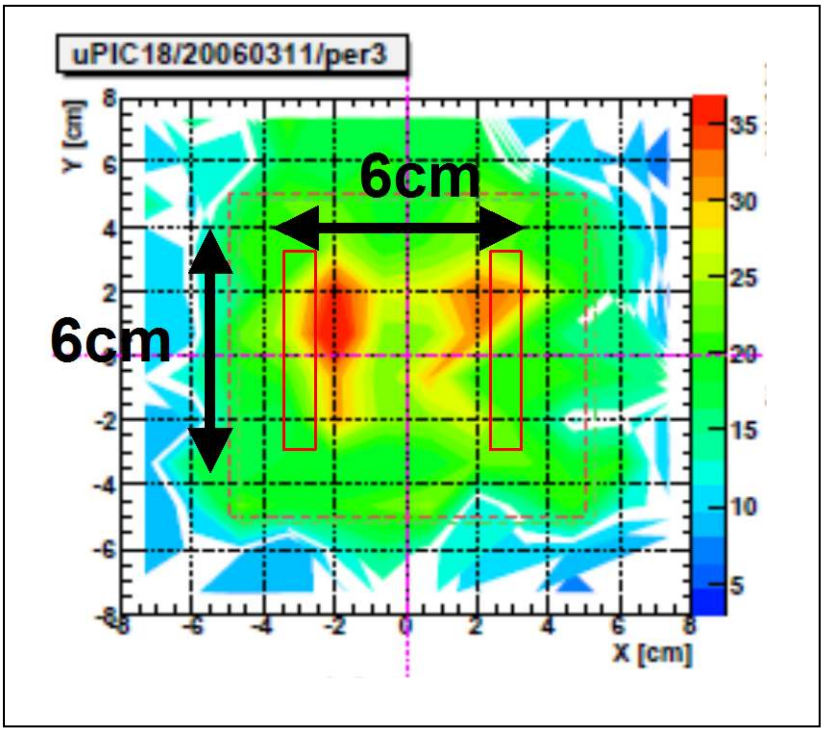

Fig.6: Image of two line sources filled with ^{131}Iodine (364keV), where red lines indicate the positions of two line sources. Two line sources are set on the 20cm front of the ETCC.

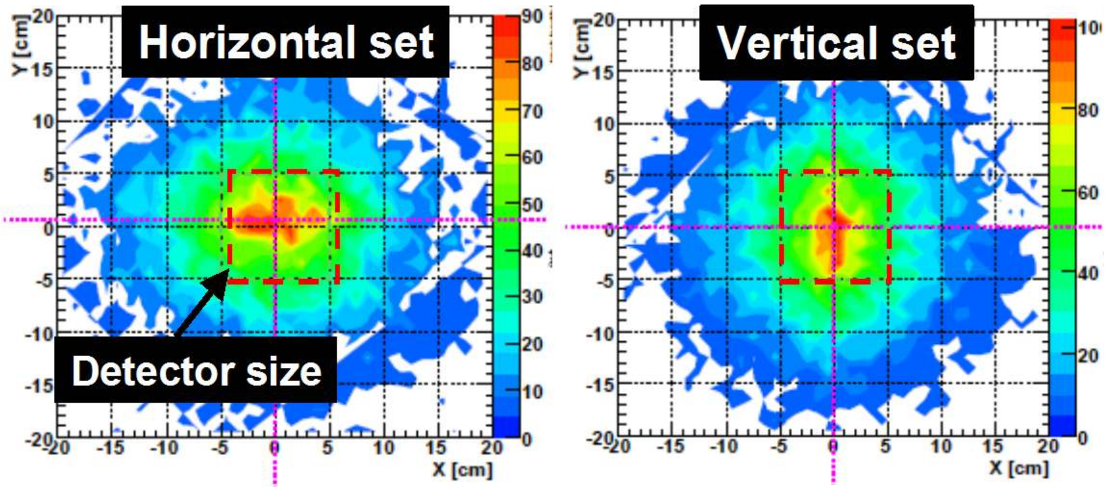

Fig.7: Images of a long line source (1cmϕx30cm) filled with ^{131}Iodine (364keV) setting on vertical and horizontal, respectively (both are set on the 20cm front of the ETCC).