REPORTS OF PRACTICAL ONCOLOGY AND RADIOTHERAPY 21 (2016) 102–107

Available online at www.sciencedirect.com

ScienceDirect

journal homepage: http://www.elsevier.com/locate/rpor

Original research article

Study on measuring device arrangement of array-type CdTe detector for BNCT-SPECT

Masanobu Manabe*, **Soichiro Nakamura**, **Isao Murata**

Division of Electrical, Electronic and Information Engineering, Graduate School of Engineering, Osaka University, Yamada-oka 2-1, Suita, Osaka 565-0871, Japan

ARTICLE INFO

Article history:
Received 12 March 2014
Received in revised form
12 December 2014
Accepted 6 April 2015
Available online 11 June 2015

Keywords:
BNCT
CdTe detector
Gamma-ray
SPECT
Collimator
Anti-coincidence

ABSTRACT

Aim: To design the measuring device arrangement of array-type CdTe detector for BNCT-SPECT.

Background: In a boron neutron capture therapy, a very serious unsolved problem exists, namely that the treatment effect for BNCT cannot be known during irradiation in real time. Therefore, we have been developing a so-called BNCT-SPECT with a CdTe detector, which can obtain a three-dimensional image for the BNCT treatment effect by measuring 478 keV gamma-rays emitted from the excited state of ^{7}Li nucleus created by the ^{10}B(n,α) reaction. However, no practical uses were realized at present, because BNCT-SPECT requires very severe conditions for spatial resolution, measuring time, statistical accuracy and energy resolution.

Materials and methods: The design study was performed with numerical simulations carried out by a 3-dimenaional transport code, MCNP5 considering the detector assembly, irradiation room and even arrangement of arrayed CdTe crystals.

Results: The estimated count rate of 478 keV gamma-rays was sufficiently large being more than the target value of over 1000 counts/h. However, the S/N ratio did not meet the target of S/N > 1. We confirmed that deterioration of the S/N ratio was caused by the influence of Compton scattering especially due to capture gamma-rays of hydrogen. Theoretical calculations were thereafter carried out to find out whether anti-Compton measurement in an array-type CdTe detector could decrease the noise due to Compton scatterings.

Conclusions: The calculation result showed that the anti-coincidence would possibly increase the S/N ratio. In the next phase, an arrayed detector with two CdTe crystals will be produced to test removal possibility of the anti-coincident event.

* *Corresponding author*. Tel.: +81 668797894.
E-mail address: mmanabe@ef.eie.eng.osaka-u.ac.jp (M. Manabe).

http://dx.doi.org/10.1016/j.rpor.2015.04.002
1507-1367/

1. Background

Recently, BNCT (boron neutron capture therapy) has attracted many scientists in a medical field as a new radiation therapy. BNCT can destroy tumor cells by alpha particles (α) and lithium nuclei (^{7}Li) emitted by the reaction of thermal neutrons or epithermal neutrons with boron (^{10}B). Ranges of emitted α and ^{7}Li particles are as long as the size of a human body cell. Hence, if ^{10}B would be accumulated only in tumor cells, it would be expected that only the tumor cells would be killed with low damage of the healthy cells.

However, this therapy has not been established yet as a commonly utilized therapy, because no medical neutron sources are available and, in addition, there are some other serious problems to be solved as early as possible. One of them is that the treatment effect cannot be known during BNCT in real time. The present study describes a SPECT system for BNCT named BNCT-SPECT, which we are now developing. This is a three dimensional imaging device to monitor the local BNCT dose (treatment effect) around the tumor in real time. The BNCT-SPECT device is shown in Fig. 1. ^{10}B(n,α)^{7} Li reaction is expressed by the next two nuclear reactions:

$$\begin{aligned} {}^{10}\mathrm{B} + \mathrm{n} &\rightarrow \alpha + {}^{7}\mathrm{Li} + 2.79\,\mathrm{MeV} \quad (6.1\%) \\ &\rightarrow \alpha + {}^{7}\mathrm{Li}^{*} + 2.31\ \ \mathrm{MeV} + \gamma \quad (478\,\mathrm{keV}) \quad (93.9\%) \end{aligned} \tag{1}$$

94% of residual ^{7}Li is in the first excited state, i.e., ^{7}Li*. ^{7}Li* decays in its half-life of 10^{-14} s to emit a 478 keV gamma-ray via transition from the first excited state to the ground state. If the intensity distribution of 478 keV gamma-rays could be measured three-dimensionally, we could obtain the three-dimensional distribution of ^{10}B(n,α)^{7} Li reaction rate in the tumor. The result of the measurement can then be regarded as the treatment effect of BNCT. Emitted 478 keV gamma-rays are collimated by the collimator and measured by many measuring devices in the array detector in order to make an image of gamma-ray intensity. The BNCT treatment effect (local tumor dose) can be estimated from the obtained three-dimensional gamma-ray image.

However, the 478 keV gamma-rays must be measured in a very high neutron flux field, because we have to use a very intense neutron source. Especially, capture gamma-rays of 2.22 MeV produced by the ^{1}H(n,γ)^{2}H reaction and annihilation gamma-rays of 511 keV to be detected just adjacent to 478 keV gamma-rays become a very large and critical background. Also, actual medical conditions, i.e., protocol, irradiation site and so on, must be considered. Taking into account the above situation, we set four design criteria as follows:

(1) The spatial resolution should be about several mm in the obtained SPECT image from the viewpoint of medical treatment.
(2) It is necessary to complete a measurement in about 60 min, because the treatment time of BNCT is normally less than 1 h.
(3) The number of counts per unit detector should be more than 1000 counts so that the statistical accuracy can be kept to be less than several percent.
(4) The energy resolution, full width at half maximum (FWHM), should be less than 33 keV (511–478 keV) so as to measure annihilation gamma-rays and 478 keV prompt gamma-rays separately.

Finally, we decided to employ a CdTe device as an elemental gamma-ray measuring device after precise investigation.[1]

Practically, at first, by theoretical calculations the necessary efficiency of the elemental CdTe detector was fixed. The size of the CdTe crystal was then determined meeting the spatial resolution requirement and keeping the necessary efficiency.[2,3] Consequently, we confirmed that the thickness of more than 30 mm was necessary in case of assuming that the incident surface was 2×1.5 mm.[4]

Next, the expected performance was confirmed through test measurements with an actually produced CdTe crystal. It was found from the measurement that 478 keV and annihilation (511 keV) gamma-rays could be measured separately, although it has been pointed out to be a critical problem so far.[5,6]

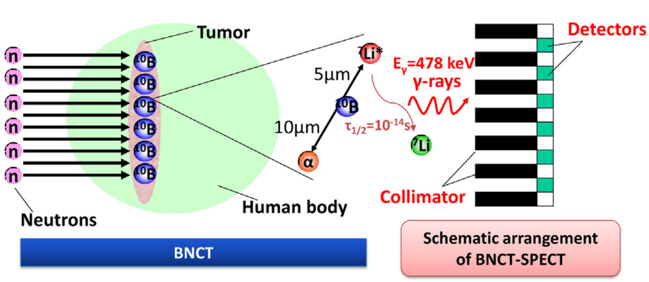

Fig. 1 – Principle of BNCT-SPECT.

2. Aim

The purpose of this study is to design the measuring device arrangement of array-type CdTe detector for BNCT-SPECT. In this study, we investigate the design feasibility of the BNCT-SPECT system considering the detector assembly, irradiation room and even arrangement of CdTe crystals. At first, we design a suitable collimator which would be a very important part of the array-type CdTe detector. As the design target, we set the number of counts being greater than 1000 for 60 min at 478 keV peak and, supplementarily, set signal to noise (S/N) ratio being greater than unity. As mentioned later in Fig. 6, the S/N ratio is not acceptable. To improve the S/N ratio at the photopeak of 478 keV, anti-coincidence measurement is examined by using an array-type CdTe detector.

3. Materials and methods

We designed a collimator for an array-type CdTe detector so that a sufficient number of counts (>1000) could be obtained within 1 h with an acceptable S/N ratio in a real BNCT scene. Fig. 2 shows a schematic drawing of the collimator. The CdTe detector size was assumed to be achievable maximum dimensions (2 mm × 2.5 mm × 40 mm) at present in Japan. The collimator hole diameter was determined to be 2 mm considering the CdTe detector size and the spatial resolution. The

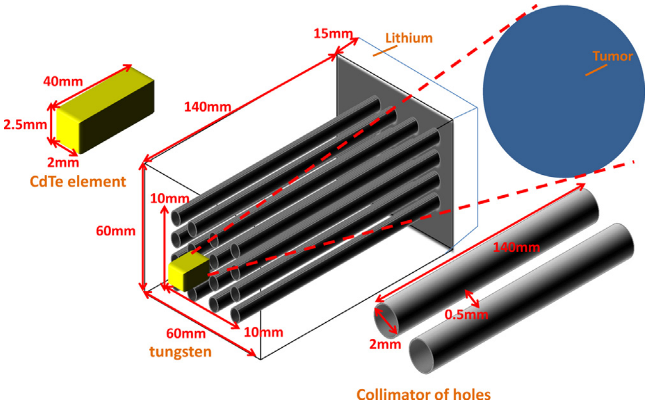

Fig. 2 – Collimator diameter and number of holes considered for design of an array-type CdTe detector.

collimator length is set to be 15.5 cm because the number of counts should be larger than 1000 counts/h. In this case, 4×4 (16 crystals) should be taken into account in the design, because the CdTe detector can view the entire tumor through the region containing these holes. By this assumption, a reasonable design, that is, the noise level reaches its maximum, is possible.

The design was carried out taking into account the actual BNCT site by using MCNP5 (Three-dimensional Monte Carlo N-Particle Transport Code). As for a neutron source condition, a 10 keV broad parallel neutron beam was assumed for simplicity of the design calculation. Considering common cases, a tumor of 3 cm in diameter was modeled in which 20 ppm boron was accumulated. We employed F4, F5 and F8 tallies of MCNP5 to estimate the reaction rate, flux and pulse height spectrum, respectively.

Calculations are divided into two steps, i.e., source term and pulse height spectrum calculations as in the following.[7]

Firstly, we calculated the source terms, concerning following 4 items:

(1) ^{10}B(n,α) reaction rate in the tumor.
(2) ^{113}Cd(n,γ) reaction rate in the detector.
(3) Neutrons reaching the detector.
(4) Gamma-rays reaching the detector except, (1) and (2) (^{1}H(n,γ) from the human body is included here).

Secondly, we calculated four pulse height spectra below measured by the CdTe detector with F8 tally by using the calculated source terms above.

(1) 478 keV gamma-ray spectrum.
(2) ^{113}Cd capture gamma-ray spectrum.
(3) Spectrum induced by neutrons reaching the detector.
(4) Spectrum induced by gamma-rays reaching the detector, except (1) and (2) (^{1}H(n,γ) from the human body is included here).

Finally, summing up the results from (1) to (4), we calculated the pulse height spectrum to be measured with the present CdTe detector.

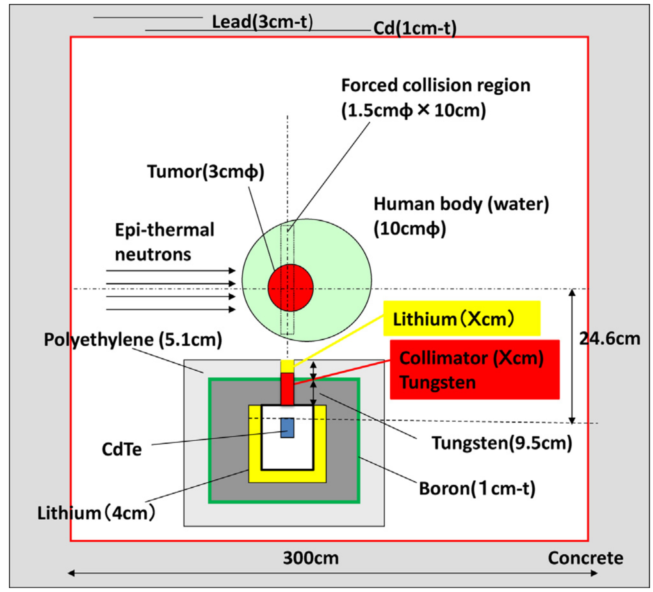

Fig. 3 – Final design result of the collimator for the present BNCT-SPECT system taking into account an even irradiation room.

4. Results

4.1. *Collimator design for array-type CdTe detector*

Fig. 3 shows the finally obtained collimator design model for the present BNCT-SPECT system.

The designed model is composed of an irradiation room, detector assembly and human body phantom. As shown in the figure, the CdTe detector is mainly shielded with polyethylene and boron and especially with tungsten, boron and lithium in the direction to the CdTe detector, because there are collimator holes in the same direction. The thicknesses of the tungsten (W) and lithium (Li) are parameters changed in the design calculations. The results of the design calculations are shown in Figs. 4–6. Fig. 4 shows the final pulse height spectrum

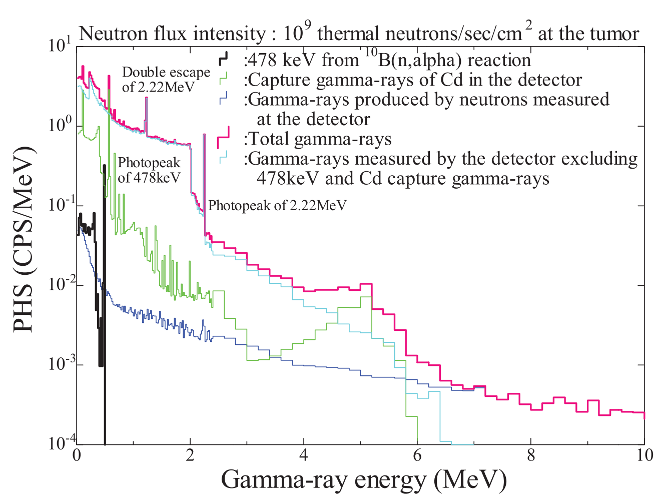

Fig. 4 – Calculated PHS of energy up to 10 MeV.

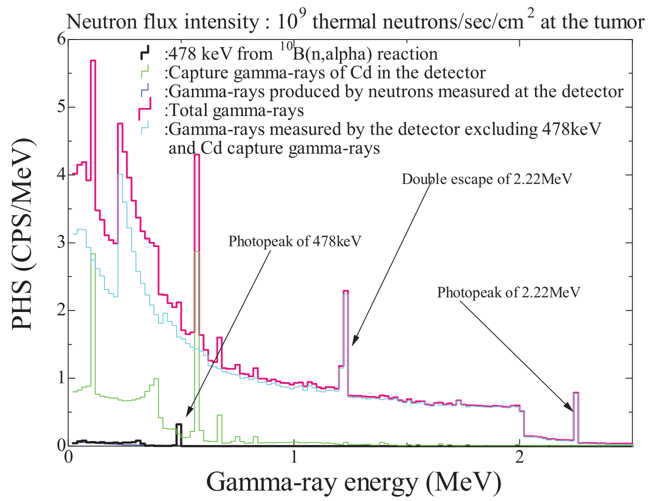

Fig. 5 – Calculated PHS at 478 keV nearby.

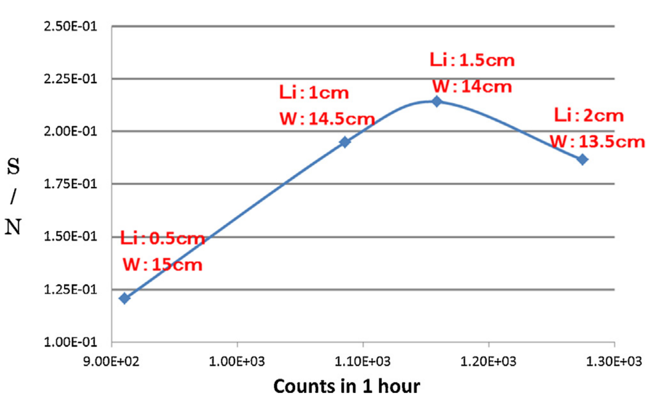

Fig. 6 – S/N ratio and count rate at 478 keV as a function of Li and W thicknesses.

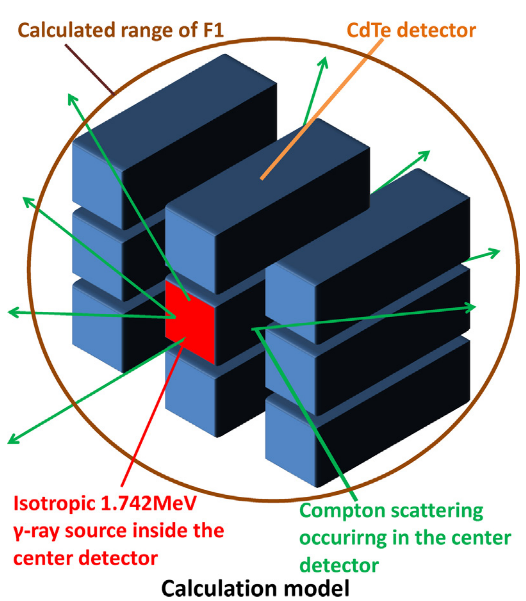

Fig. 7 – Anti-coincidence calculation model.

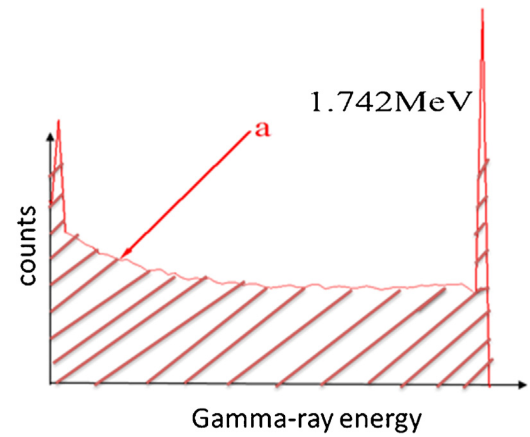

Fig. 8 – Calculated leak current.

including 4 components mentioned above. Fig. 5 is the one in the low energy region around 478 keV. From this result, it was confirmed that a photopeak of 478 keV gamma rays emitted from ^{10}B(n,α)^{7}Li reaction is distorted greatly by noises due to gamma-rays other than 478 keV gamma-rays. It can also be expected that the most dominant influence is caused by the Compton continuum created by 2.22 MeV gamma rays emitted from the ^{1}H(n,n)^{2}H reaction. At the photopeak of 478 keV, the contribution ratio of 2.22 MeV gamma-rays is 72%.

Fig. 6 shows the S/N ratio and net count rate at 478 keV as a function of W and Li thicknesses in front of the collimator. From the figure, the optimum thicknesses of W and Li are found to be 14 cm and 1.5 cm, respectively. In this case, the count rate of 478 keV is 1159 counts per 60 min, which is larger than the target value. On the other hand, the S/N ratio is 0.21 being less than the supplementary target value of unity.

4.2. *Anti-coincidence performance estimation by using an array-type CdTe detector*

4.2.1. *Anti-coincidence calculation model*

As described in the previous section, the deterioration of the S/N ratio is induced mainly by the influence of Compton scattering of gamma-rays emitted from the ^{1}H(n,n)^{2}H reaction occurring in the water phantom. To remove or decrease the Compton contribution, conventionally, anti-Compton detection technique could be applied in these cases. Fortunately,

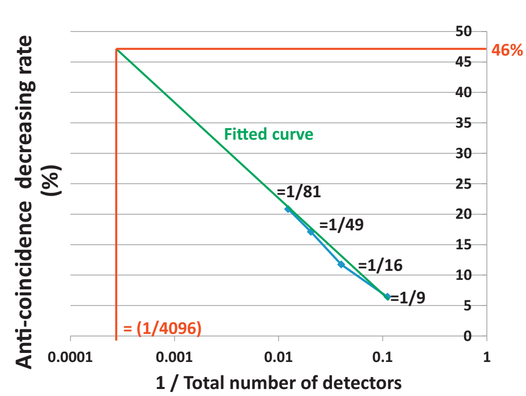

Fig. 9 – Anti-coincidence decreasing rate.

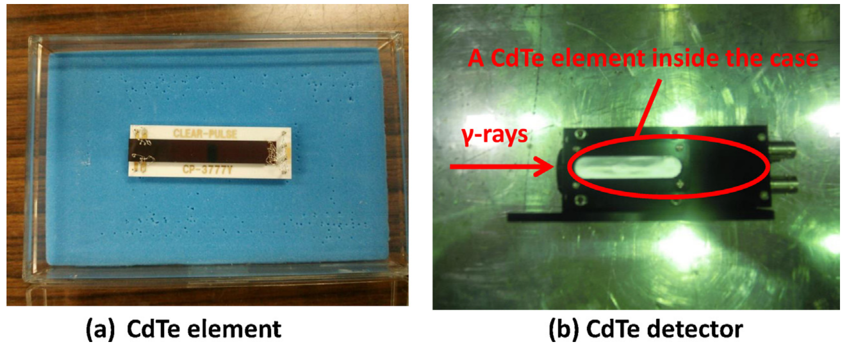

Fig. 10 – Photo of the two-element CdTe detector.

in the present BNCT-SPECT system, there are a lot of CdTe crystals stacked to form an arrayed detector system. We then examined whether anti-Compton measurement in the arrayed CdTe crystals could decrease the noise due to Compton scatterings. The analysis was carried out with MCNP5. Fig. 7 shows an example of the calculation model.

This is a case of nine elemental detector crystals arranged, in which the central detector is surrounded by 8 CdTe crystals having a function to detect Compton gamma-rays emitted from the central one to reduce the contribution of the Compton continuum. In the calculation, firstly, it was assumed that 2.22 MeV gamma-rays are incident only to the central detector in parallel with the detector axis. Compton contribution included at 478 keV photopeak is created mainly by 2.22 MeV gamma-rays. This is actually formed by 478 keV Compton electrons. In this case, the Compton scattering gamma-ray has a fixed energy of 1.742 MeV. By detecting this gamma-ray with surrounding 8 detectors, the Compton continuum can be reduced by the anti-coincidence detection technique.

Fig. 8 shows the calculated leakage current of 1.742 MeV gamma-rays emitted in the central detector by F1 tally shown in Fig. 7. If the integral value of Fig. 8 is a, full energy absorption of 1.742 MeV in the surrounding detectors is calculated to be $1 - a$.

By the next equation, the percentage of anti-coincidence detection can be deduced without so-called "event mode".

$$\eta(1 - a) \times 100\%, \quad (2)$$

where η is a correction factor for the angular distribution of secondary gamma-rays emitted in the central detector, because in the calculation an isotropic source was used. In fact, the emitted Compton scattering gamma-rays showed a forward peaked distribution. The effect, η, was thus estimated from MCNP5 calculations as follows: at first, η is defined as the ratio b/c. The value c is evaluated by counting the number of 1.743 MeV gamma-rays entering the surrounding detectors after their being emitted isotropically in the center detector. And the value b is similarly evaluated as the number of 1.742 MeV gamma-rays which are emitted in the center detector via interaction of 2.22 MeV gamma-rays incident to the center detector. As a result, it was found that η was not so largely varying and the value is converged to be around 0.9 with increase of the number of detectors.

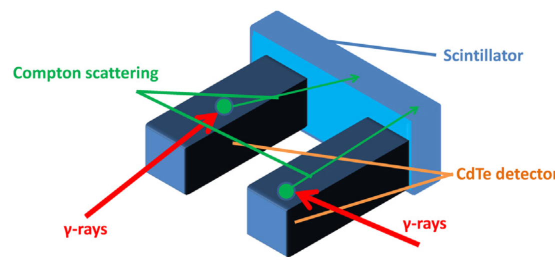

Fig. 11 – Schematic view of improved anti-coincidence detection with scintillator.

4.2.2. *Anti-coincidence decreasing rate*

Fig. 9 shows the result of anti-coincidence decreasing rate obtained by Eq. (2) as a function of the number of detectors surrounding the central detector. The horizontal axis is reciprocal of the total number of detectors and the vertical axis is the anti-coincidence decreasing rate. We calculated the anti-coincidence decreasing rates for several cases of 3×3, 4×4, 7×7 and 9×9 arrangements.

In the present BNCT-SPECT system, one arrayed detector has 4096 (64×64) CdTe crystals. And the BNCT-SPECT consists of 4 arrayed detectors. The anti-coincidence decreasing rate of the present system is estimated to be 46% by extrapolation as shown in Fig. 9. Finally the S/N ratio would be almost doubled, which is 0.45 (=0.21 (S/N ratio in the last chapter) $\times$ 1/0.46 (anti-coincidence effect)). Also, coefficient of determination R^2 was evaluated to be 0.98, that is, the fitted curve becomes mostly a straight line. Although this value is not so high, i.e., actually not beyond unity, measurement would be possible, because the estimated statistical error can be suppressed below 10% even in this case, if the peak net counts of 1000 are obtained.

4.3. *Future work*

As a future work, we are planning to carry out measurement of actual anti-coincidence detection ratio by using a two-element CdTe detector, which was designed and produced at the end of last year as shown in Fig. 10.

Furthermore, to reduce the Compton base at 478 keV more substantially, a scintillation detector is planned to be set just behind the CdTe detector as shown in Fig. 11 to carry out anti-coincidence detection also with the use of signals from the scintillator.

5. Conclusions

We investigated the feasibility of the BNCT-SPECT system by simulations with MCNP5 considering the detector assembly, irradiation room and even arrangement of arrayed CdTe crystals.

At first, collimator design was carried out for the present BNCT-SPECT. As a result, the best result was obtained in the case of a combination of 14 cm thick tungsten and 1.5 cm thick lithium shields. Practically, the number of counts is greater than the target value. However, the S/N ratio could not exceed the target value of unity.

Anti-coincidence performance estimation was thereafter carried out to improve the S/N ratio at 478 keV by using an array-type CdTe detector by simulations with MCNP5. As a result, by considering the arrangement of an arrayed CdTe detectors, the anti-coincidence decreasing rate at 478 keV due to Compton scattering of 2.22 MeV could be reduced by about 46%, that is, the S/N ratio of 0.42.

Conflict of interest

None declared.

Financial disclosure

This research presentation was received support by a part of travel expenses from Research Foundation for the Electrotechnology of Chubu.

Acknowledgment

This work was partly supported by a research granted from Research Foundation for the Electrotechnology of Chubu.

REFERENCES

1. Murata I, Mukai T, Ito M, Miyamaru H, Yoshida S. Feasibility study on BNCT-SPECT using a CdTe detector. *Prog Nucl Sci Technol* 2011;**1**:267–70.
2. Mukai T, Murata I, Ito M, Miyamaru H. Development of CdTe detector for BNCT-SPECT. In: *Proceedings of the 2009 annual symposium on nuclear data, vol.* 005. 2010. p. 75–80.
3. Murata I, Mukai T, Nakamura S, Miyamaru H, Kato I. Development of a thick CdTe detector for BNCT-SPECT. *Appl Radiat Isot* 2011;**69**:1706–9.
4. Nakamura S, Mukai T, Manabe M, Murata I. Characterization test of CdTe detector element designed and developed for BNCT-SPECT. In: *Proceedings of the 2011 annual symposium on nuclear data, vol.* 005. 2012. p. 165–70.
5. Murata I, Nakamura S, Manabe M, Miyamaru H, Kato I. Characterization measurement of a thick CdTe detector for BNCT-SPECT – detection efficiency and energy resolution. In: *Presented in 15th international congress on neutron capture therapy (ICNCT-15)*. 2012.
6. Manabe M, Nakamura S, Murata I. Separate measurement of 478 keV and annihilation gamma-rays by CdTe detector for BNCT-SPECT. In: *Presented in 7th international symposium on radiation safety and detection technology (ISORD-7)*. 2013.
7. Nakamura S, Mukai T, Manabe M, Murata I. Precise numerical simulation of gamma-ray pulse height spectrum measured with a CdTe detector designed for BNCT-SPECT. *Prog Nucl Sci Technol* 2012;**3**:52–5.