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Design study of a dedicated head and neck cancer PET system

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Abstract

The tumor-involved regions of head and neck cancer (HNC) have complex anatomical structures and vital physiological roles. As a consequence, there is a need for high sensitivity and high spatial resolution dedicated HNC PET scanner. The purpose of this study is to evaluate and optimize system design that includes detecting materials and geometries. For the detecting material, two scanners with the same two-panel geometry based on CZT and LYSO were evaluated. For the system geometry, four CZT scanners with two-panel, lengthened two-panel, four-panel, and full-ring geometries were evaluated. A cylinder phantom with sphere lesions and an XCAT phantom in the head and neck region were simulated. The results showed that the sensitivity of the 40-*mm* thickness CZT system and the 20-*mm* thickness LYSO system were comparable. However, the multiple interaction photon events recovery accuracy of the CZT system was about 20% higher. The in-panel and orthogonal-panel spatial resolutions of CZT are 0.58 and 0.74 *mm*, while those of LYSO are 0.70 and 1.40 *mm*. For system geometry, the four-panel and full-ring scanners have a higher contrast recovery coefficient (CRC) and contrast-to-noise ratio (CNR) than the two-panel and lengthened two-panel scanners. However, a 5-*mm* lesion in the XCAT phantom was visualized within 6 *min* in the two-panel system.

Index Terms—

Head and neck cancer; Dedicated PET; System design; Monte Carlo simulation

I. Introduction

Head and neck cancer (HNC) accounts for approximately 4% of all cancers in the United States [1] and the overall annual mortality rate is 23% [2]. Whole-body positron emission tomography (PET) and its combination with computed tomography (CT) are commonly used for HNC diagnosis, staging, treatment planning, and assessing response to therapy [3]–[6]. Compared to CT, magnetic resonance imaging, sonographic and histopathological findings, PET imaging shows the highest sensitivity and specificity for detecting lymph node metastases of HNC [7].

However, lesions in this region can be challenging to diagnose due to the thin, soft tissues within the neck, which require a high-resolution imaging system. The spatial resolution of whole-body PET is typically 4 to 6 *mm* [8], [9]. For structures less than twice the reconstructed image resolution, the true amount of activity is not completely captured [10]. The poor spatial resolution of whole-body PET hinders the precise delineation of the primary tumor of HNC and limits the detection of tumor involvement in lymph nodes smaller than 4 to 5 *mm*. Besides, it also results in a large number of reported false-negatives in lymph nodes (for example 80%) [11]–[13]. Researchers [14] from Duke University Medical Center have introduced a dedicated HNC PET acquisition protocol with longer scan time in the HN bed position to improve the detection ability of PET imaging in HNC. The dedicated HN protocol has advantages in detecting lymph nodes smaller than 15 *mm* compared to the standard protocol. However, in terms of evaluating primary tumors and detecting lymph nodes smaller than 10 *mm*, there are no significant differences between the standard protocol and the dedicated HN protocol [13], [14].

Due to the complex anatomy and vital physiological role of the tumor-involved structures, the goal of HNC treatment is not only to improve survival outcomes but also to preserve organ functions [15]. An improvement in resolution in a PET image to better define the boundary of tumors is significant for the treatment planning and monitoring of HNC. In supraglottic squamous cell carcinoma, for example, if it involves thyroid cartilage, it is T4 and unresectable. If it does not involve thyroid cartilage, it is T3 and can be cured with surgery. Another example is that if supraglottic squamous cell carcinoma does not cross the anterior commissure, patients can have supraglottic laryngectomy and can be cured without losing voice.

As a result, there is a need for high spatial resolution and high sensitivity PET imaging in HNC. We are designing a high-resolution add-on dedicated HNC PET scanner to complement the whole-body PET scanner. This system will image the patients right after the whole-body PET scanning without injecting any extra dose to the patient. The dedicated system will provide extra information for scenarios when radiologists are looking for small lymph nodes or well-defined tumor boundaries. FIG. 1 shows an illustration of the add-on dedicated system [16], [17].

With more accurate detection of small lymph nodes and estimation of the extent of tumor growth, the dedicated head and neck will provide physicians with more freedom to choose treatment options including surgery, radiation therapy, and chemotherapy. It helps with more accurate radiation dose planning and will lead to better patient outcomes such as preserving organ functions. It also improves confidence in differentiating post-treatment changes from tumor recurrence.

To design such a dedicated system, the first consideration is the detecting material. Cadmium zinc telluride (CZT) detectors and lutetium-yttrium oxyorthosilicate (LYSO) combined with silicon photomultiplier (SiPM) detectors are investigated for HNC PET scanner. Compared with LYSO, CZT has a high energy resolution, sub-millimeter intrinsic spatial resolution [18], [19] and intrinsically achievable depth-of-interaction (DOI) information. It has been used for the small field of view (FOV) and high-resolution PET applications such as small animal imaging [20]–[23]. The high energy resolution and high spatial resolution of CZT detectors are also important for correctly identifying the first interaction position in multiple interaction photon events (MIPEs) using Compton kinematics.

Another important consideration for designing a dedicated PET is the system geometry. Compared with a whole-body PET scanner, dedicated PET often utilizes smaller and compact geometry to adapt to the dedicated imaging environment and improve the system sensitivity [24]. For example, dedicated brain PET has ring and helmet geometries [25]–[28], and dedicated breast PET has two-panel and ring geometries [29]–[32]. In this work, we consider two-panel, lengthened two-panel, four-panel, and full-ring geometries for the sake of high sensitivity and patient's comfort.

The purpose of this study is to evaluate and optimize system design among different detecting materials and geometries. For detecting material, two scanners with the same two-panel geometry based on CZT and LYSO are compared in terms of photon coincidence sensitivity, MIPEs recoverability, noise equivalent count (NEC) rate, and spatial resolution. For system geometry, CZT scanners with two-panel, lengthened two-panel, four-panel, and full-ring geometries are compared in terms of contrast recovery coefficient (CRC) and contrast-to-noise ratio (CNR) of reconstructed images.

II. Method and materials

A. Detecting material

Two systems with the same stationary two-panel geometry based on CZT and LYSO were built in GATE [33]. Based on the human head size [34], the panel was set as 150×200 mm^2, and the distance between the two panels was 200 mm. For the LYSO system, the LYSO crystal segment size was 1×1 mm^2 and the crystal thickness was 20 mm. The fill factor was 86.5%. The energy resolution, time resolution, and DOI resolution were assumed to be 15%, 400 ps, and 2 mm, respectively [35]–[37]. The time window and energy window for selecting coincidence events were 1 ns and [400, 620] keV, respectively. For the CZT system, the crystal size was 40×40×5 mm^3, and the energy resolution, time resolution, and intrinsic spatial resolution were set to 2%, 8 ns, and 1×1×1 mm^3, respectively [19]. The time window was 15 ns and energy window was [490, 530] keV. The deadtime of the LYSO scanner and CZT scanner were set as 1 μs and 10 μs, based on PETsys TOFPET2 ASIC [38] and Kromek RENA3 ASIC [39], respectively.

The photon coincidence sensitivity was defined as the coincidence rate divided by the point source activity. Generally, only photons without scattering were used to constitute coincidences (P-P coincidence). To further improve sensitivity, MIPEs have been recovered with different methods [40]–[43]. In this paper, Compton kinematics [40] was used to recover coincidences (P-CP coincidence) between annihilation photons that had a photoelectric event (P photon) and photons that had a Compton event before the photoelectric event (CP photon). In Compton kinematics, the scattering angle can be computed by energy,

$$\theta_E = \cos^{-1}\left(1 - m_0c^2\left(\frac{1}{E_s} - \frac{1}{E_i}\right)\right), \quad (1)$$

where E_i is the incident photon energy, E_s is the scattered photon energy, and m_0c^2 is the rest mass of an electron. For PET applications, $E_i = m_0c^2 = 511\ keV$. The scattering angle can also be computed using the interaction position information,

$$\theta_p = \cos^{-1}\left(\frac{\overrightarrow{V}_i \cdot \overrightarrow{V}_s}{\left|\overrightarrow{V}_i\right| \cdot \left|\overrightarrow{V}_s\right|}\right), \quad (2)$$

where $\overrightarrow{V}_i$ and $\overrightarrow{V}_s$ are the directions of incident and scattered photons. FIG. 2 shows the principle to identify the Compton event in a P-CP coincidence. To compare the sensitivity and MIPE recoverability of the two systems and check the performance of Compton kinematics under a high single rate, 10 point sources with different activities (7, 14, …, 70 *MBq*) placed at the FOV center were simulated separately. Since the crystal size of many previously published dedicated brain PET scanners are around 2 *mm* [25], [27], [44]–[46], we compared the Compton recovery accuracy of systems based on 1-*mm* crystal size and 2-*mm* crystal size respectively.

NEC rate incorporates the noise effects of random and scatter counts and is an indicator of the signal-to-noise ratio for PET systems [47].

$$NEC = \frac{T^2}{T + S + R}, \quad (3)$$

where T, S, and R are true, random and scatter coincidence rates respectively. To consider the effect of activity outside of the FOV, the phantom used for the NEC study contained three components, which represented the brain, neck, and torso. The brain component was a 130-*mm* diameter and 80-*mm* height cylinder with a 45.6 kBq/cm^3 concentration activity [48], and the torso component was a 260-*mm* diameter and 200-*mm* height cylinder with 5.7 kBq/cm^3. Both the brain and the torso were placed outside the FOV. For the neck component [14] (cylinder, 110-*mm* diameter, 126-*mm* height), 10 different concentration activity (5.7, 11.4, …, 57 kBq/cm^3) were investigated.

For a stationary two-panel geometry, the incomplete angular sampling would cause the orthogonal-panel spatial resolution to be worse than the in-panel spatial resolution. To understand the difference, a point source was placed at the FOV center, and the line profiles along the orthogonal-panel and in-panel directions in the reconstructed image were fitted separately to measure the spatial resolution. Images were reconstructed with a list-mode 3D maximum likelihood expectation maximization (MLEM) algorithm [49] through the gpurecon program [50]. Time-of-flight (TOF) is known to have the large potential for image quality improvement and more accurate quantification (signal-to-noise ratio) for a given number of counts [51], [52], and TOF was incorporated in the image reconstruction of the LYSO system. As a comparison, the CZT system did not utilize TOF.

B. System geometry

For the system geometry study, four CZT systems with two-panel, lengthened two-panel, four-panel, and full-ring geometries were compared, as is shown in FIG. 3. All four systems have 150-*mm* thickness in the *z* axis. The panel size of the two-panel and the four-panel systems was 40×200 mm^2, while the panel in the lengthened two-panel system was 40×320 mm^2. The average human head size was 145 *mm* (head breadth) × 194 *mm* (head thickness) in the United States, and the maximum head size was 174 *mm* × 239 *mm* [34]. Since the distance between detector panels was adjustable, the distance between panels was set as 24 *cm* based on the average human head size. On the contrary, the full-ring structure did not have the flexibility to adjust its geometry, so the inner diameter was set as 30 *cm* based on the maximum human head size. The time resolution and energy resolution were set as 8 *ns* and 2%, respectively. The time window was 15 *ns* and energy window was [490, 530] *keV*. P-CP coincidences were used in image reconstruction.

For the cylinder phantom study, four hot spheres with diameters 3, 4, 6 and 8 *mm* were placed in a 126-*mm* height, 110-*mm* diameter water phantom. The background concentration activity was 5.7 kBq/cm^3, and the hot-to-background ratio was 8:1 [14]. We further compared the four geometries based on 1-*mm* crystal size and 2-*mm* crystal size. Each system had a 2-*min* data acquisition.

The same MLEM gpurecon program without TOF was used for image reconstruction. For the 1-*mm* crystal size, the voxel size of reconstructed image was 0.5×0.5×0.5 mm^3. As a comparison, the voxel size of the 2-*mm* crystal size was 1×1×1 mm^3. The image reconstruction had 10 iterations, which was chosen to maximize the CNR of the 3-*mm* sphere in the cylinder phantom study. Data corrections for scatter coincidence and random coincidence were not applied. No regularization or a post-reconstruction filter was used.

Image quality was evaluated based on CRC and CNR [53],

$$CRC = \frac{C_{hot}/C_{bkg} - 1}{a_{hot}/a_{bkg} - 1},\ CNR = \frac{C_{hot}/C_{bkg}}{\sigma_{bkg}}, \tag{4}$$

where C_{hot} and C_{bkg} are the average voxel value in a hot sphere and background region of interest (ROI), respectively, a_{hot} and a_{bkg} are the ground-truth concentration activity, and σ_{bkg} is the standard deviation of the voxel values in the background ROI.

For the XCAT phantom [54] study, simulations of the lengthened two-panel system and the four-panel system on the XCAT phantom in head and neck region (200×240×150 mm^3 in *x*, *y*, and *z* axes) were performed. The ^{18}F-FDG concentration activity in normal tissue, spinal cord, salivary gland, and brain were 3.4, 7.1, 7.8, and 16.9 kBq/cm^3 based on clinical study, respectively [14], [55]–[57]. A sphere tumor with 5 *mm* diameter was put inside the phantom and the tumor concentration activity was 27.2 kBq/cm^3. Each system had a 6-*min* data acquisition [14].

C. Comparison with a whole-body scanner

To validate the benefits of the add-on dedicated system, another simulation was performed to compare the two-panel scanner with one whole-body PET scanner (GE Discovery MI 4-ring PET scanner). The same cylinder phantom (11-*cm* diameter and 12.6-*cm* length) were simulated with both systems, respectively. Hot spheres (nine 3-*mm* diameter, nine 4-*mm* diameter, five 6-*mm* diameter, five 8-*mm* diameter) were placed in the central slice of the axial direction of the phantom. The background activity was 5700 Bq/cm^3, and the sphere to background ratio was 8:1.

III. Results

A. Detecting material

The sensitivity of the CZT and LYSO systems under different source activities is shown in FIG. 4. At 7 *MBq*, the sensitivity of the CZT and LYSO systems are 0.60% and 0.69%, and it increased to 2.43% and 2.55% respectively after recovering MIPEs. The results showed that with a 40 *mm* crystal thickness, the CZT system could achieve similar sensitivity as the LYSO system. After recovering MIPEs, the sensitivity of both systems improved approximately 3 times, which indicated the importance of MIPE recovery. Due to the poor time resolution, the P-CP sensitivity of the CZT system decreased 25.5% with the increase of activity from 7 *MBq* to 70 *MBq*. This was because if more than two interactions were detected within the same time window, all the events within this time window were abandoned. So when the count rate got higher, it was more likely that more than two interactions were detected within the same time window, and lowered the sensitivity.

The P-CP coincidences recovery accuracy of the CZT and LYSO systems based on 1-*mm* and 2-*mm* crystal is shown in FIG. 5. The results also showed that given a source activity, the recovery accuracy of the CZT system was about 20% higher than LYSO, and the recovery accuracy was not affected by the source activity. By decreasing the crystal size from 2 mm to 1 mm, the Compton recovery accuracy of the CZT system improved 4.23%, while that of the LYSO system was almost the same. This was because the Compton recovery accuracy was affected by both energy resolution and crystal size. For the CZT system, the energy resolution was high, so decreasing crystal size could improve accuracy. However, in the LYSO system, the poor energy resolution dominated the recovery error, so changing crystal size did not make an obvious influence. Details can be found in our previous study [58].

The NEC, true, random, scatter, and the total rate of the CZT and LYSO systems are shown in FIG. 6. The LYSO system showed a high NEC rate at all concentration activities, and the higher the concentration activity, the larger the difference was.

The in-panel and orthogonal-panel spatial resolution of CZT and LYSO systems based on only P-P coincidences and both P-P and P-CP coincidences are shown in Table. I. The results indicated that CZT had a better spatial resolution than LYSO, and incorporating P-CP coincidences for image reconstruction would slightly deteriorate the spatial resolution.

B. System geometry

The reconstructed images of the cylinder phantom are shown in FIG. 7. The 4, 6, and 8-*mm* hot spheres were clearly resolvable in all four systems. Due to the limited angular data sampling, the background and hot spheres were elongated in the two-panel and lengthened two-panel systems. CRC and CNR versus hot sphere diameters curves are shown in FIG. 8.

The reconstructed images of the XCAT phantom are shown in FIG. 9. The spinal cord, salivary gland, and brain were clearly resolvable in both images. The tumor was also visible in both systems within 6 *min*, but the tumor in the four-panel system was more resolvable. The tumor CRC were 0.08 and 0.12 for the lengthened two-panel system and the four-panel system, respectively, while the CNR of the tumor were 2.4 and 4.2.

C. Comparison with a whole-body scanner

The transverse slices and the sagittal slice reconstructed images of the dedicated two-panel scanner and the dedicated scanner are shown in FIG. 10. It can be shown that the dedicated system can achieve superior spatial resolution than the whole body system, which indicates the benefits of using such an add-on system.

IV. Discussion

We studied the design considerations for a dedicated HNC PET scanner. Different detecting material including CZT and LYSO, and different system geometries including two-panel, lengthened two-panel, four-panel, and full-ring geometries were investigated.

A. Detecting material

For the detecting material, CZT and LYSO were compared in terms of sensitivity, MIPEs recovery, NEC rate, and spatial resolution. The sensitivity of CZT was 13.04% lower than the LYSO, which suggests that the CZT system and the LYSO system with exactly the same geometry can achieve comparable sensitivity. After recovering MIPEs, the sensitivity of CZT was only 4.71% lower, which indicates that the CZT could recover more MIPEs than LYSO. This is because CZT has a larger Compton-to-photoelectric ratio than the LYSO. Moreover, due to the much better energy resolution, CZT showed an about 20% higher recovery accuracy. The large Compton-to-photoelectric ratio and the high energy resolution make CZT better for the MIPEs recovery.

Though the LYSO system had a larger scatter event rate, it showed a high NEC rate due to its larger true event rate and smaller random event rate. In the data processing, if photons from more than one annihilation were detected in a time window, all singles in this window were rejected. When the total activity was high, the probability to detect more annihilation photons within one window got increased. Since CZT had a wide time window (15 *ns*), the true coincidence rate of CZT gradually plateaued. The poor time resolution also led to a much higher random rate for the CZT system compared to the LYSO system. The low true rate and high random rate caused the NEC rate for the CZT system to be lower than that of the LYSO system. A lead shield for stopping singles from outside of the FOV is likely necessary for the CZT system.

Table. I shows that the in-panel and orthogonal-panel spatial resolution of the CZT decreased 3.57% and 4.23% after incorporating P-CP coincidences. As a comparison, the LYSO system decreased by 12.90% and 6.87% respectively. The spatial resolution gets worse because incorrectly recovered P-CP coincidences were used for image reconstruction. However, since LYSO has a worse recovery accuracy, the deterioration of spatial resolution is worse than CZT.

Table. I also shows that the CZT system has a better in-panel and orthogonal-panel spatial resolution than LYSO. For a stationary two-panel geometry, DOI affects both the in-panel and orthogonal-panel spatial resolution [18], which is illustrated in FIG. 11. For the sake of simplicity, a 2D case is drawn. The spatial resolution is limited by the closest distinguishable LORs. For two adjacent LORs, the blurring along the orthogonal-panel axis σ_O is equal to DOI, while the blurring along the in-panel axis σ_I is equal to $DOI \times tan(\theta)$, where θ is the angle between the LOR and the orthogonal-panel axis. In a stationary two-panel geometry system, the LOR with a small θ has a large probability to be detected. As a result, σ_O tends to be larger than σ_I, which explains why the orthogonal-panel spatial resolution is worse than the in-panel spatial resolution. The result shows the importance of DOI resolution to a two-panel geometry system.

B. System geometry

For the system geometry, four CZT systems with two-panel, lengthened two-panel, four-panel, and full-ring geometries were compared in terms of CRC and CNR of the reconstructed image. For all four scanners, both CRC and CNR improved with increasing sphere diameter, and the difference in CRC and CNR among all scanners increased as the diameter of the sphere increased. For all four geometries, the CRC and CNR of 1-*mm* crystal size were higher than the 2-*mm* crystal size, which indicated the benefits of using 1-*mm* crystal.

Compared with the two-panel scanner, the lengthened two-panel scanner had a higher CRC and CNR in 4-, 6- and 8-*mm* spheres. Compared with the two limited-angle (two-panel, lengthened two-panel) scanners, the two full-geometry (four-panel, full-ring) scanners could achieve higher CRC and CNR in all spheres. However, the improvement of CRC and CNR comes with the need for more detectors. Specifically, the ratio of the number of detectors of the two-panel, lengthened two-panel, four-panel, and full-ring geometries are 10:16:20:23, which means that the full-geometry (four-panel, full-ring) designs achieve higher CRC and CNR in large lesions with a doubled cost, due to the doubled number of detectors.

Moreover, panel-based designs have the flexibility to adjust the panel distance while the same image reconstruction method can still be used, so that the system geometry is able to be compactly adapted to individual patients. It is also very important to consider the patient's comfort in such a compact design, and two-panel can achieve this goal without blocking the line of sight.

C. Comparison with virtual-pinhole PET

Virtual-pinhole PET is known as using a high-resolution add-on PET scanner to improve the spatial resolution. Depends on different regions of interest, virtual-pinhole PET can have different geometries (full-ring for small animal imaging [59], half-ring for head and neck imaging [60] and breast imaging [61].) The idea of the virtual-pinhole PET and the dedicated PET proposed in this work are similar, both of which use smaller crystals and are placed near the region of interest to improve the spatial resolution locally.

However, the main difference is that the virtual-pinhole PET is inserted into the whole-body PET and it acquires data with the whole body PET simultaneously (a task that is not trivial and requires working with whole-body vendors), thus there are three types of coincidence events: insert-insert, insert-scanner, and scanner-scanner [62]. As a result, the insertion of the virtual-pinhole PET affects the photon detection of the whole-body PET and image reconstruction is more complex because two systems need to be modeled at the same time. As a comparison, the dedicated PET acquires data with whole-body PET separately, so photon detection of the whole-body PET is not affected. However, since we want to use the whole-body PET image as the prior image to reduce the limited artifacts in the dedicated PET image, it is necessary to track patient movement for the dedicated PET application, which brings extra complexity.

D. Limited-angle artifacts

The stationary two-panel PET system is known to have the limited-angle artifacts [63]–[67]. FIG. 7 shows that both the two-panel and lengthened two-panel systems have an elongated background and hot spheres. One method to reduce the limited-angle artifacts is to use a prior image without limited angle artifacts and penalize the dissimilarity between the target image and the prior image during the reconstruction [68], [69]. We plan to take advantage of a whole-body PET scan, which does not have the problem of limited-angle artifacts and the whole-body PET image can be used as the prior image. We are developing the penalized maximum-likelihood image reconstruction algorithm.

V. Conclusions

In this paper, we evaluate different detecting materials and system geometries through the simulation of a dedicated HNC PET scanner. For comparing the detecting material, the sensitivity of the 40-*mm* thickness CZT system and the 20-*mm* LYSO system are 0.60% and 0.69%, and it increases to 2.43% and 2.55% respectively after recovering MIPEs. The favorable energy resolution makes the CZT system have an approximately 20% higher MIPE recovery accuracy, but the poor time resolution of CZT results in a two-times lower NEC rate. However, the superior DOI resolution of CZT leads to a better spatial resolution, which is important for HNC imaging. For comparing system geometry, the four-panel and full-ring PET systems achieve higher CRC and CNR than the two-panel and lengthened two-panel PET systems. Nevertheless, the CRC of the 3-*mm* diameter hot sphere of the two-panel and lengthened two-panel PET systems is comparable to that of the four-panel and full-ring PET systems. The CRC and CNR for the two-panel system can be improved by extending the panel size. Both the two-panel and four-panel PET systems can image the patient's head and neck region and resolve a 5-*mm* lesion within 6 *min*. The disadvantage of the four-panel and full-ring PET systems is that they are in the line of sight of the patient, which compromises the patient's comfort. To summarize, the CZT system can achieve better spatial resolution and recovery accuracy of MIPEs compared to the LYSO system. However, the poor timing resolution of the CZT system yields a lower sensitivity, especially at high source activity. For system geometry, full-ring and four-panel designs have better CRC and CNR than two-panel design, but they come with a higher cost and compromised patient's comfort.

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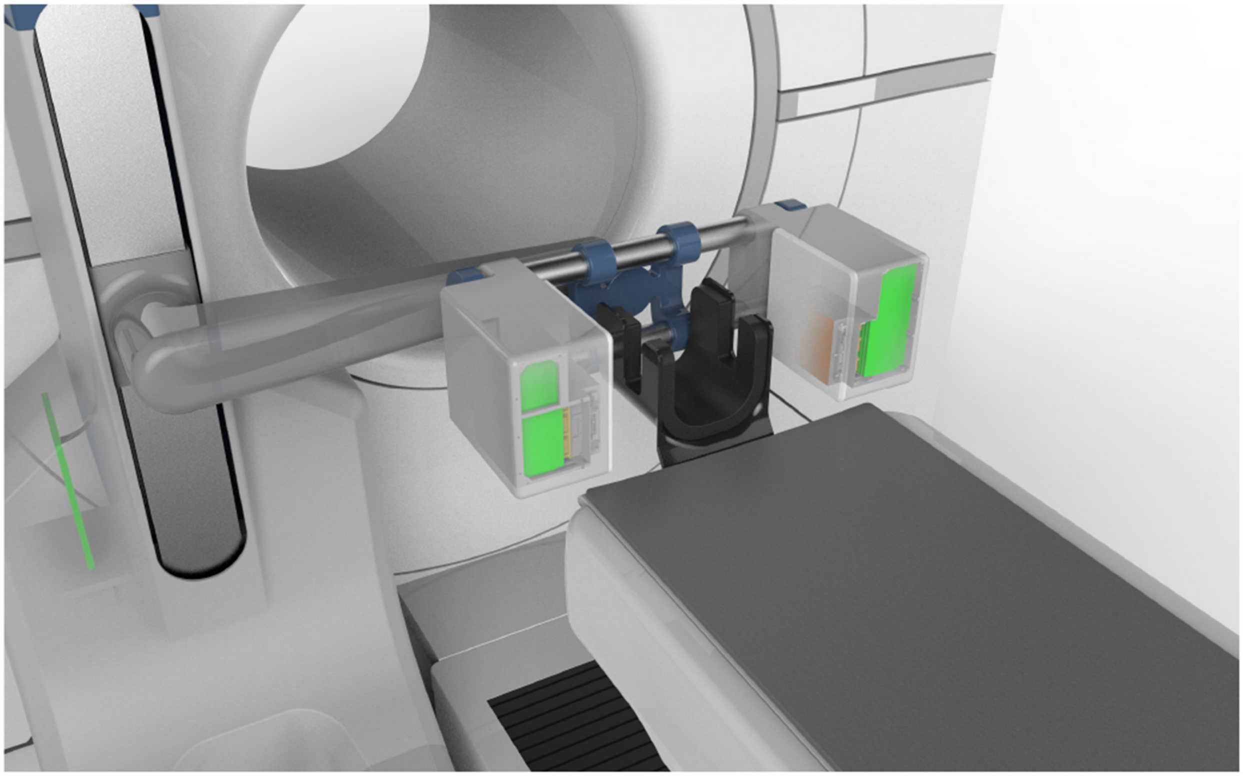

Fig. 1.

Illustration of an add-on dedicated head and neck cancer PET scanner integrated into the standard whole-body PET/CT imaging workflow. The gantry is implemented to interface with the patient bed to image the patient right after the whole-body PET scanning.

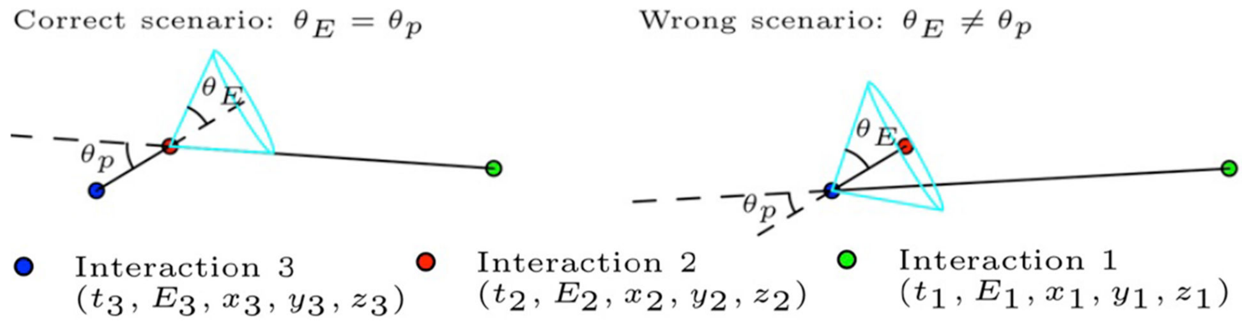

Fig. 2.
Using Compton kinematics to identify the Compton event in a P-CP coincidence. The sequence with a smaller $|\theta_E - \theta_p|$ is picked up.

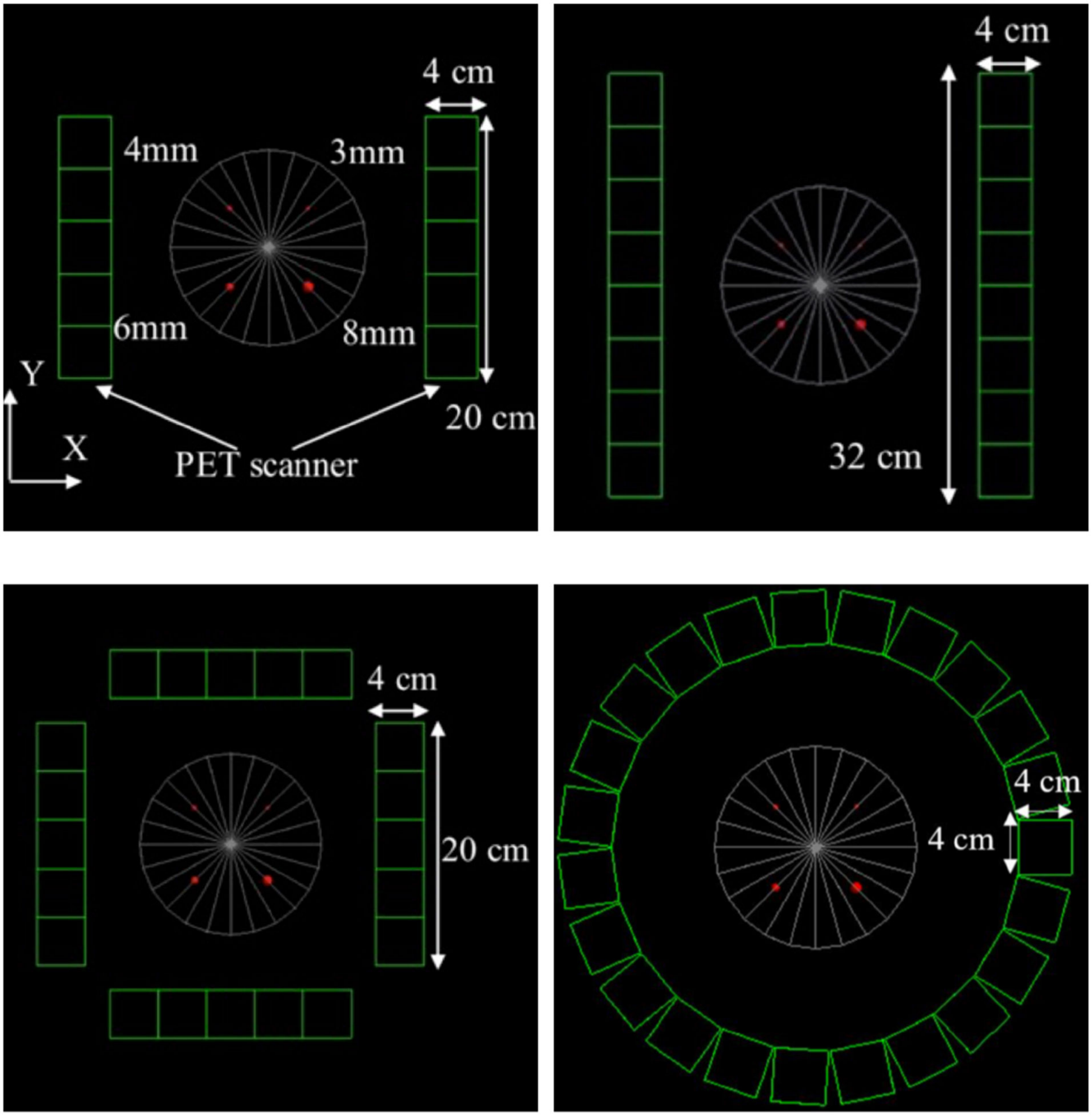

Fig. 3.

Four CZT systems with two-panel (top-left), lengthened two-panel (top-right), four-panel (bottom-left), and full-ring (bottom-right) geometries were used for the phantom study. The x and y axes are shown in the figure, and the z axis is perpendicular to the paper. All four systems have the same 150-mm thickness in the z axis.

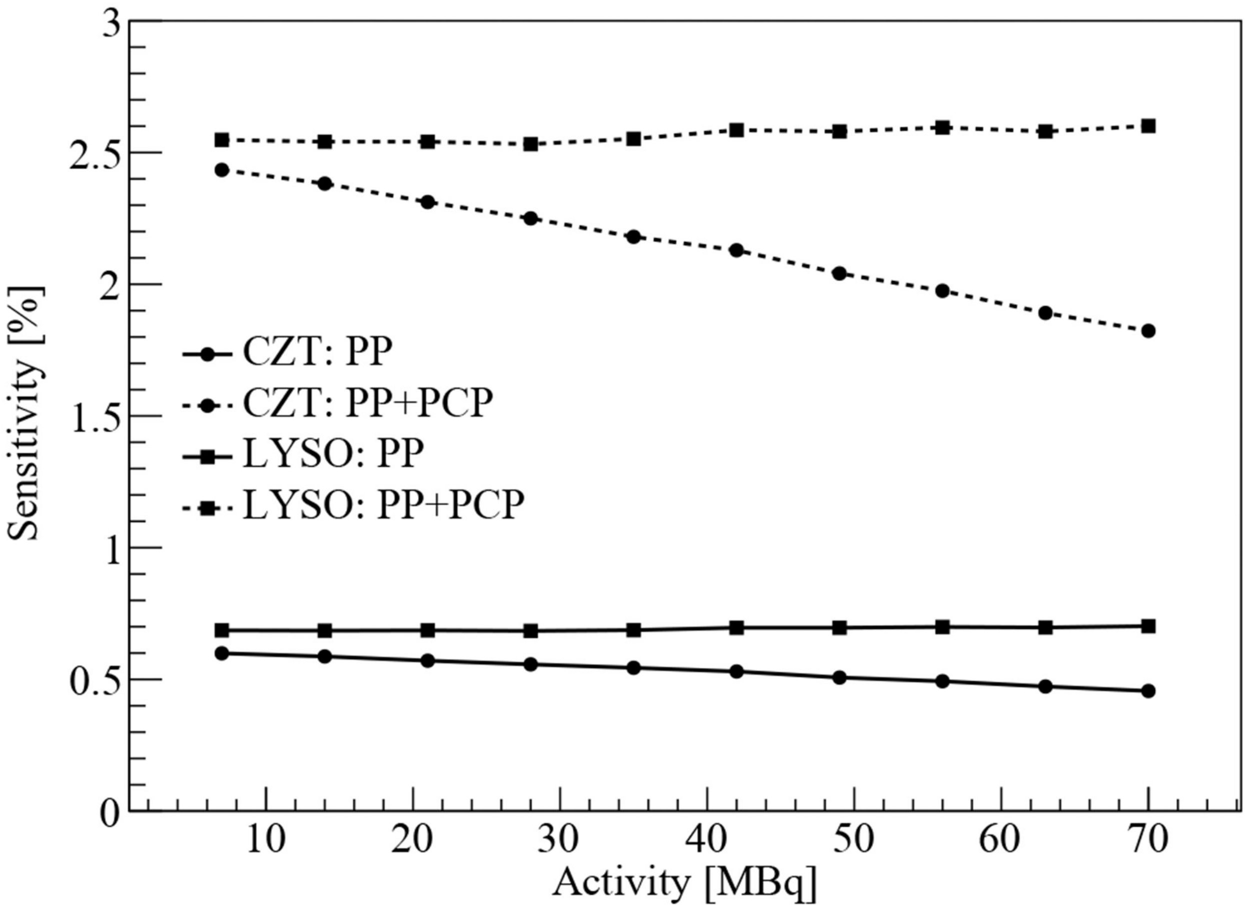

Fig. 4.
The sensitivity w/o and w/Compton recovery of the CZT and LYSO systems under different source activities. The crystal size of both system is 1 *mm*.

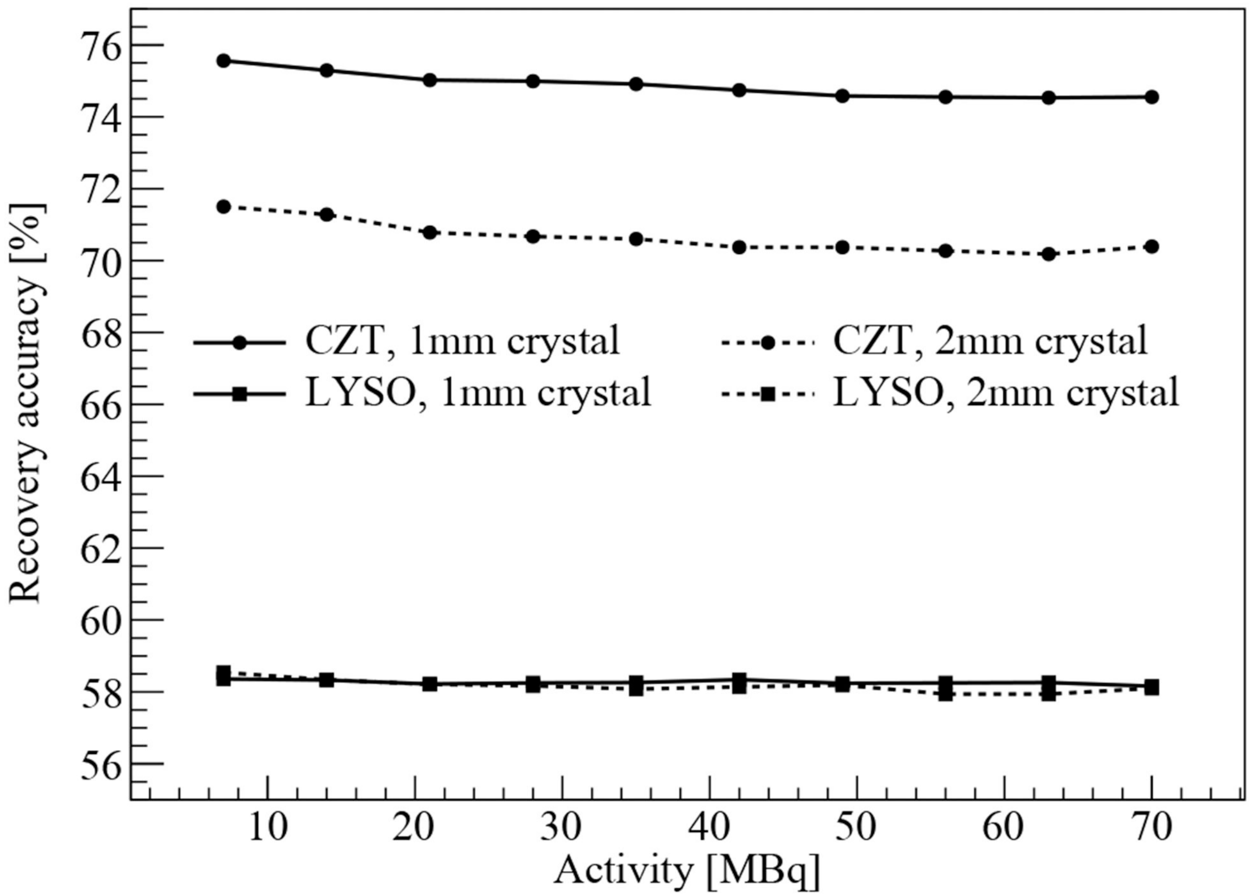

Fig. 5.
The Compton recovery accuracy of the CZT and LYSO systems based on 1-*mm* and 2-*mm* crystal size respectively.

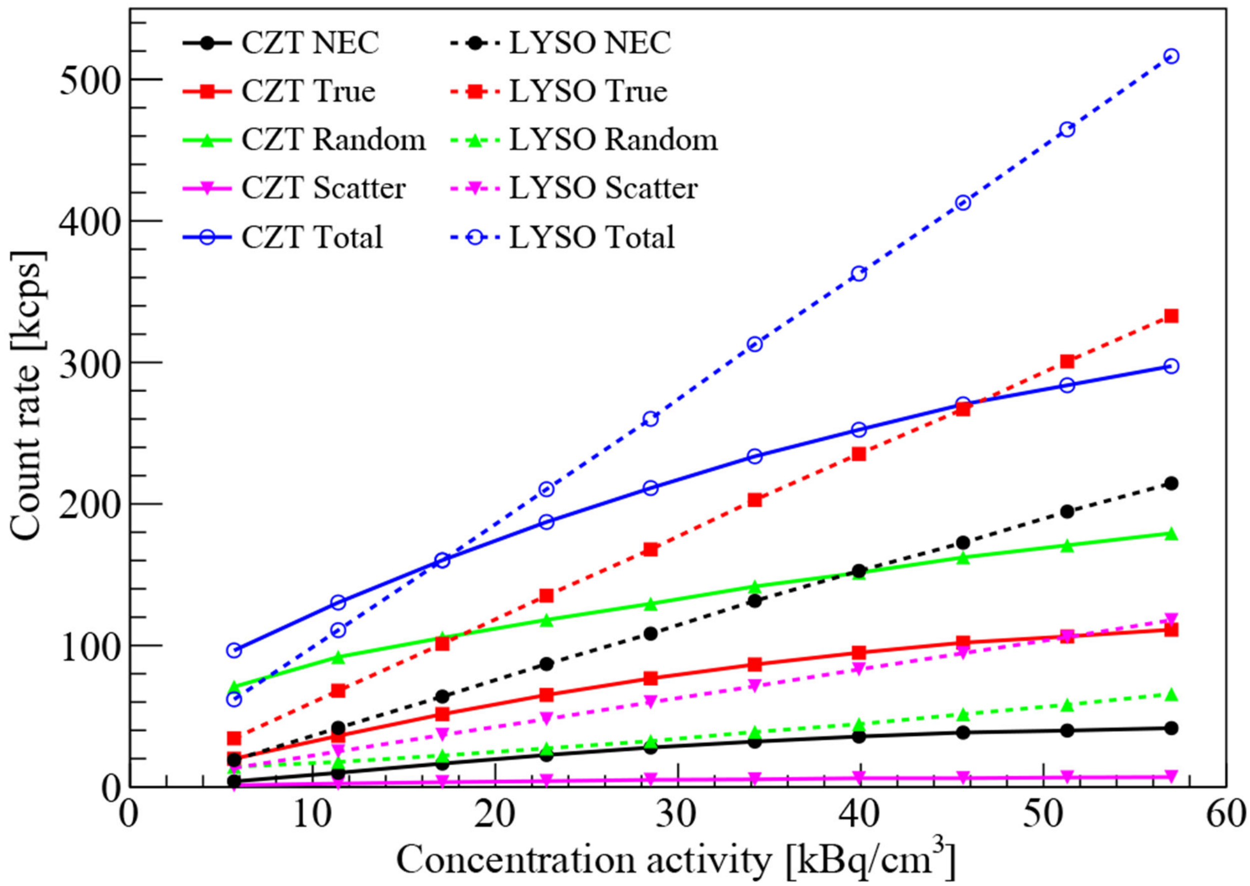

Fig. 6.

The NEC, true, random, scatter, and the total rate of the CZT and LYSO systems. The concentration activity only refers to the neck phantom, while that of the brain and the torso phantoms are kept as the same during the sweeping.

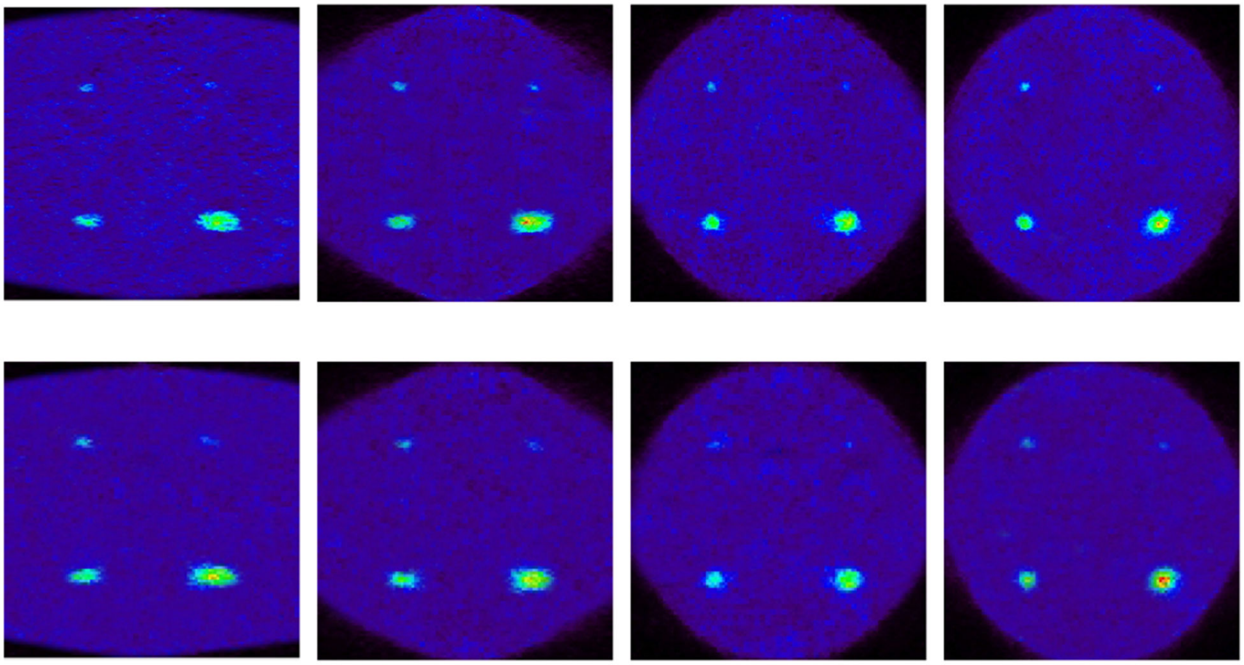

Fig. 7.

Reconstructed images of the cylinder phantom with hot spheres. The top row is the results based on 1-*mm* crystal size, and the voxel size of the reconstructed image is 0.5×0.5×0.5 mm^3. The bottom row is the results based 2-*mm* crystal size, and the voxel size is 1×1×1 mm^3. From left to right, the four systems are two-panel, lengthened two-panel, four-panel, and full-ring, respectively.

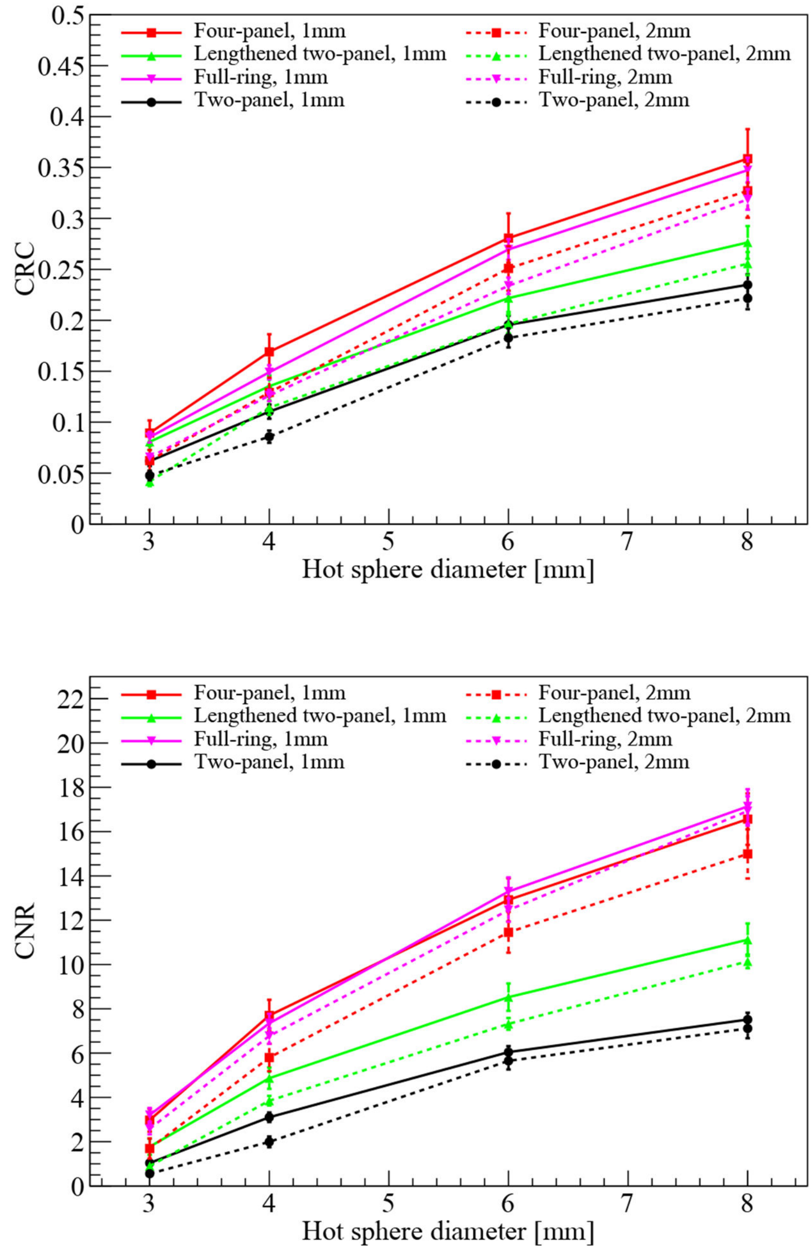

Fig. 8.

The contrast recovery coefficient (top) and contrast-to-noise ratio (bottom) versus hot sphere diameter for different system geometries and different crystal size.

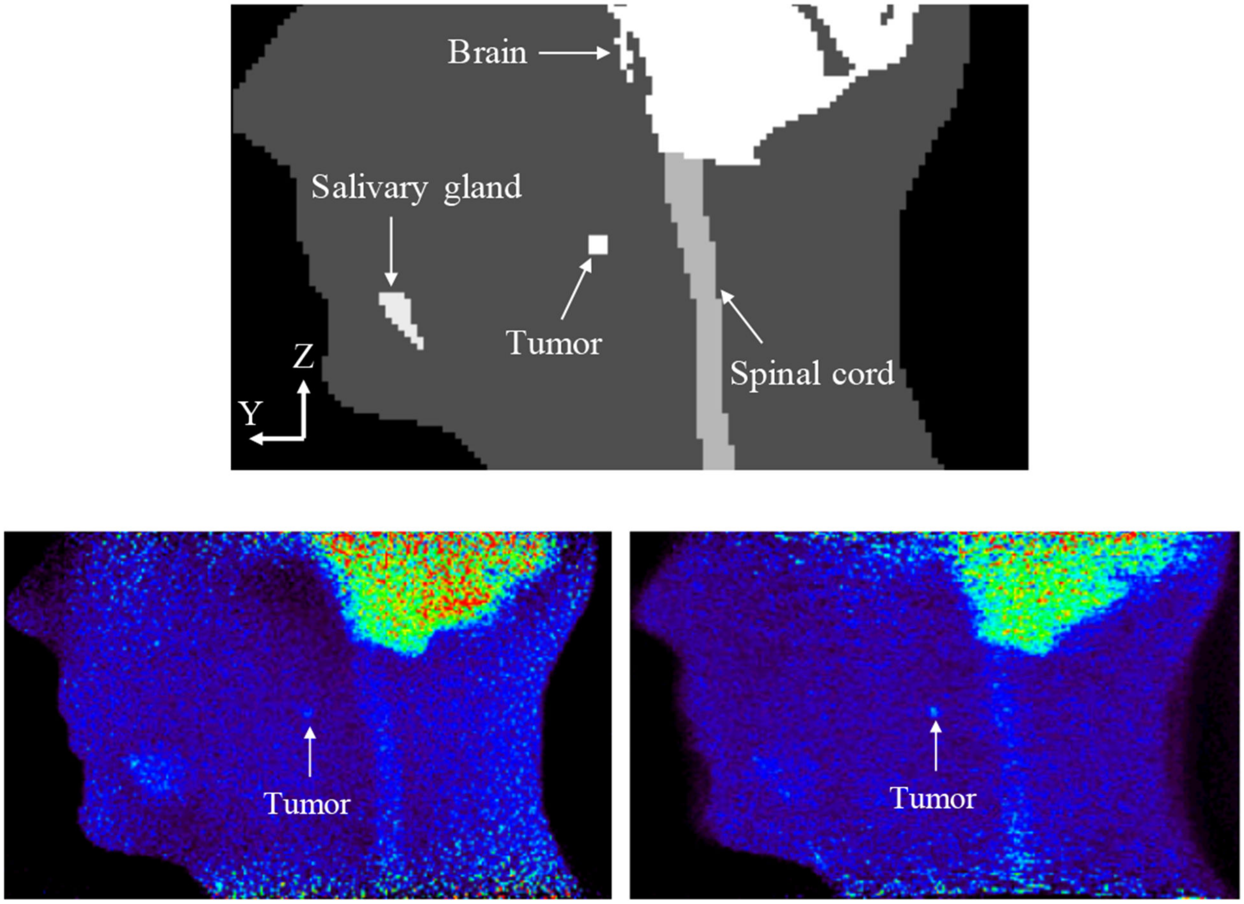

Fig. 9.
Reconstructed images of the XCAT phantom in head and neck region. Top: concentration activity map. Bottom left: the lengthened two-panel system. Bottom right: the four-panel system.

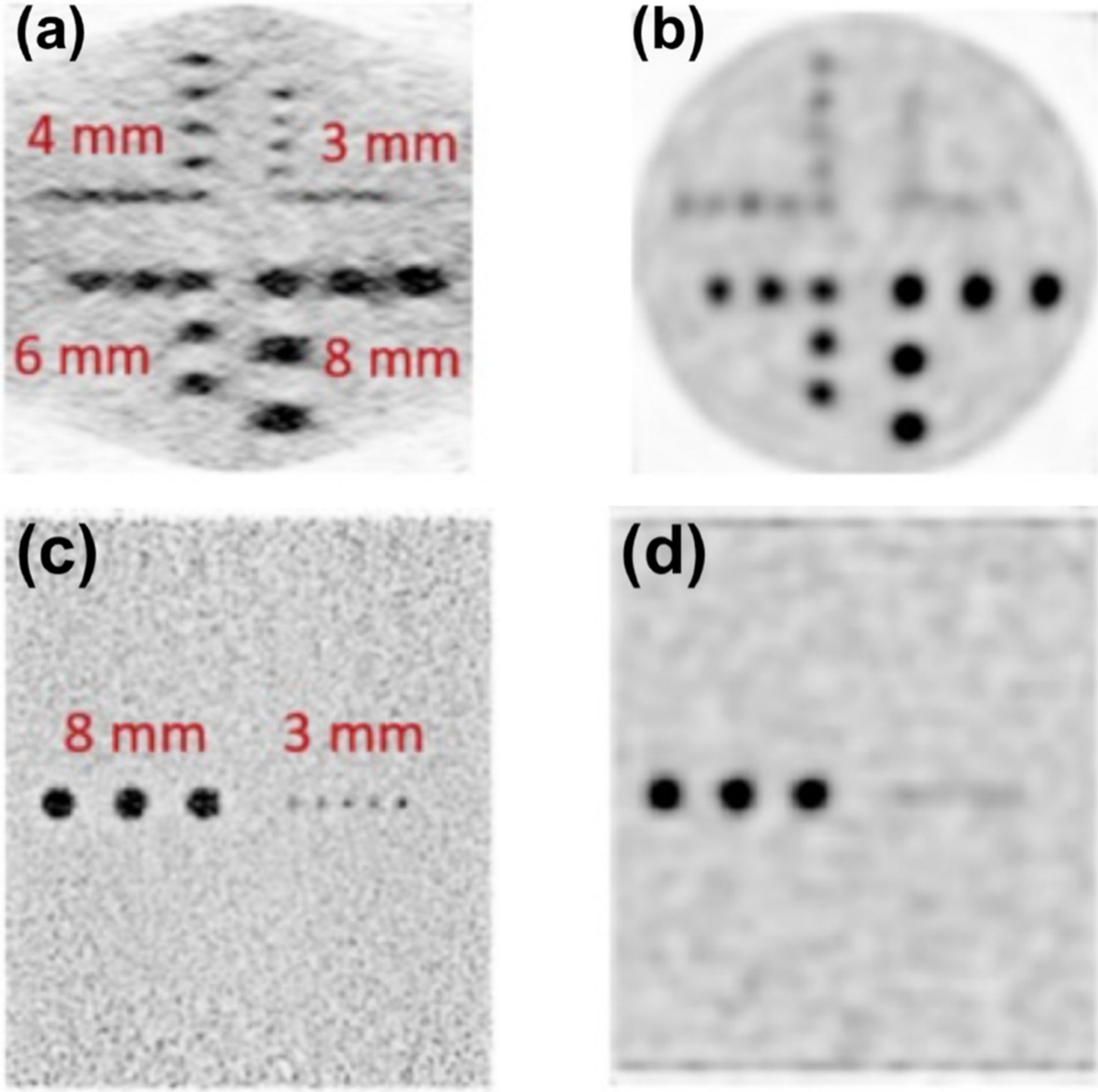

Fig. 10.

Comparison of the two-panel dedicated system and a whole-body scanner (GE Discovery MI 4-ring PET scanner). (a). the transverse slice of the dedicated scanner; (b). the transverse slice of the whole body scanner. (c). the sagittal slice of the dedicated scanner; (d). the sagittal slice of the whole body scanner.

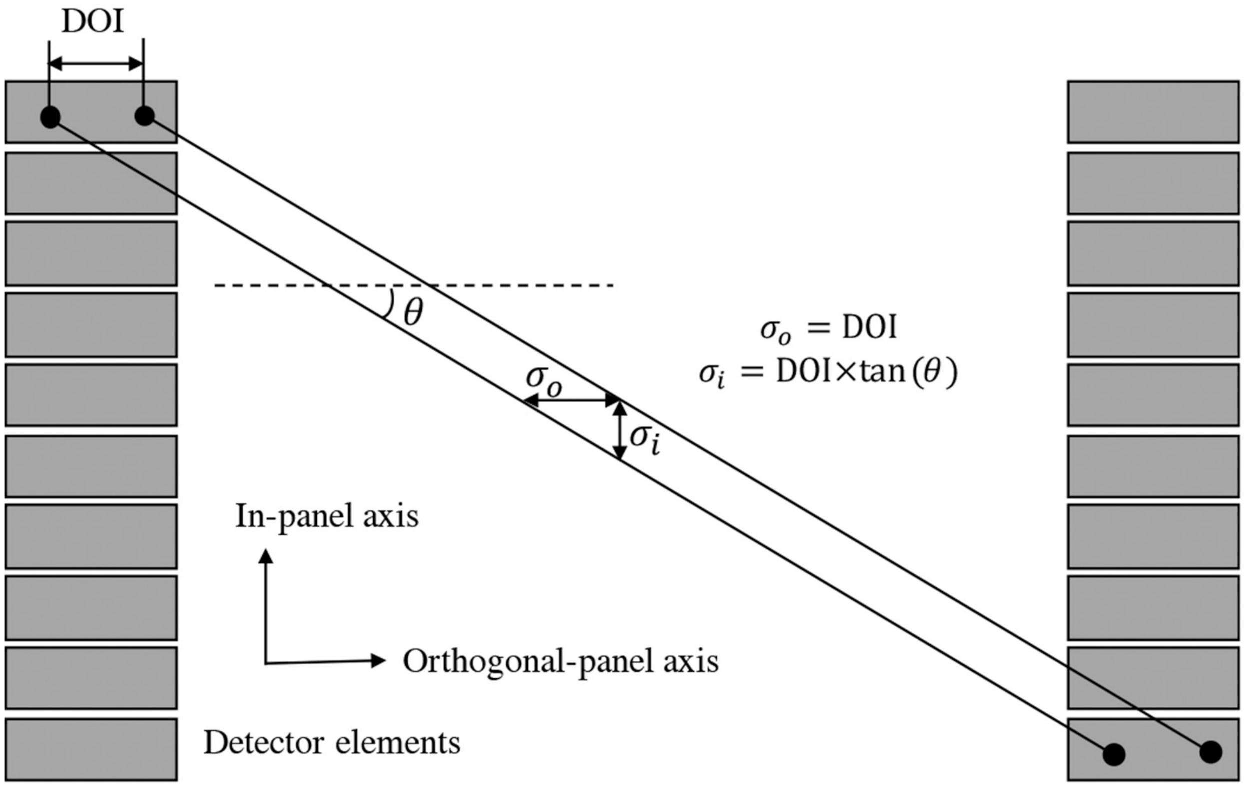

Fig. 11.
Illustration of how DOI affects the in-panel and orthogonal-panel spatial resolution in a stationary two-panel geometry PET system.

TABLE I

Point source spatial resolution of the CZT and LYSO system.

spatial resolution	P-P coincidences	P-P and P-CP coincidences
CZT in-panel	0.56 *mm*	0.58 *mm*
CZT orthogonal-panel	0.71 *mm*	0.74 *mm*
LYSO in-panel	0.62 *mm*	0.70 *mm*
LYSO orthogonal-panel	1.31 *mm*	1.40 *mm*