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Augmented Whole-body Scanning via Magnifying PET

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Abstract

A novel technique, called augmented whole-body scanning via magnifying PET (AWSM-PET), that improves the sensitivity and lesion detectability of a PET scanner for whole-body imaging is proposed and evaluated. A Siemens Biograph Vision PET/CT scanner equipped with one or two high-resolution panel-detectors was simulated to study the effectiveness of AWSM-PET

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technology. The detector panels are located immediately outside the scanner's axial field-of-view (FOV). A detector panel contains 2×8 detector modules each consisting of 32×64 LSO crystals ($1.0 \times 1.0 \times 10.0$ mm³ each). A ²²Na point source was stepped across the scanner's FOV axially to measure sensitivity profiles at different locations. An elliptical torso phantom containing 7×9 spherical lesions was imaged at different axial locations to mimic a multi-bed-position whole-body imaging protocol. Receiver operating characteristic (ROC) curves were analyzed to evaluate the improvement in lesion detectability by the AWSM-PET technology. Experimental validation was conducted using an existing flat-panel detector integrated with a Siemens Biograph 40 PET/CT scanner to image a torso phantom containing spherical lesions with diameters ranging from 3.3 to 11.4 mm. The contrast-recovery-coefficient (CRC) of the lesions was evaluated for the scanner with or without the AWSM-PET technology. Monte Carlo simulation shows 36%–42% improvement in system sensitivity by a dual-panel AWSM-PET device. The area under the ROC curve is 0.962 by a native scanner for the detection of 4 mm diameter lesions with 5:1 tumor-to-background activity concentration. It was improved to 0.977 and 0.991 with a single- and dual-panel AWSM-PET system, respectively. Experimental studies showed that the average CRC of 3.3 mm and 4.3 mm diameter tumors were improved from 2.8% and 4.2% to 7.9% and 11.0%, respectively, by a single-panel AWSM-PET device. With a high-sensitivity dual-panel device, the corresponding CRC can be further improved to 11.0% and 15.9%, respectively. The principle of the AWSM-PET technology has been developed and validated. Enhanced system sensitivity, CRC and tumor detectability were demonstrated by Monte Carlo simulations and imaging experiments. This technology may offer a cost-effective path to realize high-resolution whole-body PET imaging clinically.

Index Terms—

AWSM-PET; Cancer detection; Virtual-Pinhole PET

I. Introduction

POSITRON emission tomography (PET) with ¹⁸F-fluorodeoxyglucose (FDG) provides *in vivo* measurement of glucose metabolism in patient's body [1]. With increased aerobic glycolysis (Warburg effect [2]) in many types of cancer cells, FDG-PET/computed tomography (CT) is widely used for cancer staging, re-staging, and for monitoring response to therapy. While the diagnostic accuracy of FDG-PET/CT for metastatic cancer detection is superior to other imaging modalities [3–5], the detection of metastatic lesions that are smaller than 1 cm remains a challenge. For example, the sensitivity of FDG-PET/CT is only around 50% for the detection of metastatic pelvic lymph nodes in invasive bladder cancer patients [6, 7] and early-stage cervical cancer patients [8].

The suboptimal performance of FDG-PET/CT for detecting small lesions is partly due to the limited spatial resolution of clinical scanners but more importantly the poor signal-to-noise ratio of the data when using whole-body imaging protocols. To find distant metastases at unknown locations, a patient is often scanned from the base of skull to the thighs in multiple steps of 1–3 minutes each. With short scan times, the quality of PET images is predominantly limited by counting statistics rather than the intrinsic spatial resolution of a

scanner. As a result, many recent innovations in clinical whole-body PET/CT scanners focused on enhancing the noise property of images (e.g., time-of-flight (TOF) PET [9, 10] and total-body PET [11]).

In a recent study, Surti et. al. [12] analyzed the effect of improved detector spatial resolution versus timing resolution on lesion detectability under whole-body imaging conditions. The results demonstrated improvement in lesion detectability by detectors of either higher intrinsic spatial resolution or higher timing resolution despite the finding that the image quality of PET is significantly limited by counting statistics when using whole-body imaging protocols. This finding has prompted efforts to further improve the spatial resolution and timing performance of clinical scanners [13].

Parallel to the efforts from clinical PET/CT manufacturers, several groups developed organ-specific PET (or PET/CT) systems that place high-resolution detectors around an organ-of-interest to maximize the image resolution and system sensitivity [14–21]. However, these devices are incompatible with clinical utility of PET that focuses on whole-body imaging and cancer staging. Approaches that bring the benefits of organ-specific PET to a clinical PET/CT scanner have emerged in recent years and been independently validated [22–25], including multi-resolution PET [26, 27], zoom-in PET [28], and virtual-pinhole PET (VP-PET) [23]. We previously developed several prototype VP-PET insert systems and demonstrated its ability to improve the image resolution of a clinical PET/CT scanner without compromising its whole-body imaging capability [29, 30]. Although the VP-PET insert technique enables zoom-in imaging for an organ-of-interest, it does not directly benefit whole-body cancer staging where the location of a distant metastasis, and thus the “target” region to zoom-in on, is unknown.

To overcome this limitation, a novel embodiment of the VP-PET technology optimized for clinical whole-body PET imaging was proposed. Here, the high-resolution detectors are placed outside of a scanner’s imaging FOV along the axial direction as an “Outsert” device (as will be illustrated in Fig. 1). This new geometry can significantly improve the image resolution and sensitivity of a clinical PET/CT scanner. Most importantly, the device simultaneously acquires high-resolution PET data as a patient undergoes a standard whole-body imaging protocol without incurring any additional scan time. Thus, the augmented whole-body scanning via magnifying PET (AWSM-PET) technology is fully compatible with current clinical workflows, with potential improvement in lesion detectability.

II. Materials and Methods

A. System Design

An AWSM-PET system consists of one or two high-resolution detectors (the “Outsert”) placed outside of the scanner’s axial FOV (Fig. 1). Coincidence detection between the Outsert and the scanner detectors, as well as between Outsert detectors (if there are two or more panels), needs to be established. When a patient undergoes a whole-body scan (whether through step-and-shoot or continuous bed motion [31, 32] acquisition), annihilation gamma-rays are simultaneously detected by different detector groups (Scanner-Scanner - SS, Outsert-Scanner - OS, Outsert-Outsert - OO). As tissues are moved from the

native scanner's imaging FOV into the "augmented imaging zone" enabled by Outserts (represented by the shaded areas in Fig. 1), additional coincidence events are collected. This extended axial FOV results in a higher overall system sensitivity than the native scanner when all coincidence events are used jointly for image reconstruction. Since the Outsert detectors have high intrinsic spatial resolution and are placed closer to a patient's body, the AWSM-PET systems also benefit from the resolution enhancement from VP-PET geometry [23].

B. Monte Carlo Simulation

The sensitivity and lesion detectability of a PET scanner equipped with AWSM-PET technology was evaluated using GATE [33]. Fig. 2(a) and 2(b) show the simulated system geometries that contain either one or two Outserts integrated with a Siemens Biograph Vision PET/CT scanner. The scanner consists of 304 detector modules arranged in 8 rings (422 mm in radius). Each module is composed of a lutetium-oxorthosilicate (LSO) array of 20×10 crystals, each measuring $3.2 \times 3.2 \times 20.0$ mm³. The axial field-of-view (FOV) of the scanner is 263 mm. The native scanner has 19 Detector Electronics Assemblies (DEAs), each processing signals from 16 detector modules (2/ring $\times$ 8 rings). An Outsert consists of 2×8 high-resolution detector modules. Each module is composed of 32×64 LSO crystals, each measuring $1.0 \times 1.0 \times 10.0$ mm³. The overall dimension of the LSO arrays in an Outsert is $129 \times 263 \times 10$ mm³ which has the same active sensing area as that of a standard DEA in the Biograph Vision scanner. The center of the Outsert in Fig. 2(a) was at (0, -195, 196) mm (assuming the origin of the coordinate system is at the center of the scanner's imaging FOV) such that the Outsert is just outside the axial FOV of the scanner. In Fig. 2(b), the center of the two Outserts were at (0, ± 195 , 196) mm. The coincidence resolving time (CRT) of the scanner and the Outsert detectors was assumed to be 250 ps. The energy window used was from 435 keV to 650 keV. The energy resolution of the scanner and the Outsert detectors was 12% at 511 keV (FWHM). Both object and detector scatter were included. The default setting for event positioning for block detectors in GATE was applied.

The sensitivities of the two AWSM-PET systems were analyzed by stepping a ²²Na point source across the axial FOV of the systems at different locations ($Y = -100, 0$, or 100 mm). The total true coincidence events detected by the scanner and the Outserts were divided into three groups (SS, OS, and OO) and normalized to the number of decays from the ²²Na source. The sensitivity profiles were plotted against the source location across the axial FOV. It should be noted that the profile of the SS events represents the sensitivity of the native scanner.

An elliptical torso phantom was imaged using the two AWSM-PET systems in Fig. 2. The torso phantom measures $316 \times 228 \times 400$ mm³ and contains sixty-three spherical lesions (as illustrated in Fig. 2(c)), each 4 mm in diameter and centered axially. The torso phantom and the lesions were filled with ¹⁸F solutions with initial activity concentrations of 5.3 kBq/mL and 26.5 kBq/mL, respectively, to obtain a tumor-to-background ratio of 5. The phantom was imaged at two bed positions where the lesions were located at $Z = 0$ mm and $Z = 180$ mm, respectively. Coincidence events were acquired for 200 s at each bed position.

C. Experimental Verification of the Concept using existing flat-panel detector

Using an existing flat-panel detector [34], a proof-of-concept AWSM-PET prototype was set up, as shown in Fig. 3(a). The flat-panel detector consists of 32 detector modules arranged in a 4×8 array, each contains a 16×16 LYSO crystal array read out by a 4×4 multi-pixel photon-counters (Hamamatsu S12641-PA-50(x)) array. Each LYSO crystal measures 0.92×0.92×3 mm³ in 1 mm pitches. The overall dimension of the sensitive area of the panel is 135×67 mm². A robotic arm was used to place the flat-panel detector around a Siemens Biograph 40 PET/CT scanner [35]. The axial FOV of the scanner is 220 mm. The integration of the flat-panel detector with the scanner, including readout electronics and data acquisition were described and can be found in [30].

A phantom study was carried out to measure the contrast-recovery-coefficient (CRC) of small lesions using the prototype AWSM-PET device. Spherical glass tumors with diameters of 3.3, 4.3, 6.0, 8.0, 9.6, and 11.4 mm (Fig. 3(b)) were mounted below the lung inserts inside a torso phantom which measures 316×228×162 mm³ (Fig. 3(a)).

The torso phantom and the lesions were filled with ⁶⁴Cu solutions of an initial activity concentration (¹⁸F equivalent) of 3.47 kBq/mL and 20.82 kBq/mL, respectively, to obtain a tumor-to-background ratio of 6. The reason that ⁶⁴Cu was chosen over ¹⁸F is due to its longer half-life which permits us to image the phantom over a longer period.

The phantom was scanned at two bed positions. At the first bed position, the center of the lesions was in the slice where Z = −3 mm. The phantom was imaged for a period that is equivalent to 200 seconds scan time at each bed position when 370 MBq FDG is uniformly distributed in a 70 kg patient. The panel was placed at four different positions (−67.5, −180, 143.5) (67.5, −180, 143.5) (−67.5, −180, 210.5) (67.5, −180, 210.5) mm to mimic a single Outsert detector that has an active sensing area of 271×135 mm². Additional 200% of counts were also collected to evaluate the performance of an Outsert with 3 times the sensitivity of the current proof-of-concept prototype that only employs 3 mm thick LYSO crystals. Subsequently, the panel-detector was moved to four locations above the scanner's central axis with the detector centered at (−67.5, 180, 143.5), (67.5, 180, 143.5), (−67.5, 180, 210.5), and (67.5, 180, 210.5) mm to repeat the data collection. The above procedures were repeated after the phantom was moved along the Z-axis by 110 mm to mimic a whole-body scan of the torso phantom at a second bed-position. A CT scan of the phantom was also acquired for attenuation correction.

D. Image Reconstruction and Analysis

The two bed positions, both in simulation and experimental studies, were chosen to ensure that the spherical tumors were in the FOV of the scanner at the first bed position and go outside of the FOV of the scanner but in the augmented imaging zone(s) at the second bed position. All images were reconstructed using custom list-mode image reconstruction software [36] that computes the system matrix on-the-fly using 4 GPUs [37, 38]. Attenuation correction is estimated by forward projection using a CT image derived attenuation map. A simplified version of the component-based normalization [39, 40] which takes into account the geometric effect and individual crystal efficiency has been

implemented. The 3D image volume is a $400 \times 400 \times 400$ voxel rectilinear space, with each voxel a $1 \times 1 \times 1 \text{ mm}^3$ cube. TOF information was applied in the reconstruction of simulated data. Prompt events were used to generate the images in this study. Currently, we have not implemented any scatter and random corrections in the reconstruction.

Receiver operating characteristic (ROC) curves [41] were estimated for reconstructed images in the simulation study by plotting the True Positive Fraction (TPF) against the False Positive Fraction (FPF). In the calculation of TPF, a spherical region-of-interest (ROI) with a diameter of 4 mm was drawn over the center of each lesion to estimate the count density. A hexagon grid was drawn over the entire 3D image volume to simplify the calculation of FPF. Spherical ROI with the same size was drawn over the vertex of every hexagonal grid to compute the count density. TPF and FPF were computed across all possible threshold values [42, 43].

CRC was computed for each spherical lesion in the torso phantom by using a spherical ROI drawn over the center of the corresponding sphere in the image volume to estimate the mean count density, $C_{T,i}$ in the i -th lesion. The diameter of the ROI is the same as that of the lesion under evaluation. In the images reconstructed using experimental data, co-registered CT images were used to define the center positions of the lesions. The mean counts within a ROI was determined by the weighted sum of counts in voxels that overlap with the ROI. The background count density, $C_{b,i}$ was computed as the mean number of counts in a square ROI drawn over the 3D image from 20 adjacent slices. The CRC_i for the i -th sphere was calculated according to the National Electrical Manufacturers Association (NEMA) NU2–2001 definition [44]:

$$CRC_i = 100\% \times \frac{\left(\frac{C_{T,i}}{C_{b,i}} - 1\right)}{(uptake - 1)} \quad (1)$$

where uptake is 5 in the simulation study and 6 in the experimental study. The CRC is reported as an average of the seven tumors of one group in the simulation study, and three tumors of the same diameter in the experimental study.

III. Results

A. Sensitivity Profile

Fig. 4 shows the sensitivity profile of the system when a point source was stepped across the axial FOV at different vertical positions ($Y = 100, 0$ and -100 mm). The overall sensitivity enhancement for the single-panel and dual-panel AWSM-PET systems over the native scanner was calculated by comparing the area under the curve (AUC). The results are shown in Table I. The improvement in sensitivity by a dual-panel AWSM-PET device was 35.7% when the source is located at $Y = 0 \text{ mm}$, and 42.0% when $Y = 100$ or -100 mm .

B. Monte Carlo Simulation

Fig. 5(a) shows reconstructed images after the 30th iteration using 200 seconds of coincidence events with TOF information from the native scanner, or a scanner equipped

with the AWSM-PET technology. The number of iterations was evaluated by looking at the ROC curves as a function of the number of iterations of the reconstructed images. After the 30th iteration, the ROC curves become stable. Table II list the number of events used to generate the images. Scanners with one and two Outsert devices gain 16% and 32% sensitivity, respectively, when a distributed source was imaged.

Scanners with the AWSM-PET technology gain visible improvement in tumor contrast when compared to the native PET scanner. ROC curves of the three images are shown in Fig. 5(b). For any given FPF, the AWSM-PET systems offer higher TPF than the native PET scanner. For example, when the FPF is restricted to be 5%, the TPF is 79.4%, 85.7% and 95.2% for the native scanner, a scanner with one Outsert, and a scanner with dual Outserts, respectively. The corresponding AUCs are 0.962, 0.977 and 0.991, respectively.

The dependence of the enhancement in plane by the AWSM-PET devices was studied by analyzing the average CRCs of the 7 tumors in each group for all three images in Fig. 5(a). The results were plotted in Fig 5(c). With one panel Outsert, the enhancement of CRC at region 7, 8 and 9 are higher than regions that are farther away from the panel. With a dual-panel Outsert, the enhancement of the CRCs for tumors at different regions is more uniform. Axial dependence of the benefit from the AWSM-PET technology was not studied because continuous bed motion or small step-and-shoot will be implemented for a future system.

Fig. 5(b) and 5(c) show significant improvement in tumor detectability and contrast recovery of small lesions by the AWSM-technology. However, the enhancement in lesion detectability and CRCs may be the result of higher system sensitivity, improved image resolution, or both. To identify the contribution from individual factors, we reconstructed images using the equivalent number of counts from the native scanner that were acquired in a 200 second scan while the scanner was equipped with the AWSM-PET technology. Reconstructed images and ROC curves are shown in Fig 6. With 232- and 264-second scans, the native scanner acquires the equivalent number of events as the scanner equipped with a one-panel or dual-panel AWSM-PET device. Tumor detectability was improved by increasing counts for the native scanner. However, an Outsert device (single-panel or dual-panel) significantly outperforms a native scanner with a longer equivalent-count acquisition time. This suggests that while the performance enhancement by the Outsert is the result of both improved system sensitivity and improved image resolution, resolution enhancement appears to be the more significant factor improving tumor detectability.

C. Phantom Study

Fig. 7(a) shows the reconstructed images of the phantom study after the 30th iteration using events equivalent to 200 seconds of a typical clinical FDG scan. From left to right, the images are from the native scanner, a scanner with one Outsert, a scanner with dual Outserts, and a scanner with dual Outserts that employ longer crystals to offer three times the sensitivity of the current 3 mm thick panel detectors. The total number of events used to generate these images were 79344156, 81544908, 85664743, 98817149, respectively. The dual Outserts image was obtained by adding up OS events with the single panel below and above the phantom. With high-resolution OO events that could be acquired by a real dual-panel device, the image quality could be further improved. Comparing the image from the

native scanner and scanners equipped with Outsert(s), visible improvement in tumor contrast can be observed, especially for smaller tumors. Fig. 7(b) shows the average CRCs as a function of tumor sizes. The average CRCs are also listed in Table III.

The average CRCs are 2.8%, 4.2%, 8.4%, 16.1% for lesions with diameters of 3.3, 4.3, 6.0, and 8.0 mm, respectively, by the native scanner. A scanner equipped with a single Outsert improves the CRCs to 7.9% and 11.0% for the 3.3 mm and 4.3 mm diameter lesions, respectively. With dual Outserts, the CRCs of the 3.3 mm and 4.3 mm diameter lesions were further enhanced to 8.8% and 12.6%, respectively. For lesions with larger diameters, e.g. 9.6 mm and 11.4 mm, the enhancement in CRCs is less significant as the image resolution of the native scanner is adequate in resolving these larger lesions.

Since the sensitivity of our existing panel detector that employs 3 mm thick LYSO crystals can be easily tripled by employing longer crystals, the acquisition time for the OS events was tripled to mimic a dual-Outsert AWSM-PET system with detectors of 3 mm FWHM depth-of-interaction (DOI) resolution. Without considering the effects of additional scatter events, this result represents the best-case scenario for estimating the performance of a future AWSM-PET device with DOI detectors. Using such detectors, a dual-Outsert AWSM-PET system improves the CRCs of the 3.3 mm and 4.3 mm diameter lesions to 11.0% and 15.9%, respectively, enabling the detection of sub-5 mm lesions more reliably. It is worth mentioning that the spatial resolution of the Biograph 40 PET/CT scanner is more than 4.1 mm FWHM as reported in literature [35].

IV. Discussion

We are unaware of any close-form analytical image reconstruction algorithm that can model the unconventional geometries of AWSM-PET systems in this work. Thus, instead of attempting to quantify the image resolution of an AWSM-PET system versus a native PET scanner based on statistically reconstructed images, the tumor detectability and CRCs of spherical lesions of different sizes were studied to demonstrate the effectiveness of the AWSM-PET technology.

Depending on the design of the Outsert detectors, the overall system sensitivity and/or image resolution may be further improved. The choice of detector technology is often a tradeoff between its cost-effectiveness and the desired performance, constrained by other design considerations such as complexity of other system components (e.g., cooling of sensors) and data correction techniques. In the case of AWSM-PET detector design, it is desirable to match the Outsert detector characteristics to those of the scanner's in order to minimize the need for system re-development. For example, a Siemens Biograph Vision scanner has a CRT of less than 250 ps FWHM. In order to fully benefit from the TOF-PET feature of such scanners, the CRT of the Outsert detectors should have a comparable timing performance as we try to achieve high intrinsic spatial resolution in detector innovation. The actual design for the AWSM-PET Outsert detectors will also depend on the capability of a scanner's readout electronics. For example, smaller crystal cross-section will improve the intrinsic spatial resolution of the detector. However, if the total number of crystals per detector block is constrained by light sensors (e.g. SiPM array) or readout electronics, then the choice of

smaller crystals will limit the total sensing area of the Outsert detectors thus the overall system sensitivity. Similarly, shorter crystals will enhance the timing performance and reduce potential parallax error but at the expense of fewer OS and OO events. To maximize the overall system sensitivity, one may choose to introduce detectors with DOI capability but at the expense of complex readout schemes and normalization algorithms. Therefore, the actual design of Outsert detectors will need to be optimized before an AWSM-PET system is implemented. Additional studies to identify the most cost-effective detector and system design for the proposed AWSM-PET technology will be conducted.

In addition to the optimization of the Outsert detector design, new hardware, firmware and software needs to be developed, in order to establish coincidence detection between the scanner and the Outsert detectors without compromising the scanner's native functions.

The Biograph Vision PET/CT scanner has 19 DEAs, while the Outsert(s) simulated in the Monte Carlo study has the same active sensing area as that of a DEA. That is, a dual-Outsert AWSM-PET device will only require detector materials and read-out electronics that are equivalent to two additional DEAs. This additional material cost is roughly 10% of the detector cost of a native whole-body PET/CT scanner, while the overall system sensitivity is improved by 36%–42% when using the AWSM-PET technology. This suggests that the AWSM-PET technology could become a cost-effective solution to bring high-resolution PET imaging to clinical applications, especially for its compatibility with whole-body imaging protocols.

In this study, data from two bed positions was acquired and reconstructed to demonstrate the feasibility of an AWSM-PET device for enhancing tumor detectability using a commercial PET/CT scanner. However, the use of step-and-shoot scanning with two bed positions is non-ideal and was done only to simplify the image reconstruction work for this proof-of-concept study. The axial dependence of lesion detectability was not studied because it will depend on many factors when using step-and-shoot motions. For example, the step size of each bed position will affect the overlapping fraction between the native scanner FOV and the augmented imaging zones by the Outsert. Depending on the overlapping fraction, the performance enhancement will vary along the axial direction. Acquisition using continuous-bed-motion will eliminate this axial dependence. However, it will require careful implementation of all correction techniques. This optimal implementation is beyond the scope of this work and will be studied in the future.

In the current simplified version of the component-based normalization, only the geometric effects and crystal efficiency component have been considered. Additional components such as the crystal interference pattern of the Outsert detectors, axial effects and dead time parameter will be included after the device hardware is fully developed. The position of the Outsert panels relative to the scanner will not be changed once the system is built. Therefore, the geometry of the AWSM-PET system is fixed. The time-consuming part of the calibration for normalization (such as crystal interference pattern, axial effects and dead-time parameters) only needs to be measured once. The actual dead-time of the system will depend on the radioactivity injected and patient size, and it will be estimated and corrected based on detector blocks' single event rate. With a roughly 10 mm thick device close to the patient,

random correction and scatter correction techniques are critical in order to produce fully quantitative PET images using such a system. These correction techniques will be implemented for our GPU-based list-mode image reconstruction framework [36].

V. Conclusion

A novel AWSM-PET technology, which integrates a high-resolution Outsert device outside the axial FOV of a PET/CT scanner to augment whole-body scans, has been developed and evaluated. Simulation studies show an enhancement of overall system sensitivity of 37%–42% for a dual-Outsert AWSM-PET system. This technology improves the lesion detectability of a whole-body PET scanner. The experimental imaging study validates the technology and shows higher contrast recovery when an Outsert detector was integrated with a Siemens Biograph 40 PET/CT scanner, especially for lesions that are borderline detectable by the native scanner. Optimization of the Outsert detector is underway in order to implement the AWSM-PET technology for a current state-of-the-art PET/CT scanner, the Biograph Vision.

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Fig. 1.

(a) A basic AWSM-PET system consists of high-resolution detectors (an Outsert) placed outside the scanner's axial FOV to survey the body in conjunction with scanner detectors. At the first bed position, the three tumors (represented by red circles) are in the native scanner's imaging FOV. When the patient is moved to the second bed position, these tumors are outside the scanner's FOV but inside the augmented zone 1. These additional coincidence events detected by an Outsert are information-rich. Joint image reconstruction using SS events from the bed position 1 and OS events from bed position 2 can enhance both the image resolution and system sensitivity. (b). An AWSM-PET system with dual-panel Outserts creates additional augmented zones 2 and 3 to further improve the sensitivity and image resolution of the scanner.

Fig. 2.

(a) AWSM-PET with one Outsert detector; (b) AWSM-PET with dual-Outsert; (c) Tumor pattern simulated in an elliptical torso phantom and definition of tumor region numbers.

Fig. 3.

(a) Setup of the torso phantom imaging experiment with a flat-panel detector placed outside a scanner's axial FOV; (b) Photo showing the fillable lesions of different sizes. For each size group, only three lesions were filled with blue dyes and radioactivity. The others were filled with water. All lesions were aligned to be visible in the same image slice.

Fig. 4.

Sensitivity profiles at different positions across the axial FOV for (a) a single-panel AWSM-PET system as shown in Fig. 2(a), and (b) a dual-panel AWSM-PET system as shown in Fig. 2(b).

Fig. 5.

(a) Reconstructed images of a torso phantom containing 63 lesions (4 mm in diameter, T/B = 5) measured by a simulated Siemens Biograph Vision PET/CT scanner, and the same scanner equipped with one Outsert or dual Outserts as shown in Fig. 2; (b) ROC curves of the three systems; (c) Average CRCs of the 7 tumors in each group as a function of different regions for all images in (a).

Fig. 6.

(a) Reconstructed images using events from 232- and 264- second scans with the native scanner. (b) ROC curves of the five different configurations.

Fig. 7.

(a) Reconstructed images of the torso phantom in Fig. 3 ($T/B = 6$) imaged by a native Biograph 40 PET/CT scanner or the same scanner equipped with a single-Outsert, with dual-Outserts, or dual-Outserts that have higher sensitivity; (b) Average CRCs measured by the above four configurations are shown as a function of the lesion sizes.

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TABLE I

Improvement in System Sensitivity by the AWSM-PET Technology

Configurations	One-panel Outsert			Dual-panel Outsert		
	OS/SS	OO/SS	(OS+OO)/SS	OS/SS	OO/SS	(OS+OO)/SS
Sensitivity Enhancement						
Y = 100 mm	10.9%	-	10.9%	36.3%	5.7%	42.0%
Y = 0 mm	14.9%	-	14.9%	27.7%	8.0%	35.7%
Y = -100 mm	26.9%	-	26.9%	36.3%	5.7%	42.0%

TABLE II

Total number of different types of events to generate the images in Fig. 5.

Number of events used to generate the images	Scanner Only	One Panel Outsert	Dual-Panel Outsert
Prompt (% of counts gained by the Outsert)	169,203,762	195,498,026 (15.54%)	223,162,841 (31.89%)
True (% of Prompt)	109,843,714 (64.9%)	126,291,724 (64.6%)	143,716,869 (64.4%)
Random (% of Prompt)	2,901,947 (1.7%)	3,518,964 (1.8%)	4,016,931 (1.8%)
Scatter (% of Prompt)	56,458,101 (33.4%)	65,687,336 (33.6%)	75,429,040 (33.8%)

TABLE III

Average CRC of different size of tumors for different systems

Diameter of the spheres	3.3 mm	4.3 mm	6 mm	8 mm	9.6 mm	11.4 mm
Scanner Only	0.028 ± 0.029	0.042 ± 0.009	0.084 ± 0.020	0.16 ± 0.030	0.18 ± 0.017	0.22 ± 0.005
Panel at bottom	0.079 ± 0.020	0.11 ± 0.012	0.21 ± 0.012	0.31 ± 0.007	0.32 ± 0.025	0.37 ± 0.006
Two panels	0.088 ± 0.042	0.13 ± 0.016	0.24 ± 0.021	0.36 ± 0.035	0.33 ± 0.026	0.39 ± 0.012
Two panels with 3 OS	0.11 ± 0.005	0.16 ± 0.023	0.32 ± 0.040	0.47 ± 0.029	0.40 ± 0.031	0.49 ± 0.021