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Effective Count-Rates for PET Scanners with Reduced and Extended Axial Field of View

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Abstract

We investigated the relationship between noise equivalent counts (NEC) and axial field of view (AFOV) for PET scanners with AFOVs ranging from one-half to twice those of current clinical scanners. PET scanners with longer or shorter AFOVs could fulfill different clinical needs depending on exam volumes and site economics. Using previously validated Monte Carlo simulations, we modeled true, scattered, and random coincidence counting rates for a PET ring diameter of 88 cm with 2, 4, 6, and 8 rings of detector blocks (AFOV 7.8, 15.5, 23.3, and 31.0 cm). Fully 3D acquisition mode was compared to full collimation (2D) and partial collimation (2.5D) modes. Counting rates were estimated for a 200 cm long version of the 20 cm diameter NEMA countrate phantom and for an anthropomorphic object based on a patient scan. We estimated the live-time characteristics of the scanner from measured count-rate data and applied that estimate to the simulated results to obtain NEC as a function of object activity. We found NEC increased as a quadratic function of AFOV for 3D mode, and linearly in 2D mode. Partial collimation provided the highest overall NEC on the 2-block system and fully 3D mode provided the highest NEC on the 8-block system for clinically relevant activities. On the 4-, and 6-block systems 3D mode NEC was highest up to ~300 MBq in the anthropomorphic phantom, above which 3D NEC dropped rapidly, and 2.5D NEC was highest. Projected total scan time to achieve NEC-density that matches current clinical practice in a typical oncology exam averaged 9, 15, 24, and 61 min for the 8-, 6-, 4-, and 2-block ring systems, when using optimal collimation. Increasing the AFOV should provide a greater than proportional increase in NEC, potentially benefiting patient throughput-to-cost ratio. Conversely, by using appropriate collimation, a two-ring (7.8 cm AFOV) system could acquire whole-body scans achieving NEC-density levels comparable to current standards within long, but feasible, scan times.

Introduction

Positron emission tomography (PET) image quality is strongly dependent on photon counting efficiency, and the fidelity of the detected photons used for image reconstruction. PET is generally considered to be a 'count-limited' imaging modality; in clinical practice PET images are substantially smoothed due to statistical noise that results from photon counting limitations.

Increasing PET scanner sensitivity is a common goal of advancing scanner design. Increased PET scanner sensitivity can be used to reduce patient dose, decrease scan time, or reduce image noise level, while maintaining other image characteristics constant. The noise equivalent count (NEC) parameter is a measure of valid ("true") coincident events relative to unwanted scattered and accidental coincidence events recorded by the scanner, and has been widely used as a measure of effective count sensitivity in PET scanners. NEC is proportional

[†]The relationship was credited to an internal report by authors in [4]

to the square of signal-to-noise ratio (SNR) [1], and NEC density (counts-per-unit object volume) has been shown to correlate (under ideal conditions) with lesion detection in simulated observer studies [2].

One way to increase absolute counting sensitivity is to increase the geometric sensitivity (solid-angle subtended between the object and scanner), which can be achieved through reducing the diameter and/or extending the axial length of the detector ring. Common clinical PET scanners have ring diameters of 80–90 cm, and axial field of view of roughly 15 – 20 cm. Reducing the ring diameter increases transaxial parallax errors that degrade spatial resolution, and the ring diameter can only be reduced so much and still accommodate a reasonable range of patient sizes. Furthermore, there is currently interest in increasing the patient port size to accommodate radiation therapy planning.

Increasing scanner axial length can increase geometric sensitivity without the associated drawbacks of reducing the ring diameter. Increased system cost is the main constraint to extending PET AFOV. Extended axial field-of-view (AFOV) PET systems have been the subject of several simulation studies [3–8]. These studies focused on significantly longer AFOV systems (up to 350 cm) operating in fully 3D (septaless) acquisition mode. The detector scintillators included BGO and LSO. Additionally, Jakoby and colleagues compared NEC rates measured on two clinical PET scanners of different AFOV [9]. The increase in NEC rate with scanner AFOV was predicted to be a quadratic relationship, and that relationship was confirmed for a standardized cylindrical test object for the two scanners tested. However, when estimated for patient scans, the NEC increase was less than quadratic. Reasons for this discrepancy likely include the difference in activity distributions (uniform and concentrated in the phantom, non-uniform and distributed in the patient), and the limited axial length of the standard phantom (70 cm).

Naturally a system with longer AFOV will have higher sensitivity because there are more detectors. We frame our analysis in the context of NEC-density, which will refer to NEC per unit AFOV in this study. If the rate of increase of NEC sensitivity with scanner AFOV is greater than linear (e.g. NEC increases as the square of AFOV), then NEC-density will also increase with AFOV. This could potentially benefit the patient throughput-to-cost ratio, making the longer AFOV system advantageous for a clinic with high patient volumes. On the other hand, such a relationship means shorter scanners have lower NEC-density. Hence, on a short AFOV scanner, in addition to the larger number of bed positions required to cover a given axial range, the lower NEC-density would also require longer scan time per bed position to collect equivalent integrated noise equivalent count density in the final image. Minimal total scan duration is important not just for patient throughput, but also for patient comfort and for limiting patient motions that become more likely for longer scans. In this study we examined NEC variations with scanner AFOV, and estimated how exam times change for different AFOV and operating modes. Given that scan time will be longer on shorter AFOV PET scanners, we tried to determine if whole-body scans can be acquired in a feasible time from the patient-motion viewpoint.

We used simulations to characterize NEC for BGO scanners that have both longer and shorter AFOV's than the current clinical PET axial length. We considered a modular axial design, which may reduce the costs of incorporating this type of a flexible approach, and evaluated scanners with between one-half and twice typical AFOV (i.e. ~8–30 cm). The axial modules were based on an existing detector block design and fixed ring diameter. We used objects longer than the standard cylinder to account for scattered and random coincidence events originating farther outside the field of view, as would be present in clinical scanning. We included the use of full and sparse axial septa in the simulations following previous work that showed potential advantages of sparse septa [10–11]. We then

estimated total scan times for whole-body exams that acquire integrated NEC density consistent with current oncology protocols.

Materials and Methods

We used the SimSET Monte Carlo simulation package [12], and data modeling techniques previously employed and validated for estimating relative global count-rates on different PET scanner configurations [11, 13]. We simulated true, scattered (Compton and coherent), and random coincidence counts detected by a PET scanner based on a clinical scanner, and also with reduced and extended AFOV. The PET scanner was based on the Discovery STE PET/CT, which consists of four rings of bismuth germinate (BGO) detector blocks arranged axially. The ring inner diameter is 88 cm, and the detector block axial extent is 3.88 cm. The four detector blocks in the DSTE result in an AFOV of 15.5 cm. The transaxial FOV is 70 cm diameter. We compared results from a 4-block ring system (DSTE), with results from scanners with 2-, 6-, and 8-detector block rings, with corresponding scanner AFOVs of 7.8, 15.5, 23.3, and 31.0 cm.

The DSTE operates in either fully 3D acquisition mode, or 2D mode. Each block contains 6 BGO crystals axially, resulting in 24 rings of crystals axially (6.4 mm per crystal). Tungsten septa 5.4 cm long (radial) and 0.8 mm thick were placed between each crystal ring in 2D mode. We also simulated the scanner with one septum between every two crystals. This collimation mode is referred to as 2.5D partial collimation, and was shown in a previous study to have advantages over 2D and 3D mode for NEC on the Advance and DSTE scanners [10–11]. Our simulations of reduced and extended AFOV modeled these three collimation modes. In all cases lead end-shielding at the axial ends of the scanners were also 5.4 cm in the radial dimension.

Two objects were used in the simulations: a solid cylinder with an embedded line source, and an anthropomorphic object based on a patient whole-body scan. The cylinder was 20 cm diameter, 200 cm long, solid polyethylene with a 3 mm diameter line source running along its length, and 4.5 cm from the cylinder axis. This object emulates the standard NEMA NU-2 count-rate phantom used for NEC measurements (20 cm diameter by 70 cm long cylinder [14]), but has been extended in axial length for the present study. Since activity well beyond ± 35 cm axially can contribute singles events (and thus scattered and random coincidences), it is important to use a phantom that extends near to the limits of where the longest AFOV modeled (31.0 cm) can be exposed to single photon events in 3D-mode. The standard 70 cm long cylinder phantom will not capture important NEC trends because as AFOV is increased, the outside-FOV activity contribution is not concordantly increased. The 200 cm long cylinder provides increased outside-FOV activity for each AFOV studied here, and is longer than most patients (figure 1).

The second object was derived from a typical whole-body FDG PET/CT patient scan on the DSTE PET/CT, and voxelized for input into the SimSET program. This patient was 160 cm tall, and 80 kg. Figure 1 illustrates the relative orientation of the different AFOV scanners and objects. Dashed lines show the axial extent that the central crystal could be exposed to outside-FOV activity in 3D mode. Note that the dashed lines in figure 1 indicate 3D AFOV of the central crystals, and that crystals at the axial edge can be exposed to activity even farther than indicated.

Figure 1 shows the relative ^{18}F activity distribution within the anthropomorphic phantom. Figure 2 shows the distribution of attenuation coefficients in the voxelized phantom as determined by the patient CT scan. The continuous range of linear attenuation coefficients

found in the patient CT scan (fig. 2A) was segmented into appropriate SimSET attenuation index values as shown in fig. 2B.

The version of SimSET used in this study (v2.6.2.6) models the detector as a solid annulus of scintillation material. Adjustments for sensitivity losses due to inter-crystal and inter-block material in the block detector were made with a multiplicative packing fraction [11]. Random coincidence counts were estimated from the rate of single photons absorbed in the detector following Stearns et al. [15]. Coincidence events, and singles for calculating random event rates were accepted within an energy window of 425–650 keV. The coincidence window was set to 9.75 ns for computing randoms from singles.

Simulated data were binned according to the DSTE crystal positions, and then into 2D sinograms using single-slice rebinning (SSRB). Different axial acceptances within the SSRB algorithm were investigated. NEC was then calculated directly from the simulated true (T), scattered (S), and random (R) coincidence counts using the following equation

$$NEC = \frac{T^2}{T + S_{ROI} + R_{ROI}} \quad (1)$$

The ROI subscript refers to using scattered and random events within a 24 cm wide window, centered on the object, from sinogram data that has been sorted into 2-dimensional bins using single-slice rebinning, as prescribed in the NEMA NU2 standard [14]. The term R_{ROI} in the denominator is not amplified in this case since random coincidence events were estimated from singles event rates (i.e. low-noise estimation). Scatter fraction (SF) is calculated from the same windowed sinogram data as $SF = S_{ROI}/(T + S_{ROI})$. Calculating NEC and SF from measured data requires separating prompt events into constituent T, S, and R. Simulated data separates these types of events inherently. Here we calculated NEC and SF directly from inherently separate simulated T, S, and R coincidence counts. We used the true sinogram to create a window then applied to the scattered and random events sinograms for calculating NEC and SF in the anthropomorphic phantom.

To relate results in this work to standard NEC measurements, a comparison was made between NEC from the axially-extended 200 cm cylinder and the 70 cm long NEMA standard cylinder from which it was derived.

Evaluating 'optimal' NEC can be challenging because NEC varies with activity in the object or patient. Evaluating systems based on the maximum NEC achieved only makes sense if that maximum occurs with a clinically relevant amount of activity in the object, which is not the case for 2D acquisition mode. Instead we compared NEC of the systems for a given object activity. For this work we compared NEC between systems when there was 260 and 370 MBq in the objects. An activity of 260 (370) MBq corresponds approximately to an injection of 370 (555) MBq that has decayed for 60 minutes, without considering metabolic clearance or voiding.

Using the optimal collimation mode, i.e. that which maximized NEC-density in the anthropomorphic phantom, we estimated the total scan time for each of the four AFOV systems to cover an axial field of 75 cm and achieve a typical clinical NEC-density. An axial bed position overlap of 25% of the system AFOV was used in calculating the number of bed positions needed. As a reference for a typical clinical NEC-density we used a 4-minute per bed position acquisition on the 4-block ring system (AFOV = 15.5 cm). The calculations did not account for non-uniform activity distribution in the patient, i.e. the varying NEC-density with bed position.

Results

Comparison of count rates in 70 cm and 200 cm long cylinders

The relationship between NEC rate and phantom activity changes when using the 200 cm long cylinder phantom instead of the conventional NEMA-standard 70 cm long cylinder. The peak NEC rate is reduced from 100 kcps to 85 kcps and the activity it occurs at increases from 300 MBq to 700 MBq. Figure 3 compares NEC rate versus phantom activity on the DSTE scanner in 3D acquisition mode (4-block rings, 15.5 cm AFOV) for the 200 cm and 70 cm long cylinder phantoms, and from the anthropomorphic phantom.

NEC vs. axial acceptance angle

Axially oblique coincidence events are restricted by septa in 2D and 2.5D modes, to ~11 cm and 22 cm axial separation, respectively. Increasing axial acceptance of coincident events beyond these limits adds primarily scattered and random events in theory, each of which degrade NEC. In our simulations peak 2D-mode NEC was highest when restricting coincidence line of response (LOR) acceptance to a maximum axial distance of ~6.4 cm between photon pairs. The same was true of 2D NEC with 370 MBq in the 200 cm long cylinder phantom. Figure 4 shows the relationship between NEC and axial acceptance for the longest system that we modeled (AFOV = 31.0 cm). Peak NEC continued to increase in 2.5D and 3D acquisition modes when larger axially oblique LORs were accepted, and the same was true for 2.5D and 3D NEC with 370 MBq in the 200 cm long cylinder phantom (Figure 4). For 2.5D mode the increase in NEC between accepting events separated by ~13 cm axially, and 31 cm (accepting all LORs) is small (5%). Note the difference in relative NEC values between collimation modes when comparing peak NEC, regardless of phantom activity, with NEC achieved with 370 MBq in the phantom (Fig. 4A vs. Fig. 4B).

These relationships were similar for other AFOV systems, and for 260 MBq in the phantom. In the following sections we present count rate estimations obtained using maximum axial acceptance that gave the highest NEC, i.e. 6.4 cm in 2D mode, and acceptance of all axial oblique lines-of-response in 2.5D and 3D.

Count rates vs. axial field of view

20 cm diameter by 200 cm long cylinder phantom—In 2D mode the true, scattered, random, and NEC event rates increased linearly with AFOV, whereas in 3D mode the relationships were quadratic. Partial collimation (2.5D) count rates also appeared quadratic, with smaller coefficients than 3D mode. The modeled count rates as a function of AFOV with 260 MBq in the 200 cm cylinder phantom are shown in figure 5. NEC-rate densities (NEC/AFOV) for 260 MBq and 370 MBq phantom activities are given in table 1. The *relative* 3D mode true, random and NEC event rates for the same phantom arrangement are shown in figure 5D. There were similar relative relationships for 2D and 2.5D modes but with their associated AFOV functional dependence. In the relative comparison, as expected randoms rates increase more than true events with increasing AFOV. Scatter fraction (SF) also increased with AFOV (table 2). Both increasing relative random counts and SF act to degrade NEC, yet NEC was seen to closely follow the increasing trends of the true coincidence rates for each collimation mode (fig 5D shows the case of 3D mode).

Anthropomorphic Object—Relative count rate trends between collimation modes in the anthropomorphic phantom differed slightly from those found using the 200 cm long cylinder phantom. Again the 3D and 2.5D NEC rates were higher than 2D NEC for clinical activities. The NEC count-rate curves versus anthropomorphic phantom activity are shown in figure 6 for each collimation mode and AFOV system, and the ratio of 2.5D NEC to 3D NEC is also

plotted in fig. 6D. Tables 3 and 4 contain the NEC-density for 260 and 370 MBq and scatter fractions in the anthropomorphic phantom.

The 2D NEC in the anthropomorphic phantom varies nearly linearly with system AFOV, meaning NEC-density in 2D acquisition modes is nearly constant with AFOV. Fully 3D mode NEC rises approximately quadratically, as in the 200 cm long cylinder phantom.

Tables 5 and 6 give the expected total acquisition time at 260 MBq and 370 MBq phantom activity levels for a typical clinical oncology protocol.

Discussion

In the interest of increasing PET sensitivity and reducing scan times, extended AFOV PET scanners have been considered and reported on previously [3–8]. In some of these studies the axial length was extended by an order of magnitude (to ≥ 200 cm) over current systems to achieve whole-body PET scanning in a single acquisition [4, 6, 7]. Here we reported on shorter than conventional AFOV scanners as well as moderately longer (7.8–31 cm). Our results showed that axial septa collimation could benefit NEC rates for short (7.8 cm) AFOV PET systems, while septaless 3D mode provided the highest NEC rates for long (31.0 cm) AFOV systems. For the 7.8 cm AFOV system, estimated total scan time to achieve an integrated NEC density matching current PET acquisition protocols averaged 61 minutes. This is considered long by today's standards, but is in line with historical PET protocol duration and clinically feasible. Thus, in an institution with low patient volumes, a short AFOV PET scanner at a lower cost may be preferred over the 15 – 20 cm AFOV scanners available today. At the other end of the spectrum, the high 3D-mode NEC for AFOV=31cm could provide a cost-to-patient throughput advantage. Our analyses did not include system cost or pricing projections. We only assumed the costs of a PET system changes with the numbers of detectors required to make longer AFOV systems.

We used collimating septa in this analysis based on earlier studies that suggested partial collimation could provide advantages in NEC rates [10–11]. Those results were supported by measured count rates in partial collimation mode on the DSTE PET/CT system [11]. In theory septa limit the detection of true coincident event LORs beyond an axial difference of approximately 10 cm in 2D, and 20 cm in 2.5D. The fact is that annihilation photons penetrate the septa, leading to detection of additional true coincidence events beyond the nominal limits of the collimation. However, we found 2D NEC was highest when axial acceptance was limited to ~ 6.5 cm, which is less than the restrictions of the septa geometry. This suggests a relative increase in the number of scattered and random counts over true events for axial acceptance between the observed and theoretical optimal limits. For 2.5D the highest NEC was achieved using all axial combinations, but reached 95% of that highest value when limiting acceptance to ~ 16 cm (fig. 4 shows results for the 8-block ring system; similar trends held for 4- and 6-block systems, with less of a plateau region). In 3D mode the NEC rate was highest for full axial acceptance also, and approached a significantly higher plateau value than 2D and 2.5D for clinical activity in the phantom (fig. 4B). Maximum axial acceptance could be a significant consideration for long AFOV systems where the number of possible coincidence LORs becomes very large. Limiting the axial acceptance angle reduces the data processing burden. Also, axially oblique LORs can degrade spatial resolution through parallax effects and possible simplifications in system modeling in the image reconstruction method. The optimal global NEC is only one of several considerations for determining the optimal axial acceptance. Restricting oblique LORs would only be necessary for longer AFOV systems, for which 3D mode NEC was found to be considerably higher than collimated modes, even when restricting axial acceptance (fig. 4B). This is true when considering clinically relevant activities. The highest overall peak NEC was achieved

in 2D-mode (fig. 4A), but this occurred at a phantom activity above 2500 MBq, which is more than would be injected for typical oncology scanning. NEC in 2D mode did not surpass NEC in other modes until there was approximately 740 MBq in the phantom. These results were derived from the cylinder phantom.

The axially symmetric cylinder phantom was 200 cm long in order to capture the effects of random and scattered coincidence events originating farther outside of the FOV than provided by the standard 70 cm long cylinder count-rate phantom. The scanner geometry considered here was susceptible to single events from as far as 127 cm from the center of the scanner (8-ring system, 3D acquisition mode, fig. 1). While events from that distance contribute proportionally less than within-FOV events due to $1/r^2$ effect and longer attenuation paths through the object, we considered it important to include the presence of this activity in a study comparing scanners with variable AFOV. The phantom activity that gave rise to peak NEC was substantially different for a 200 cm long cylinder phantom compared to a 70 cm long cylinder phantom (fig. 3). The peak NEC value was 25% lower for the 200 cm cylinder in this case. The reduction in peak NEC rate in the 200 cm phantom relative to the 70 cm long phantom demonstrates that outside-of-FOV activity was affecting system count rates, and stems from increased singles rates incident on the detectors that leads to: (i) more accidental coincidences events, and (ii) fewer recorded true events because of a higher dead-time fraction. Each of these effects reduces NEC. Cylinder phantoms with larger diameters than 20 cm (e.g. 35 cm) have also been used for NEC analysis [10, 11, 16] since 20 cm is generally considered too small to accurately model scattered events from a patient. Previous work showed that NEC rates were significantly reduced in a 35 cm diameter cylinder relative to a 20 cm diameter cylinder phantom, and the activity of peak NEC shifted slightly lower, but that relative trends among different collimation modes were consistent [10–11]. We elected to use the NEMA standard phantom diameter and not extend these analyses to a 35 cm diameter phantom, although given the scatter environment of typical patients, a 35 cm diameter cylindrical phantom was found to be more appropriate [6]. Despite our use of a 200 cm long cylinder phantom for the present study, the NEMA NU2 guideline is still believed to be a suitable protocol for routine characterization of current clinical systems because of its relative ease of implementation. Analyses using the 20 cm diameter by 200 cm long cylinder are susceptible to the same limitations reported for the standard 70 cm long cylinder (e.g., [17–18]), which motivated use of the anthropomorphic phantom.

The anthropomorphic phantom represented a more realistic activity distribution than the cylinder phantom. This phantom is based on a patient that is shorter and heavier than the median patient and so represents an object towards the opposite extreme of typical relative to the long thin cylinder phantom. The same general trends held for both phantoms, although there were some differences between optimal collimation methods for each phantom (table 1 vs. table 3). The differences were likely due primarily to different activity distributions and scatter environments (tables 2 and 4).

In each phantom true, scattered, random, and NEC event rates increased as a quadratic function of AFOV for 3D and 2.5D modes, and linearly for 2D mode (fig 5 shows the results for the cylinder). Notably, the random event rates increased considerably faster with AFOV than did true rates (fig. 5D). This result agrees with previous work reporting an increased randoms fraction with AFOV [3]. In addition to randoms fraction, scatter fraction increased with scanner AFOV (tables 2, 4). However, the nature of the NEC metric (eqn. 1) resulted in NEC varying in accordance with true event rates despite the relatively larger increase in random event rates and increased SF. Fig. 5 shows results for 260 MBq in the cylinder phantom; the nature of these relationships held at other activity levels as well, but with different parametric values.

It is important to consider the behavior of NEC rate with object activity since clinical scanning exposes the system to a range of activities. The activities in the anthropomorphic phantom that gave peak NEC rate in 3D-mode were near doses used clinically (~250–450 MBq) for each scanner modeled here, whereas peak 2D NEC occurred for activities well above clinical doses (> 1000 MBq), and peak 2.5D NEC occurred near clinical limits (~500–700 MBq). Ideally the dose injected into a patient could be customized so the scanner operates at peak NEC rate during the scan, such as proposed by Watson and colleagues [18]. In general, scanners will experience a variety of count rates, so optimizing NEC over that expected range is desirable. We compared NEC-density at two clinically relevant activity levels: 260 and 370 MBq total in each phantom. While we used these two activity levels as benchmarks, we will consider our conclusions below in context of the broader NEC vs. activity relationship using fig. 6.

Somewhat unexpected was the fact that the septa benefited the shorter AFOV systems. The general finding was that, for clinical activity concentrations in the anthropomorphic object, the short AFOV system (7.8 cm) had highest NEC-density in 2.5D acquisition mode, and the long AFOV system (31 cm) had highest NEC-density in 3D mode (fig. 6B, table 3). We note, however, that for the short two-ring (7.8 cm AFOV) system the difference in NEC between 3D and 2.5D modes was small. If we examine the NEC vs. phantom activity curves in figure 6 we can draw the following conclusions regarding optimal collimation mode for each AFOV system. For the 2-block ring AFOV system, each collimation mode provided roughly equivalent NEC up to about 125 MBq (3mCi) in the phantom, after which 2D and 2.5D modes provided higher NEC. On the 8-block ring system 3D mode provided the highest NEC up to 475 MBq (13mCi), which covers most clinical protocols. There was a transition between these two cases for the 4-, and 6-block ring systems in which optimal NEC was a trade-off between 2.5D and 3D modes, with 3D NEC being higher at lower activities, but surpassed by 2.5D NEC above ~300 MBq (8mCi). The 2.5D NEC rate curve was relatively broad, thus giving a wider range of high-NEC operation. In 3D mode the NEC rate dropped rapidly after peaking. A potential secondary benefit of using partial collimation is that the reduced photon flux will reduce event mispositioning from pulse pile up that can cause spatial and quantitative distortions. Conversely, partial collimation acquisitions can complicate iterative image reconstruction [19]. We did not consider those effects here, focusing instead on NEC sensitivity and scan acquisition times.

We estimated clinical scan times for whole-body imaging on each AFOV system using the NEC rate density. Based on the premise that NEC is proportional to the square of the scanner AFOV, it would follow that doubling or halving the scanner AFOV would reduce or increase scan time by a factor of four, respectively (to achieve a given NEC-density for a given scan length). This highlights the benefit of longer AFOV, and exposes a potentially prohibitive limitation to a short AFOV scanner. However, we found that by using septa collimation, NEC varied linearly with AFOV, and for the system with 7.8 cm AFOV the highest NEC-density was achieved in 2.5D acquisition mode (table 3). Thus, using the 'optimal' acquisition mode (i.e. the mode providing highest NEC rate), the projected whole-body scan time for the 2-block ring system was found to be roughly a factor of two longer than on the 4-ring system (tables 5, 6), rather than a factor of four longer as predicted by a quadratic relation between NEC and AFOV. While a quadratic reduction in whole-body acquisition time was not achieved on the longer AFOV systems (6-, and 8-ring systems) relative to the 4-ring system, the improvement was better than linear (tables 5 and 6).

As mentioned, most prior studies have focused on longer PET AFOV, aiming for shorter scan duration (e.g. Eriksson). The quadratic nature of 3D mode NEC vs. AFOV has been claimed by others for other phantom objects [4, 9], but never explicitly demonstrated to our knowledge[†]. Siemens Molecular Imaging Inc. now offers a PET scanner with an axial length

of either 16.2 cm (Biograph TruePoint with 3 detector block rings) or 21.8 cm (Biograph TruePoint TrueV with 4 block rings). In a report comparing the two scanners, Jakoby et al. [9] showed peak NEC to be 73% greater for the longer system (closely matching the predicted quadratic increase of 78%) when measured with the standard 70 cm long phantom. The improvement was just 49% for patient data whose count-rate response was modeled using the methods described by Watson et al. [18]. Our simulated results for the 200 cm long cylinder in 3D mode agreed in that we found peak NEC rate increased by 78% from the 4-block to 6-block systems (data not shown; peak NEC occurred for ~660 MBq). The increase was 91% and 87% with 260 and 370 MBq in the phantom, as can be calculated from table 1 data. We also found similar increases for the anthropomorphic phantom (3D); 72%, 67%, and 69% NEC increase from 4-block to 6-block AFOV with activities of 260 MBq, 370 MBq, and at peak NEC, respectively (peak NEC occurring ~324 MBq). One limitation of our study was that we only evaluated NEC from the anthropomorphic phantom using one bed position and one patient weight (shown in fig. 1). This limited analysis could be a reason for disagreement with the patient results modeled by Jakoby et al.

Our previous results from similar simulation studies showed good agreement between relative measured and simulated count-rate variations on the Discovery STE PET/CT scanner (15.5 cm AFOV) [11]. In that work, the absolute true-coincidence count-rates were in general over-estimated by the model, which likely extends to this analysis. However, relative peak NEC, and activities giving rise to peak NEC in the model, agreed with measurements. Emphasis should be placed on the relative comparisons between scanners in this work, rather than comparing with measurements on existing systems.

In this work we modeled a tomograph that used BGO scintillation crystals. Nearly all new PET scanners use lutetium-based scintillation crystals (e.g. LSO), which have considerably faster decay time (~40 ns time constant for lutetium-based, vs. 300 ns time constant for BGO), and higher scintillation light yield, but lower stopping power and photoelectric interaction cross section than BGO. Our study considers scanners with AFOV longer and shorter than current typical models for which BGO is a viable choice of crystal; a shorter AFOV scanner would in theory be a less expensive system, and using BGO would aid in reducing system costs, as presently BGO is considerably less costly than lutetium-based crystals. Longer systems require more detector components and BGO could again be used to contain costs. This study used the scanner model that incorporates empirically determined live-time characteristics and has been compared to measured data. Several factors need consideration when extending this analysis to an LSO-based system, notably scanner live-time and crystal thickness. For reference, a comparison of measured NEC curves on similar geometry LSO and BGO scanners shows equivalent NEC at 300 MBq in the NEMA count-rate cylinder phantom [9, 11], indicating that system performance depends on more than just crystal properties. Current LSO-based systems have increasingly fast system electronics for time-of-flight (TOF) capability. Analyses of optimal count-rates on TOF systems is beyond the scope of this work, but analyses analogous to ours using an appropriate model of an existing LSO-based system would be of interest. Based solely on crystal properties, we believe similar or improved results to those found here could be achieved on a scanner using lutetium-based crystals. In particular, whole-body PET scanning on a scanner with just 7.8 cm AFOV should be clinically feasible.

Conclusions

Whole-body PET scans with NEC-density comparable to current imaging protocols should be clinically feasible on a short two-ring (7.8 cm axial field of view) BGO system. The

[†]The relationship was credited to an internal report by authors in [4]

required scan time would be in the range of 60 minutes; relatively long by today's standards, but in line with historical PET protocols. NEC rate-per-AFOV was optimized on the short scanner by using partial collimation (2.5D). Conversely, longer AFOV systems (31 cm) achieved the highest NEC in 3D acquisition mode, which provided a quadratic increase in NEC with AFOV.

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Figure 1. PET scanners with 2-, 4-, 6-, and 8-block rings axially, and the position of the anthropomorphic object, and cylinder object. The dashed lines show the axial extent of the solid angle subtended by the center of the scanner and the end-shielding in 3D mode for the 2-, 4-, and 8-ring scanners (6-ring scanner omitted for clarity). The figure is roughly to scale.

Figure 2. Voxelization and segmentation of the attenuation coefficients of the anthropomorphic phantom. The linear attenuation coefficients (A) were calculated from a CT image via conventional multi-linear conversion on the DSTE. The indicated regions were assigned to SimSET attenuation indexes according to the best match. (B) Segmented attenuation object; central sagittal slice. SimSET material indexes are shown.

Figure 3. NEC rates on the DSTE (15.5 cm AFOV) scanner in 3D mode for the 200 cm long cylinder phantom used in this work, the NEMA standard 70 cm long cylinder phantom, and the anthropomorphic phantom.

Figure 4. Peak NEC (A) and NEC at 370 MBq (B) vs. maximum axial acceptance in SSRB for the 200 cm long cylinder scanned with the 8-block ring system.

Figure 5.

(A) True, (B) random, and (C) NEC event rates vs. scanner AFOV for each collimation mode. (D) 3D mode true, random, and NEC rates were normalized to a value of one at 7.8 cm AFOV and displayed as a function of scanner AFOV to show relative increases for each count type. Rates shown correspond 260 MBq within the 200 cm cylinder phantom.

Figure 6.

(A): NEC rates as a function of activity in the anthropomorphic phantom, for each AFOV system, and each collimation mode. Note the differences in vertical scales.

Table 1

200 cm Cylinder NEC

AFOV (cm)	NEC density (keps/cm) at 260 MBq				NEC density (keps/cm) at 370 MBq			
	2D	2.5D	3D	2D	2.5D	3D	2.5D	3D
7.8	1.6	2.1	2.2	2.2	2.9	2.8		
15.5	1.9	3.0	4.0	2.6	3.9	4.7		
23.3	2.0	3.4	5.0	2.7	4.4	5.9		
31.0	2.0	3.7	6.3	2.8	4.7	7.3		

Table 2

200 cm Cylinder Scatter Fraction

AFOV (cm)	SF (%)		
	2D	2.5D	3D
7.8	23	32	47
15.5	28	40	50
23.3	28	46	54
31.0	29	52	54

Table 3

Anthropomorphic Phantom

AFOV (cm)	NEC density (keps/cm) at 260 MBq			NEC density (keps) at 370 MBq		
	2D	2.5D	3D	2D	2.5D	3D
7.8	1.5	1.8	1.5	2.0	2.2	1.7
15.5	1.7	2.2	2.3	2.1	2.6	2.3
23.3	1.6	2.3	2.6	2.1	2.6	2.6
31.0	1.6	2.3	3.0	2.1	2.6	3.0

Table 4

Anthropomorphic Phantom Scatter Fraction

AFOV (cm)	SF (%)		
	2D	2.5D	3D
7.8	25	38	58
15.5	30	48	64
23.3	32	56	70
31.0	32	61	71

Table 5

Projected acquisition times to achieve equivalent NEC density from each scanner based on optimal^a NEC-density with **260 MBq** in the anthropomorphic phantom

AFOV (cm)	Number of beds needed ^b	Optimal collimation	Time per bed (min)	Total scan time in min. (rel%)
7.8	12.6	2.5D	5.1	63.8 (261%)
15.5	6.1	3D	4.0	24.4 (100%)
23.3	4.0	3D	3.5	13.8(57%)
31.0	2.9	3D	3.0	8.6 (35%)

^a optimal NEC-density: use of collimation mode that provides highest NEC-density

^b covering a 75 cm AFOV, calculated using a bed-overlap of 25% AFOV

Table 6

Projected acquisition times to achieve equivalent NEC density from each scanner based on optimal^a NEC-density with **370 MBq** in the anthropomorphic phantom

AFOV (cm)	Number of beds needed ^b	Optimal collimation	Time per bed (min)	Total scan time in min. (rel%)
7.8	12.6	2.5D	4.6	58.3 (239%)
15.5	6.1	2.5D	4.0	24.4 (100%)
23.3	4.0	3D	3.9	15.5 (63%)
31.0	2.9	3D	3.5	10.0 (41%)

^a optimal NEC-density: use of collimation mode that provides highest NEC-density

^b covering a 75 cm AFOV, calculated using a bed-overlap of 25% AFOV