Abstract—In this simulation study, we evaluate the performance of a limited angular coverage PET system consisting of two/four fast-timing 50 ps FWHM CTR flat-panel detectors made of 5–20 mm long pixelated lutetium oxyorthosilicate crystals. We studied image quality and count rates following the National Electrical Manufacturers Association standard, spatial resolution by imaging a Derenzo phantom and a hot rod, and investigated the sensitivity of different scanner designs. We demonstrated the possible use of such a scanner by imaging a human head and a torso of the extended cardiac-torso (XCAT) digital phantom. All the designs were compared to the reference scanner, based on Siemens Biograph Vision PET/CT scanner geometry. We show that good coincidence timing resolution (CTR) can compensate for lower detection efficiency or smaller angular coverage. Good image quality can be obtained with a simple limited-angle PET system without distortions or artefacts. Substantial degradation of the spatial resolution with increased crystal length is observed in the two-panel design due to the parallax error, but not in the four-panel design. The four-panel design simulated with a CTR of 50 ps FWHM is comparable to that of the current state-of-the-art clinical PET/CT scanner. Similar fast-timing limited-angle planar detectors could enable much less expensive total-body or single organ (dynamically selectable) imaging devices.

Index Terms—CASToR, extended cardiac-torso (XCAT), fast timing detectors, GATE, Geant4, limited angle PET, Monte Carlo simulation, National Electrical Manufacturers Association (NEMA) NU 2-2018, time-of-flight (TOF) PET.

I. INTRODUCTION

IN TIME-OF-FLIGHT (TOF) PET, the difference in the arrival times of the two annihilation photons is measured with high precision that helps localize the emission point along the line-of-response (LOR), and this additional information helps during image reconstruction, leading to reduced noise correlations, and improved image signal-to-noise-ratio (SNR) [1]. A clinical Siemens Biograph Vision PET/CT scanner demonstrated coincidence timing resolution (CTR) of 214 ps FWHM [2]. A record CTR of 58 ps FWHM was recently reported using a pair of small LSO:Ce:Ca crystals (2 × 2 × 3 mm³) coupled to silicon photomultipliers of 4 × 4 mm² [3]. This value is close to the intrinsic limit of current generation fast inorganic scintillators [4], however it was measured with high power readout electronic circuits, thus limiting the scalability from laboratory prototype devices to real-world PET systems, composed of thousands of channels. The progress in 3-D microelectronics integration opens exciting perspectives for a performance breakthrough [5].

Limited-angle TOF-PET systems are being investigated for intraoperative surgical applications [6] and for multimodal compatible flexible PET imaging to fit existing magnetic resonance imaging (MRI) or computed tomography (CT) [7]. Dual panel detectors are being studied to be used for in-beam PET systems used for monitoring proton or heavy ion therapies [8]–[10] and also for positron emission mammography [11]–[13].

PET scanners with limited angular coverage generally generate distorted 3-D images with artefacts. However, with improved timing resolution, we do not need as many projections and the angular sampling requirement to obtain distortion-free, and artefact-free PET images decreases [14]–[16]. This relaxed requirement enables PET systems consisting of two planar detectors with excellent timing resolution. Such a system enables a potential use of modular and flexible designs where two opposite panels could be adjusted to focus on the dynamically selected field of view within the patient anatomy. Four modules can be assembled into a more sensitive PET scanner. These designs could substantially reduce the cost and complexity of scanners at the acceptable loss of sensitivity. In this work, the performance of a limited angular coverage PET imaging system consisting of two or four fast timing
panel detectors made with lutetium oxyorthosilicate (LSO) crystals is studied using Monte Carlo simulations. We decided to explore systems with coincidence time resolutions of currently reachable 200 ps down to 50 ps FWHM, selected by the above-mentioned intrinsic scintillator limit, with intermediate steps at 100 and 75 ps.

II. MATERIALS AND METHODS

Our pilot study is based on Monte Carlo simulations of digital phantoms and different PET scanner designs. The performance of PET scanners was compared using different qualitative and quantitative approaches. We investigated the sensitivity of different scanner designs by imaging a point source at different positions. We evaluated count rates and image quality following the standard defined by the National Electrical Manufacturers Association (NEMA).1 We imaged a Derenzo phantom for a visual inspection of the resolution, and we quantitatively evaluated spatial resolution at a single position near the center of the field-of-view. In the end, we evaluated the feasibility of different scanner designs by imaging the head and thorax region of a digital phantom.

We also compared the performance of panel-based PET scanners to the reference scanner designed following the geometry of the state-of-the-art Siemens Biograph Vision PET scanner. Where available, simulation results were compared to measured data of the Siemens Biograph Vision [2].

A. Limited Angular Coverage PET System

Two groups of different detector geometries with limited angular coverage were studied, shown in Fig. 1. The first consisted of two parallel flat panel detectors and the second of two orthogonal pairs—four panels. The selected size of the panels was $30 \times 30$ cm$^2$, and the distance between the opposite panels in each pair was fixed at 40 cm. Panel detectors were built from arrays of LSO crystals with a fixed transversal size of $3 \times 3$ mm$^2$ while having different crystal lengths $-5, 10, 15,$ and $20$ mm. The reflective gap between the crystals was 0.01 mm. Each scintillation crystal was coupled to one channel of the photosensor. Notation $N_{\text{panels}} \times d_{\text{mm}} \times t_{\text{ps}}$ is used to reference a specific detector design, where $N$ is the number of panels, $d$ is the length of the crystals, and $t$ represents CTR of the specific design.

B. Reference System

The reference system was constructed following the design of Siemens Biograph Vision PET/CT scanner [2], one of the best current state-of-the-art PET scanners available on the market. The system has a 78-cm bore and an axial field of view of 26.3 cm. The scanner has eight detector rings and 38 blocks per ring. Each detector block contains a $4 \times 2$ arrangement of mini blocks (four in the trans-axial and two in the axial direction). A mini-block consists of a $5 \times 5$ LSO array of $3.2 \times 3.2 \times 20$ mm$^3$ crystals. The system’s CTR is 214 ps FWHM, while a 4.1 ns coincidence time window and a standard energy window of 435–585 keV were used for accepting events in the simulation. Note that the reference scanner uses a similar signal processing chain (measured deposited energy in the crystal/module is converted in the single event represented by its detection time, position, and energy—the information recorded by the front-end electronics in the real system) as the limited angle scanners, thus its performance is not fully compatible with the Siemens Biograph Vision.

C. Simulations

The limited angle TOF PET imager was studied using GATE [17] version 8.1, a Geant4 [18] Application for Tomographic Emission: a simulation toolkit for PET and SPECT medical imaging. Information about the simulated systems and their parameters is summarized in Table I.

| Scintillator       | Limited angle scanner | Reference scanner |
|-------------------|-----------------------|-------------------|
| Crystal size      | $3 \times 3 \times 5/10/15/20$ mm$^3$ | $3.2 \times 3.2 \times 20$ mm$^3$ |
| Panel detector size | $30 \times 30$ cm$^2$ | / |
| Axial field of view | 30 cm | 26.3 cm |
| Distance between panels | 40 cm | / |
| Ring diameter | / | 78 cm |
| Energy resolution | 10% | 10% |
| Energy window | 435 - 585 keV | 435 - 585 keV |
| Coincidence time resolution | 200 ps, 100 ps, 75 ps, 50 ps | 214 ps |
| Coincidence time window | 2 ns | 4.1 ns |

1https://www.nema.org/standards/view/Performance-Measurements-of-Positron-Emission-Tomographs
policy was used to determine the hit position within the mini-block—module-based readout. Energy resolution was set to 10% and 435–585 keV energy window was used (Table I).

Singles counts found within a coincidence time window were selected as coincidence events, and in the case when more than two singles were found in coincidence, all good coincidence pairs were included (GATE takeAllGoods policy). In the case of four-panel detectors, we considered only the coincidence pairs originating from singles from opposite panels—no side panel interactions were considered due to the limited view of the object and a large parallax error. The inclusion of such events will be a subject of further studies.

The coincidence event was characterized by two times: 1) the time of the first gamma interaction in the first crystal and 2) the time of the first gamma interaction in the second crystal. We added a normally distributed random value to the interaction times to account for the combined effects of time spread due to interaction depth, photo-detector, and readout electronics. We tuned the width of the normal distribution to achieve targeted CTR resolution between detector pairs. We did not take into account the intrinsic detector nor the acquisition dead time. The majority of simulations were performed on the Slovenian national super-computing network (SLING).

D. Image Reconstruction

We performed an iterative 3-D image reconstruction taking into account the TOF information using open-source CASToR—Customizable and Advanced Software for Tomographic Reconstruction [19]. We used MLEM—Maximum-Likelihood Expectation Maximization [20] optimization algorithm to reconstruct an image from the list-mode data obtained from the simulation. We only considered true coincidences (gammas originating from the same source and not scattered in the phantom) in the reconstruction not to deal initially in this pilot study with the impact of scatter and random event correction algorithm on the image quality.

Correction factors for normalization and attenuation correction were precomputed by CASToR and embedded in the data file used for reconstruction. We used the true attenuation map for computing attenuation correction factors. Images were additionally normalized by normalization factors, obtained by imaging a homogeneous box source with the size of the field-of-view. We obtained a minimum of $10^8$ true coincidences for each scanner design, and the normalization factors corresponded to the inverse values of the values in the reconstructed image, that was smoothed by a broad (15-mm FWHM) Gaussian filter.

We used an accelerated Siddon projector [21], a ray-tracing algorithm that computes the exact path length of a line through the voxels in most reconstructions, except for spatial resolution evaluation. Because artefacts (Moiré patterns) were observed when reconstructing the image onto a matrix with small voxel size using a Siddon projector, we used a distance-driven projector [22], based on computations of the overlap between a pair of detection elements and voxels, instead. Note that this makes spatial resolution methodology particular for comparing presented systems, rather than an absolute parameter that can be referenced and compared to existing systems.

E. Performance Evaluation

1) Sensitivity and NECR: We investigated the sensitivity, i.e., detection efficiency of a PET scanner, by imaging a point source at different positions in two directions: 1) y-direction (toward the panels or in the radial direction in the case of the reference scanner) and 2) z (axial) direction. The point source was moved in steps of 1 cm, and $10^7$ back-to-back gamma emissions were simulated per position. We determined the sensitivity as the ratio between the true event rate and source emission rate.

Noise equivalent count rate (NECR) was used for relating count rates to image signal-to-noise ratios. Following the NEMA NU 2-2018 standard, the phantom used for this study consisted of a line source of uniform activity inside of the 70-cm long polyethylene cylinder with a diameter of 20 cm. Because true, scatter and random count rates were accurately known from the simulation, we determined NECR as

$$NECR = \frac{T^2}{T + S + R}$$

(1)

where $T$, $R$, and $S$ represent the true, random, and scatter coincidence count rates, respectively. Those values can be affected by user-controlled parameters in an actual PET system [23]. In this work, we defined them in the following way: true coincidences were considered those having both their singles initiated from the same annihilation event, scatter coincidences were considered the true coincidences for which one of the two single photons (or both) interacted with the material before reaching the detector, and random coincidences were those for which the coincidence event was formed by two gamma rays from different annihilation events.

Because we did not take dead time into account, we determined NECR for different scanner designs at a single activity of 5 kBq/cm$^3$ and the acquisition contained at least $5 \cdot 10^5$ prompt counts as the NEMA standard suggests.

Assuming the dependence of gamma detection efficiency $\varepsilon$ on scintillator thickness $d$ as $\varepsilon = 1 - e^{-\mu d}$, where $\mu = 0.87$ cm$^{-1}$ is the linear attenuation coefficient of 511 keV photons in LSO, and quadratic dependence of true coincidence count rate on detection efficiency, $T \propto \varepsilon^2$, we can calculate the ratios between the rates for longer 10, 15, and 20 mm crystals relative to thin 5-mm crystals as 2.7, 4.3, and 5.5.

The standard NECR calculation does not take into account the SNR gain due to TOF information. To account for it, we used an effective NECR (NECR$_{TOF}$), usually defined as

$$NECR_{TOF} = \frac{D}{\Delta x} \cdot NECR = \frac{2D}{\Delta x} \cdot NECR$$

(2)

where $D$ is the radial dimension of the subject to be imaged and $\Delta x$ is the spatial uncertainty associated with the CRT—$\Delta t$ of the scanner. A 20-cm diameter ($D$) was used to evaluate and compare NECR$_{TOF}$ between different scanner designs in this study.

2) Spatial Resolution: The filtered back-projection (FBP) algorithm used in the NEMA standard is linear, making interpretation, and quantitative analysis of images more straightforward. However, FBP reconstructs only specific features of the original object and creates streak artefacts and missing boundaries in the limited angle tomography [24].
Reconstruction with iterative statistical algorithms can be difficult because they are nonlinear, and the non-negativity constraint can artificially enhance the apparent spatial resolution, but they can yield relevant resolutions for sources embedded in background activity and suitable source contrasts [25]. Taking this into account, we used the MLEM algorithm in the reconstruction, and hot sources were inserted into warm background.

For visual inspection of the resolution, we imaged a Derenzo phantom with hot rods in warm background. The phantom had six groups of rods with a diameter of 5.0, 4.0, 3.0, 2.5, 2.0, 1.5 mm, and the separation between the rods was the same as the rod diameter (Fig. 2). The activity concentration ratio between the hot rods and the background was 100:1, and the length of the rods was 4 cm.

We also quantitatively evaluated the resolution by imaging a hot rod with a diameter of 0.5 mm in the warm background, placed 1 cm above the center in the orthogonal direction to the detector plane. We performed the reconstruction on a matrix with $0.5 \times 0.5 \times 20 \text{ mm}^3$ sized voxels. We determined the FWHM of the line spread function in the $x$ direction (parallel to the two-panel detectors) and in the $y$ direction (perpendicular to the panel detectors) following the NEMA standard—FWHM was determined by linear interpolation between adjacent pixels at half the maximum value of the response function. We subtracted the average background, and then we determined the maximum value by a parabolic fit using the peak point and two nearest neighboring points, respectively.

3) Image Quality: The percent contrast and percent background variability were determined using NEMA NU 2-2018 phantom. Following the standard, we filled the (virtual) phantom with a sphere to background ratio of 4:1 and the background activity was set to 5.3 kBq/cm$^3$ (0.14 $\mu$Ci/cm$^3$). For contrast recovery calculations, we used true sphere masks as regions of interests (ROIs), and we also accounted for partial pixels. We positioned the phantom in the center of the scanner, and we simulated 1-min scans. We estimated the variance of contrast recovery coefficients and percent background variability from ten independently simulated 1-min scans. We determined the percent contrast and percent background variability from images that we reconstructed using MLEM with 50 iterations and we post-filtered the images with a Gaussian filter with FWHM of 5 mm.

4) Limited Angle PET System Applications: We used a 4-D extended cardiac-torso (XCAT) highly anatomically detailed phantom [26], [27] to evaluate the performance of a limited angular coverage system for imaging of humans. These phantoms were developed to provide accurate computerized models of human anatomy and physiology. We used the default male head and torso voxelized models in the simulation ($100 \times 100 \times 100$ matrix, $3 \times 3 \times 3 \text{ mm}^3$ voxels). The average activity in the phantom was 5.3 kBq/cm$^3$ corresponding to the background activity used in the NEMA image quality study, and we simulated 1-min scans.

We studied several two-panel and four-panel detector designs. For the two-panel system, we simulated the two panels at the sides of the head, parallel to the sagittal plane (Fig. 3). For torso imaging, we placed the plates below and above the torso. MLEM algorithm with 50 iterations, TOF data, and 5 mm FWHM Gaussian postfilter were used in the reconstruction. We reconstructed the datasets on a matrix with $3 \times 3 \times 3 \text{ mm}^3$ sized voxels.

We used two quantitative measures to evaluate the differences between the reference (simulation input) and the reconstructed images: 1) normalized root-mean-square error (NRMSE) and 2) the structural similarity index (SSIM). There are no consistent means of normalization of the root-mean-square error (RMSE). In this study, we used the mean of the measured data

$$\text{NRMSE} = \frac{1}{y} \left[ \frac{1}{n} \sum_{i=1}^{n} (y_i - x_i)^2 \right]$$

where $y_i$ is the intensity in the $i$th voxel of the reconstructed image, $\bar{y} = 1/n \sum_{i=1}^{n} y_i$ is the average intensity, and $x_i$ is the intensity in the $i$th voxel of the simulation input—ground truth image.

We used structure similarity index as a quantitative perceptual measure that accounts for patch-wise image statistics

$$\text{SSIM}(x, y) = \frac{(2\mu_x\mu_y + C_1)(2\sigma_{xy} + C_2)}{\mu_x^2 + \mu_y^2 + C_1}(\sigma_x^2 + \sigma_y^2 + C_2)$$

where mean ($\mu$), variance ($\sigma^2$), and covariance ($\sigma_{xy}$) represent local statistics in a patch of the reference image—$x$ and reconstructed image—$y$. $C_1$ and $C_2$ are stabilizing terms. Following [28], we used a sliding window with a side length of 11 to move voxel-by-voxel over the entire image, and each patch had its mean and variance spatially weighted by a normalized Gaussian kernel with a standard deviation of 1.5. An SSIM index map is obtained in this manner. NRMSE and the SSIM were applied to evaluate the quality of the reconstructed head region of the phantom, where the mask was derived from the input image. We used mean structural similarity index (MSSIM) index as a single overall quality measure of the entire reconstructed head region.

As another potential use case of a small field-of-view limited angular coverage system, we also imaged the thorax using the default XCAT male model ($120 \times 120 \times 120$ matrix, $3 \times 3 \times 3 \text{ mm}^3$ voxels). Note that, the panels were not centered on any specific organ. The XCAT phantom was imaged in a single bed position and was larger than the field-of-view of any particular scanner design we studied. NRMSE and MSSIM...
III. RESULTS

A. Sensitivity and NECR

We evaluated and compared the sensitivity and NECR of different scanner designs. Fig. 4 shows sensitivity in two directions for different scanner designs. For panel detectors, the readout was simulated at a crystal level, while for reference scanner, the readout was simulated at module level and also at crystal level for reference. The readout at the module level records roughly twice as many coincidences as the readout at the crystal level. With model-based readout, intercrystal scatterings can result in accepted events, increasing the detection efficiency at the cost of spatial resolution—the position (crystal) of the first gamma interaction is unknown, and the position of the event inside the module is therefore estimated with, e.g., centroid calculation.

Effective NECRs of different scanner designs are shown in Fig. 5. At a fixed CTR, ratios close to 3, 5, and 7 can be extracted from Fig. 5 between the NECR\textsubscript{TOF} values of 10, 15, and 20 mm long crystal scanner design, and the 5-mm long crystal design. The ratios are close to the estimates of 2.7, 4.3, and 5.5, obtained in Section II-E1. The number of events acquired in a PET scan is directly proportional to the average geometric coverage of the scanner, so it is not surprising that a four-panel scanner registers roughly twice as many events as the two-panel scanner.

B. Spatial Resolution

Fig. 6 shows transverse views of the reconstructed images of the Derenzo phantom for different scanner designs. In the two-panel design, the spatial resolution in the y-direction is worse than in the x-direction, and this difference is larger for scanners with poorer timing resolution or scanner designs with longer crystal. In the four-panel scanner design, the resolution is the same in the x and y direction, and it also appears to be quite homogeneous in the field-of-view. In contrast, in the
Fig. 6. Transverse views of the reconstructed images of the Derenzo phantom. The two panels were placed at the top and bottom. The MLEM reconstruction with 50 iterations was done on a matrix with $0.5 \times 0.5 \times 20$ mm$^3$ sized voxels.

Fig. 7. Spatial resolution in the $x$ and $y$ direction for different scanner designs, as a function of the scintillation crystal length $d$. Spatial resolution in the $x$-direction for a 2-panel system is represented by a single thick line for different CTRs as CTR does not notably affect spatial resolution in the $x$-direction.

two-panel design, a varying resolution within the field-of-view can be observed, especially noticeable is the degradation of resolution in the $y$-direction when moving in the $x$-direction away from the centre.

Fig. 7 shows spatial resolution in $x$ and $y$ direction for detector designs with different crystal lengths and different coincidence time resolutions. A clear degradation of spatial resolution in the $y$-direction can be observed with increasing crystal length in the case of a two-panel scanner design. CTR has a considerable impact on the spatial resolution only in the $y$ direction for a two-panel design, while CTR and also crystal length do not significantly impact spatial resolution in the four-panel design.

Fig. 8. Transverse views of the reconstructed images of the NEMA image quality phantom for different detector designs and for different CTRs. The images display the transverse section through the axial center of the hot spheres for different detector designs marked on the top of each image. A Gaussian postfilter with 5 mm FWHM was applied on all images. The number of events corresponds to the approximate number of detected true coincidences detected in 1 min by the scanner.

Fig. 9. Line profile with a cross section of a single voxel, obtained from reconstructed images of the NEMA image quality phantom for different detector designs. A Gaussian postfilter with 5 mm FWHM was used on the images. The profile passes through the 17 and 37 mm diameter hot spheres.

C. Image Quality

We reconstructed and compared the images of the NEMA image quality phantom acquired by different scanners. Fig. 8 shows transverse images of the reconstructed phantom for different scanner designs, and a profile through some of these images is shown in Fig. 9. One can notice a deterioration of
the image quality with decreased CTR and elongation of the spheres in the reconstructed images in the direction perpendicular to the panels in the two parallel panel design. The relations between percent contrast and background variability for different scanner designs are shown in Fig. 10. The relations were calculated from a series of images filtered by a Gaussian post-filter of different widths.

**D. Limited Angle PET System Applications**

As a possible application of the studied panel systems, we simulated images of the human head and torso. Fig. 11 shows reconstructed images of XCAT phantom of the head and neck region for several different panel detector designs and compares them to the image obtained with the reference scanner. Table II lists MSSIM and NRMSE for different scanner designs. MSSIM increases with improved timing resolution, while the NRMSE decreases. Filtering reduces both the MSSIM and the NRMSE. Reconstructed images of the thorax region are shown in Fig. 12. Note that, the imaged thorax digital phantom is larger than the field-of-view of any particular studied scanner design.
IV. DISCUSSION

In this work, we have investigated key performance characteristics of a limited angular coverage TOF PET system consisting of two or four flat-panel detectors with a CTR down to the intrinsic limit of current state-of-the-art PET scintillators of 50 ps by studying sensitivity, NECRs, spatial resolution, and image quality. Finally, we demonstrated the performance by imaging an XCAT digital human phantom. The sensitivity of panel detectors shows a strong positional dependence (Fig. 4). While the axial direction’s sensitivity decreases for both the panel designs and the reference scanner, panel designs also significantly lose sensitivity in the direction toward the panels (y-direction). As expected, the sensitivity does not vary that much in the radial direction for the reference scanner. We compared NECR TOF curves for different scanner designs (Fig. 5). The NECR TOF as a measure of image quality would predict better image quality for 2plates_20mm_100ps scanner compared to 2plates_10mm_75ps, contrary to what is observed (Fig. 10), indicating that the increase in NECR TOF does not reflect in the image quality. Distortions play an essential role in limited angle imaging which NECR TOF does not account for and is of limited use as a predictor of image quality in limited angle PET systems.

Spatial resolution was evaluated by imaging a hot rod near the centre, placed in a warm background, and reconstructed using the MLEM algorithm. Because evaluating spatial resolution at a single position in the FOV is not reflective of the performance of the system at other locations, we also imaged a Derenzo phantom to qualitatively evaluate the positional variation of the resolution (Fig. 6). In the two-panel setting, the spatial resolution in the y-direction (perpendicular to the panel detectors) is worse than in the x-direction (parallel to the two-panel detectors). The resolution in the y-direction can be improved by better timing resolution, however it can be severely degraded, by as much as a factor of 2 by increasing crystal length from 5 to 20 mm due to the parallax error (Fig. 7). The asymmetry in the spatial resolution results in elongation artifacts of hot spheres in the image quality study (Fig. 8). Due to increased angular coverage in the four-panel system, CTR or crystal length does not significantly affect the spatial resolution.

In the image quality study, we have imaged the NEMA phantom for an equivalent simulation time to match 1-min data collection with different scanner designs. Background variability decreases with improved CTR (Figs. 8 and 10). For two-panel designs, Fig. 10 shows similar contrast-to-background relations for designs with 10, 15, and 20 mm long crystals, indicating that improvement in setup efficiency is canceled out by the parallax error. The image quality of the 4panels_10mm_75ps scanner is similar to the image quality of the reference scanner.

We imaged the XCAT head and torso models as two possible applications of fast timing limited angle systems. The benefit of having a better timing resolution is demonstrated by evaluating the MSSIM and NRMSE in the reconstructed images of the XCAT phantoms (Table II). These quantitative
Fig. 12. Reconstructed images of the digital XCAT phantom of the thorax region for different scanners. Transverse (left column), coronal (middle column), and sagittal (right column) views are shown. Images have been reconstructed on a matrix with $3 \times 3 \times 3$ mm$^3$ voxels using MLEM—50 iterations. A Gaussian postfilter with FWHM of 5 mm was applied to all the images. The images were normalized with the activity within the $18 \times 18 \times 18$ cm$^3$ central region and scaled with the max voxel value in the true—reference image.

measures are usually not used to evaluate diagnostic PET images since the underlying reference distribution is often unknown but can be used to evaluate simulated images. MSSIM is a quantitative perceptional measure designed to correlate well with human subject ratings of image quality/similarity. The exact value of MSSIM depends on image complexity and on the imaging task and is therefore mainly used as a comparative measure. Both NRMSE and MSSIM agree well with the image quality study results as they predict similar images between the two-panel designs with 10, 15, and 20 mm long crystals and between 4panels_10mm_75ps scanner and the reference scanner (Table II).

To summarize, in the two-panel design, we observe the tradeoff between sensitivity and resolution, resulting in similar image quality between the designs with 10, 15, and 20 mm long crystals. The optimal crystal length should therefore be selected based on the specific clinical application and its performance requirements, i.e., high resolution or low noise images. Taking the cost and the dead-time effects into account, one might prefer to build a two-panel system with shorter crystals. Longer crystals have better detection efficiency but are impacted more by the parallax error, which reduces image quality. With a measurement of the depth of interaction, parallax error can be reduced or even eliminated at the cost of the complexity of the system (e.g., dual-ended readout [29]). A potential approach to correct for the parallax error or other resolution blurring effects such as insufficient angular sampling in the case of limited angle imaging is through image-based resolution modeling [30].

The simple, robust limited angular coverage systems we studied in our work can be made portable and suitable for bed-side/in-bed patient imaging, enabling their use in intensive care units (ICUs). They can be implemented in real-time PET-guided biopsy, intraoperative guidance and monitoring, simultaneous multimodality imaging, and also as in-beam PET scanners for monitoring the dose delivery in proton and heavy-ion therapy machines. Nontraditional PET scanner geometries similar to what we studied or scanners with significant gaps between detectors in the tangential and axial direction are also of interest for PET-MR hybrid systems as they permit novel ways of combining the PET and MR hardware components.

V. Conclusion and Prospects

Current clinical PET systems reach a CTR of several hundred ps FWHM. The experimental coincidence time resolution of 58(98) ps, measured with high-power electronics, of a single pair of 3(20) mm long LSO:Ce:Ca scintillation crystals coupled to silicon photomultipliers [3] cannot be easily scaled to large devices. Further disruptive design changes in both the silicon photomultipliers and the front-end electronics are necessary to achieve a large system CTR performance of 50 to 70 ps FWHM. For example, an extremely low input impedance FastIC, highly configurable 65 nm CMOS ASIC for fast timing applications [31], currently under initial tests, will enable to reduce electronics noise contribution compared to the present state-of-the-art technology. With the 3-D integration of the electronics and photosensors in a fully hybrid detector [32],
large area systems reaching an intrinsic CTR limit of inorganic scintillators will be possible. Based on those solid developments, our work is aimed to show that excellent image quality without noticeable distortions can be obtained with a simple, robust limited angular coverage TOF PET system consisting of two or four fast timing flat-panel detectors.

We have confirmed that TOF plays a crucial role in recovering sensitivity and improving resolution in the simplest two-panel design, especially in the direction perpendicular to the panels. Our simulations show that TOF resolution can partially compensate for a reduced gamma detection efficiency due to shorter crystals or smaller angular coverage. Studies with several phantoms suggest a comparable performance due to shorter crystals or smaller angular coverage. To confirm the potential modularity, where multiple panels can be combined, we present small-sized limited angle systems can be used in mobile bedside imaging of a small region-of-interest. A simple, fast timing flat-panel PET scanner is presented with multi-modality compatibility, where two or four fast timing flat-panel detectors. Our simulations show that TOF resolution can partially compensate for a reduced gamma detection efficiency due to shorter crystals or smaller angular coverage. Studies with several phantoms suggest a comparable performance due to shorter crystals or smaller angular coverage. We show a substantial degradation of spatial resolution with increased crystal length in the two-panel design due to the parallax error. Our results indicate that the depth-of-interaction information is crucial to improving spatial resolution in limited angular coverage systems with longer/standard crystal sizes (15–20 mm).

Presented small-sized limited angle systems can be used in mobile bedside imaging of a small region-of-interest. A mobile system built with two or more panels could be used in ICUs to perform diagnostic imaging of many different organs, such as the brain, heart, lungs, kidneys, liver, and assist in surgery and radiation therapy. We believe that a significant advantage of the panel detector design lies in its flexibility and potential modularity, where multiple panels can be combined to assemble longer axial-length PET scanners.

In the next steps, we plan to evaluate the performance of several limited angular coverage PET designs (e.g., torso and multiorgan/total body) and experimentally confirming their feasibility in a prototype device.

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REFERENCES

[1] S. Surti, “Update on time-of-flight PET imaging,” J. Nucl. Med., vol. 56, pp. 98–105, Jan. 2015.
[2] J. van Sluis, R. M. Turtos, E. Auffray, M. Paganoni, and P. Lecoq, “Performance characteristics of the digital biography vision PET/CT system,” J. Nucl. Med., vol. 60, no. 7, pp. 1031–1036, 2019.
[3] S. Gundacker, R. M. Turtos, E. Auffray, M. Paganoni, and P. Lecoq, “High-frequency SiPM readout advances measured coincidence time resolution limits in TOF-PET,” Phys. Med. Biol., vol. 64, no. 5, 2019, Art. no. 055012.
[4] P. Lecoq et al., “Roadmap toward the 10 ps time-of-flight PET challenge,” Phys. Med. Biol., vol. 65, no. 21, 2020, Art. no. 21BM01.
[5] S. Enoch, A. Gola, P. Lecoq, and A. Rivetti, “Design considerations for a new generation of SiPMs with unprecedented timing resolution,” J. Instrum., vol. 16, Feb. s2021, Art. no. P02019.
[6] S. Sajedi, L. Blackberg, S. Majewski, and H. Sabet, “Limited-angle TOF-PET for intraoperative surgical applications: Proof of concept and first experimental data,” 2021. [Online]. Available: arXiv:2104.03257
[7] Y. Yoshiyuki et al., “Development of a dual-head mobile DOI-TOF PET system having multi-modality compatibility,” in Proc. IEEE Nucl. Sci. Symp. Med. Imag. Conf. (NSS/MIC), 2014, pp. 1–3.
[8] G. Sportelli et al., “First full-beam PET acquisitions in proton therapy with a modular dual-head dedicated system,” Phys. Med. Biol., vol. 59, no. 1, pp. 43–60, 2014.
[9] Y. Shao et al., “In-beam PET imaging for on-line adaptive proton therapy: An initial phantom study,” Phys. Med. Biol., vol. 59, pp. 3373–3388, Jul. 2014.
[10] M. Gaond et al., “First results from all-digital PET dual heads for in-beam beam-on proton therapy monitoring,” IEEE Trans. Radiat. Plasma Med. Sci., early access, Dec. 2, 2020, doi: 10.1109/TRPMS.2020.3041857.
[11] M. C. Abreu et al., “Design and evaluation of the clear-PEM scanner for positron emission mammography,” IEEE Trans. Nucl. Sci., vol. 53, no. 1, pp. 71–77, Feb. 2006.
[12] R. R. Raylman et al., “The positron emission mammography/tomography breast imaging and biopsy system (PEM/PET): Design, construction and phantom-based measurements,” Phys. Med. Biol., vol. 53, no. 3, pp. 637–653, 2008.
[13] L. MacDonald, J. Edwards, T. Lewellen, D. Haseley, J. Rogers, and P. Kinahan, “Clinical imaging characteristics of the positron emission mammography camera: PEM flex solo II,” J. Nucl. Med., vol. 50, no. 10, pp. 1666–1675, 2009.
[14] Q. Xie, L. Wan, X. Cao, and P. Xiao, “Conceptual design and simulation study of an ROI-focused panel-PET scanner,” PLoS One, vol. 8, no. 8, pp. 1–9, 2013.
[15] Y. Gravel, Y. Li, and S. Matej, “Effects of TOF resolution models on edge artifacts in PET reconstruction from limited-angle data,” IEEE Trans. Radiat. Plasma Med. Sci., vol. 4, no. 5, pp. 603–612, Sep. 2020.
[16] S. Krishnamoorthy, B.-K. Teo, W. Zou, J. Mardonough, J. S. Karp, and S. Surti, “A proof-of-concept study of an in-situ partial-ring time-of-flight PET scanner for proton beam verification,” IEEE Trans. Radiat. Plasma Med. Sci., vol. 5, no. 5, pp. 694–702, Sep. 2021.
[17] S. Jan et al., “GATE: A simulation toolkit for PET and SPECT,” Phys. Med. Biol., vol. 49, no. 19, pp. 4543–4561, 2004.
[18] S. Agostinelli et al., “Geant4—a simulation toolkit,” Nucl. Instrum. Methods Phys. Res. A, Accelerators Spectrometers Detectors Assoc. Equip., vol. 506, no. 3, pp. 250–303, 2003.
[19] T. Merlin et al., “CASTOR: A generic data organization and processing code framework for multi-modal and multi-dimensional tomographic reconstruction,” Phys. Med. Biol., vol. 63, no. 18, 2018, Art. no. 185005.
[20] L. A. Shepp and Y. Vardi, “Maximum likelihood reconstruction for emission tomography,” IEEE Trans. Med. Imag., vol. 1, no. 2, pp. 113–122, Oct. 1982.
[21] F. Jacobs, E. Sundermann, B. De Sutter, M. Christiaens, and L. Lemahieu, “A fast algorithm to calculate the exact radiological path through a pixel or voxel space,” J. Comput. Inf. Technol., vol. 6, no. 1, pp. 89–94, 1998.
[22] B. D. Man and S. Basu, “Distance-driven projection and backprojection in three dimensions,” Phys. Med. Biol., vol. 49, pp. 2463–2475, May 2004.
[23] D. Nikolopoulos et al., “GATE simulation of the biograph 2 PET/CT scanner,” J. Nucl. Med. Radiat. Ther., vol. 6, no. 1, pp. 1–6, 2014.
[24] J. Friel and E. T. Quinto, “Characterization and reduction of artifacts in limited angle tomography,” Inverse Problems, vol. 29, no. 12, 2013, Art. no. 125007.
[25] K. Gong, S. R. Cherry, and J. Qi, “On the assessment of spatial resolution of PET systems with iterative image reconstruction,” Phys. Med. Biol., vol. 61, no. 5, pp. N193–N202, 2016.
[26] W. P. Segars, G. Sturgeon, S. Mendonca, J. Grimes, and B. M. W. Tsui, “4D XCAT phantom for multimodality imaging research,” Med. Phys., vol. 37, pp. 4902–4915, Sep. 2010.
[27] W. P. Segars, B. M. W. Tsui, J. Cai, F.-F. Yin, G. S. K. Fung, and E. Samei, “Application of the 4-D XCAT phantoms in biomedical imaging and beyond,” IEEE Trans. Med. Imag., vol. 37, no. 3, pp. 680–692, Mar. 2018.
[28] Z. Wang, A. C. Bovik, H. R. Sheikh, and E. P. Simoncelli, “Image quality assessment: From error visibility to structural similarity,” IEEE Trans. Image Process., vol. 13, pp. 600–612, 2004.
[29] M. Ito, S. J. Hong, and J. S. Lee, “Positron emission tomography (PET) detectors with depth-encoders (DOE) capability,” Biomed. Eng. Lett., vol. 1, pp. 70–81, Jun. 2011.
[30] S. Matej, Y. Li, J. Panetta, J. S. Karp, and S. Surti, “Image-based modeling of psf deformation with application to limited angle pet data,” IEEE Trans. Nucl. Sci., vol. 63, no. 5, pp. 2599–2606, Oct. 2016.
[31] R. Ballabriga, M. Campbell, and D. Gascón, FastIC and FastICpix Developments, Newslett. CERN EP Dept., Geneva, Switzerland, 2020.
[32] D. Gascón et al., “FastICpix: Integrated signal processing of active hybrid single photon sensors with ps time resolution,” in Proc. ATTRACTION Final Conf., 2020.