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Optimization of the Performance of Segmented Scintillators for Radiotherapy Imaging through Novel Binning Techniques

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Abstract

Thick, segmented crystalline scintillators have shown increasing promise as replacement x-ray converters for the phosphor screens currently used in active matrix flat-panel imagers (AMFPIs) in radiotherapy, by virtue of providing over an order of magnitude improvement in the DQE. However, element-to-element misalignment in current segmented scintillator prototypes creates a challenge for optimal registration with underlying AMFPI arrays, resulting in degradation of spatial resolution. To overcome this challenge, a methodology involving the use of a relatively high resolution AMFPI array in combination with novel binning techniques is presented. The array, which has a pixel pitch of 0.127 mm, was coupled to prototype segmented scintillators based on BGO, LYSO and CsI:Tl materials, each having a nominal element-to-element pitch of 1.016 mm and thickness of ~1 cm. The AMFPI systems incorporating these prototypes were characterized at a radiotherapy energy of 6 MV in terms of MTF, NPS, DQE, and reconstructed images of a resolution phantom acquired using a cone-beam CT geometry. For each prototype, the application of 8×8 pixel binning to achieve a sampling pitch of 1.016 mm was optimized through use of an alignment metric which minimized misregistration and thereby improved spatial resolution. In addition, the application of alternative binning techniques that exclude the collection of signal near septal walls resulted in further significant improvement in spatial resolution for the BGO and LYSO prototypes, though not for the CsI:Tl prototype due to the large amount of optical cross-talk resulting from significant light spread between scintillator elements in that device. The efficacy of these techniques for improving spatial resolution appears to be enhanced for scintillator materials that exhibit mechanical hardness, high density and high refractive index, such as BGO. Moreover, materials that exhibit these properties as well as offer significantly higher light output than BGO, such as CdWO₄, should provide the additional benefit of preserving DQE performance.

Keywords

Megavoltage cone-beam CT; flat-panel imager; electronic portal imaging device; segmented crystalline scintillators; high x-ray detection efficiency

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I. INTRODUCTION

The accurate delivery of external beam radiation therapy relies on imaging techniques that permit adequate identification and tracking of anatomical details in the patient. These techniques are intended to facilitate image-guided radiation therapy (IGRT), which aims at maximizing radiation dose to tumor volumes while minimizing dose to surrounding normal tissues and critical organs. Toward achieving this goal, x-ray imaging techniques that provide 3D information in the treatment room are routinely employed to separate anatomical clutter and provide visualization of soft-tissue structures – allowing assessment of anatomical changes over the course of radiation treatment (Barker *et al.*, 2004). Such techniques have been introduced to complement or even supplant portal imaging. Portal imaging involves the use of the megavoltage (MV) treatment beam to produce 2D projection images and is typically performed with electronic portal imaging devices (EPIDs) based on the technology of active matrix flat-panel imagers (AMFPIs) (Antonuk, 2002). A 3D imaging technique that has seen widespread clinical implementation involves the acquisition of cone-beam computed tomography (CBCT) images using a kilovoltage (kV) x-ray source and an AMFPI, both mounted orthogonally to the treatment gantry (Jaffray and Siewerdsen, 2000; Jaffray *et al.*, 2002).

While kV CBCT is a very clinically useful technique, it suffers from a number of disadvantages that could be addressed through an alternative approach in which the portal imager is used to acquire 3D images in a CBCT geometry using the MV treatment beam (Lewis *et al.*, 1992; Mosleh-Shirazi *et al.*, 1998; Seppi *et al.*, 2003; Sillanpaa *et al.*, 2006; Monajemi *et al.*, 2006; Rathee *et al.*, 2006; Morin *et al.*, 2006; El-Mohri *et al.*, 2011). While the advantages of this MV approach (including reduced streak artifacts caused by metal implants) would be beneficial (Pouliot *et al.*, 2005; Yin *et al.*, 2005), conventional MV AMFPIs unfortunately require relatively large doses to achieve clinically useful soft-tissue contrast resolution (Groh *et al.*, 2002). This limitation is a consequence of the rather low detection efficiency of the x-ray converter (~2% at 6 MV), which typically consists of a relatively thick $\text{Gd}_2\text{O}_2\text{S}:\text{Tb}$ phosphor screen (e.g., ~133 mg/cm²) coupled to a Cu plate. The resulting zero-spatial-frequency detective quantum efficiency (DQE) values of only ~1% (El-Mohri *et al.*, 2001) are considerably lower than maximum DQE values of over 70% provided by kV AMFPIs used for diagnostic imaging (Marshall *et al.*, 2011).

Currently, the potential for overcoming the DQE limitations of conventional MV AMFPIs through substitution of the phosphor screen converter by a thick, segmented scintillator is under investigation. Such scintillators consist of a two-dimensional matrix of optically-isolated scintillator elements. Small-area prototypes of such scintillators employing CsI:Tl and $\text{Bi}_4\text{Ge}_3\text{O}_{12}$ (BGO) material have demonstrated DQE values up to ~25% (Wang *et al.*, 2009) – considerably greater than that of conventional MV AMFPIs. As a result of such significant improvement in DQE, a prototype MV AMFPI based on an ~11 mm thick segmented BGO scintillator has demonstrated visualization of low contrast objects from reconstructed MV cone-beam CT images at a scan dose of only ~4 cGy (El-Mohri *et al.*, 2011) – an amount of radiation comparable to that required for a single portal image obtained from a conventional MV AMFPI.

The performance enhancement provided by these segmented scintillator prototypes was found to be most pronounced at low spatial frequencies. At higher spatial frequencies, the enhancement is primarily limited by the following factors: (a) radiation scatter; (b) less-than-optimal optical isolation provided by the septal walls that separate the scintillator elements – causing optical cross-talk; (c) beam divergence effects for off-central-axis locations which become progressively more pronounced as scintillator thickness exceeds ~10 mm (Wang *et al.*, 2010; Liu *et al.*, 2012); and (d) *misregistration* of the scintillator elements with the underlying AMFPI array pixels. While the effects of radiation scatter can be somewhat reduced (and the x-ray detection efficiency increased) through introduction of metal into the septal wall construction (Sawant *et al.*, 2005), the poor optical reflectivity of metals that can be practically incorporated into wall construction (such as aluminum) favors metal-less walls constructed with plastic sheets and glue that minimize cross-talk and maximize reflectivity (Krus *et al.*, 1999; Sawant *et al.*, 2005). In addition, the effects of beam divergence can be partially countered through incorporation of focused scintillator elements (Liu *et al.*, 2012). With regard to misregistration, the realization of good registration is challenging for a number of reasons. For example, it requires an accurate and robust means to achieve and maintain registration of the scintillator with the array. However, even with good mechanical engineering design to precisely locate and immobilize the relative orientation of the array and scintillator, the achievable degree of registration will still be limited by the extent to which the elements in the scintillator itself form a perfectly regular matrix – a quality that shall be referred to as *alignment*. In the case of the segmented scintillator prototypes developed for our research, the misalignment that occurs during fabrication is, in some instances, considerable. Such misalignment is particularly noticeable when compared with the almost perfect alignment of the pixel matrices of AMFPI arrays that are fabricated through highly precise photolithographic techniques. As a consequence, for a scintillator with a pitch equal to that of the AMFPI array, matching the location of the scintillator elements to that of the underlying array pixels is challenging, and any resulting misregistration leads to optical light emitted by a given scintillator element being detected by more than one array pixel. Such unintended sharing of the optical signal results in spatial resolution loss, potentially affecting DQE performance. Although some improvement in alignment beyond that achieved by the best of our prototypes may be possible, it is uncertain as to whether full-size scintillators suitable for ~40×40 cm² portal imagers would be sufficiently aligned to render this problem moot – providing motivation for exploring methods of overcoming the problem of misregistration.

In this paper, the problem of spatial resolution loss caused by misregistration is explored. The effects of misregistration on spatial resolution are examined by means of a theoretical model as well as through empirical investigation of the performance of various segmented scintillator prototypes. Techniques to mitigate the effects of misregistration are described and the effectiveness of the application of these techniques in recovering spatial resolution is reported.

II. METHODS

II.A Segmented scintillator prototypes

The segmented scintillator prototypes employed in this study have designs similar to those reported in previous publications and were produced as part of a program of research to create optimized, large-area segmented scintillators for radiotherapy imaging (Sawant *et al.*, 2006; Wang *et al.*, 2009; El-Mohri *et al.*, 2011). Specifications for these prototype detectors are given in Table I. The prototypes (manufactured by Saint-Gobain Crystals, OH) include an ~11.3 mm thick BGO detector, an ~11.4 mm thick CsI:Tl detector and a pair of ~10.3 mm thick $\text{Lu}_{1.8}\text{Y}_{0.2}\text{SiO}_4\text{:Ce}$ (LYSO) detectors – referred to as BGO-50, CsI-50, LYSO-50 and LYSO-115, respectively. Each detector consists of a two-dimensional matrix of 120×60 scintillator elements with a nominal element-to-element pitch of 1.016 mm, resulting in an active area of $\sim 122 \times 61 \text{ mm}^2$. Each element consists of a scintillator crystal laterally surrounded by septal walls, which generally comprise a polymer reflector sandwiched by thin layers of transparent glue. Two septal wall thicknesses were used in the present segmented scintillator prototypes: ~0.05 mm for BGO-50, CsI-50 and LYSO-50, and ~0.115 mm for LYSO-115. The choice of septal wall thickness is an important design parameter as it represents a tradeoff between optical isolation and fill factor (defined as the fraction of detector volume that efficiently absorbs radiation and generates light). While thinner septal walls advantageously provide larger fill factors (and thus higher DQE), they generally result in less effective optical isolation (and thus lower spatial resolution). Beyond this tradeoff, the examination of two different septal wall thicknesses was undertaken in the spirit of exploring the dependence of element-to-element alignment on this parameter.

For purposes of empirical performance assessment, each segmented scintillator prototype was paired with an indirect detection AMFPI array operated with a set of custom-made acquisition electronics (Huang *et al.*, 1999). This array, the performance of which has been previously characterized (Antonuk *et al.*, 2009), has a pixel format of 1024×1024 , a pixel pitch of 0.127 mm, and an optical fill factor of ~80%. The combination of this relatively high resolution AMFPI array with the comparatively coarser pitch prototypes was employed in an earlier study investigating the performance of thick, segmented scintillators for MV CBCT imaging (El-Mohri *et al.*, 2011). With a pitch that is smaller by exactly a factor of eight compared to that of the prototypes, the array enables the acquisition of x-ray images that allow visualization of the detailed structure of the segmented scintillator elements. Such a high degree of over-sampling was previously found to simplify the process of optimal registration during imager assembly, making only angular alignment necessary and eliminating the need for further mechanical registration of the scintillator elements with the array pixels (Wang *et al.*, 2009). Since spatial resolution is nominally limited by the resolution of the prototype scintillator and not by that of the finer pitch AMFPI array, it is appropriate to sum the signals from groups of 8×8 array pixels (a procedure referred to as “binning”) in order to achieve a sampling pitch consistent with the 1.016 mm element-to-element pitch of the prototypes. Furthermore, in the current study, the ability to visualize the structure of the elements is used to quantify the degree of alignment of the elements of each prototype by means of a metric, as well as to facilitate increasingly sophisticated binning techniques intended to improve some aspect of prototype performance. These metric and

binning techniques are introduced in conjunction with the presentation of results obtained from the prototypes in Section III.A.

II.B Measurement methods

Data acquisition for all prototype measurements was performed using a 6 MV treatment beam from a Varian 21EX radiotherapy linear accelerator (LINAC). The radiation output of the LINAC, measured in monitor units (MU), is calibrated so that, for each MU, a dose of 0.8 cGy is delivered at a 10 cm water depth, at a source-to-detector distance (SDD) of 100 cm, and for a field of $10 \times 10 \text{ cm}^2$. Data was acquired for the determination of the modulation transfer function (MTF), noise power spectrum (NPS) and for CBCT imaging. For NPS and CBCT, and for MTF, SDD was set to $\sim 130 \text{ cm}$ and 138 cm , and the x-ray field size was limited to $13 \times 9 \text{ cm}^2$ and $2 \times 8 \text{ cm}^2$ at the isocenter, respectively.

In the case of the NPS and CBCT measurements, the LINAC was operated at a dose rate of 100 MU/min, which delivers ~ 36 beam pulses per MU, corresponding to a dose of $\sim 0.022 \text{ cGy}$ per pulse. (The case of MTF measurements is discussed in the next section.) At this dose rate, the $5 \mu\text{s}$ long beam pulses were generated at a pulse frequency of 60 Hz, corresponding to a time interval of $\sim 16.7 \text{ ms}$ between consecutive pulses. These pulses were used to trigger array readout, allowing synchronization between radiation delivery and image acquisition in fluoroscopic mode. The acquisition of image frames at the lowest available dose (1 beam pulse) was achieved by reading out the array within the $\sim 16.7 \text{ ms}$ time interval between consecutive beam pulses. However, due to readout speed restrictions imposed by the acquisition system, which requires $\sim 113 \text{ ms}$ for full array readout, it was possible to address only a portion of the array between beam pulses – a contiguous area of 1024×120 pixels corresponding to the central part of the scintillator. For NPS, data frames were obtained under two irradiation conditions corresponding to the delivery of a constant number of beam pulses (1 and 4) per frame. Since the pulse rate was fixed at $\sim 60 \text{ Hz}$, achieving a radiation dose per frame corresponding to 1 and 4 pulses required triggering array readout every ~ 16.7 and 66.7 ms , respectively. For CBCT, data frames were obtained with only a single beam pulse per projection image.

For each prototype, close, uniform physical contact with the array was achieved by virtue of the high degree of planarity of the opposing array and scintillator surfaces and the non-negligible weight of the scintillator, without the use of an additional coupling medium. A 1 mm Cu plate was positioned on top of the scintillator with a black reflector sandwiched in between. While the reflector was introduced to optimize spatial resolution (Wang *et al.*, 2009), the Cu plate acts as a radiation buildup layer as well as an absorber of scattered radiation. For all measurements involving the BGO prototype, a total of 2000 MUs were used to irradiate the scintillator immediately prior to the start of the data acquisition. This pre-irradiation was necessary in order to bring the scintillator into a state of stable light output that is independent of accumulated dose (Wang *et al.*, 2009).

II.B.1 MTF, NPS and DQE—Spatial resolution of AMFPIs incorporating the various segmented scintillator prototypes was characterized in terms of the presampled MTF, using the angled slit technique (Fujita *et al.*, 1992; Sawant *et al.*, 2006; Wang *et al.*, 2009). For

these measurements, a custom-made slit, consisting of two adjoining blocks of tungsten, each having dimensions of $4.25 \times 8.5 \times 19 \text{ cm}^3$, separated by a narrow gap of 0.1 mm (Sawant *et al.*, 2007) and positioned so as to provide an entrance surface of $\sim 119 \text{ cm}$ from the radiation source and a SDD of $\sim 138 \text{ cm}$, was used. The slit was oriented in such a way that the narrow gap formed an $\sim 3^\circ$ angle with respect to one direction of the AMFPI array. The LINAC was operated at a dose rate of 600 MU/min with beam delivery controlled by the data acquisition electronics. For each measurement, five radiographic images of the slit were obtained with the slit positioned at the center of the field. After averaging the images, correcting for unintended radiation penetration (Wang *et al.*, 2009) and application of gain and offset corrections, the resulting slit image yielded a line-spread function (LSF). The MTF was then obtained from the Fourier transform of that LSF. Note that for all empirical image acquisitions (including those for MTF, NPS and CBCT), raw images were first filtered using a 3×3 median filter to remove pixel and line defects, followed by pixel binning.

The methodology used to measure NPS follows that described in previous publications (Siewerdsen *et al.*, 1998; El-Mohri *et al.*, 2001) and is summarized as follows. For each prototype, a total of 1000 consecutive radiation flood frames, with 1024×120 pixels per frame acquired in fluoroscopic mode under irradiation conditions corresponding to 1 and 4 beam pulses per frame, were used in the NPS analysis. Each frame was then cropped to a smaller region of interest away from the edges of the scintillator and subsequently binned. Gain and offset corrections were then applied to the flood frames, resulting in processed images consisting of 100×14 pixels. The pixel signal, measured in analog-to-digital converter (ADC) units, was then converted to electrons (where $1 \text{ ADC} = 7400 \text{ e}$). NPS was determined from the processed flood frames using the synthesized slit technique (Dainty and Shaw, 1974; Giger *et al.*, 1984; Maidment and Yaffe, 1994). For each scintillator, irradiation condition and binning technique, the processed flood frames were averaged along the short dimension, forming 1000 independent, 1D realizations. After subtraction of low-frequency background trends and the application of a Hanning window function, a Fourier transform was applied to each of the realizations. The results were then normalized (Siewerdsen *et al.*, 1998) to produce power spectra, the average of which yielded NPS. A correction for lag, estimated to be $\sim 3\%$, was applied to each NPS, resulting in the measured NPS (Granfors *et al.*, 2003). The normalized NPS (NNPS) was determined from

$$NNPS(f) = \frac{NPS(f) \times \bar{q}_0}{\bar{S}^2}, \quad (1)$$

where $\bar{S}$ is the average signal and $\bar{q}_0$ is the incident x-ray fluence for the corresponding irradiation condition. The value of $\bar{q}_0$, $\sim 245,400 \text{ X rays/mm}^2$ per beam pulse at a SDD of 130 cm, was obtained through radiation transport simulation of the LINAC calibration conditions at 1 MU. For each scintillator and irradiation condition, the DQE was determined from the measured NPS and MTF using

$$DQE(f) = \frac{MTF^2(f)}{NNPS(f)}. \quad (2)$$

II.B.2 CBCT image acquisition—To explore the effectiveness of various binning techniques in restoring loss of spatial resolution due to misalignment, a spatial resolution phantom was scanned in a CBCT geometry. The setup used for acquisition of the tomographic images has been described previously (El-Mohri *et al.*, 2011) and is only briefly summarized below. Figure 1 shows the bench-top scanning system comprising an aluminum frame used to support a cylindrical water-equivalent rod onto which the phantom is inserted. The phantom is made of an epoxy mix and contains 2 mm-thick aluminum contrast line-pair inserts with resolution groups ranging from 1 to 21 lp cm⁻¹ (High Resolution Module, CTP528, The Phantom Laboratory, Salem, NY). During image acquisition, the phantom was rotated around its axis above the scintillator while the LINAC gantry was kept at a fixed position. The rotation of the phantom was controlled by a stepper motor, which was operated at a constant speed asynchronously with the AMFPI array readout. The speed of the motor was set so that, given the need to obtain 1 projection image per beam pulse (i.e., every ~16.7 ms), 180 projection images were obtained for a 360° scan. For each tomographic acquisition, 4 scans, obtained from a total of 720 projection images, were averaged to produce a single scan corresponding to a dose of ~16 cGy. A Feldkamp-based algorithm employing a ramp filter was used to reconstruct the spatial distribution of attenuation coefficients for the phantom using gain and offset-corrected projection and flood images (Wang *et al.*, 2008). The reconstructed voxel size and slice thickness were set to 1.016 mm, matching the pitch of the scintillator prototypes.

II.C Calculational models

II.C.1 Model to quantify the effects of misregistration—The extent of spatial resolution degradation that is created by misregistration was quantified by means of a model based on simple geometrical considerations. This geometrical model entails creation of images of an object by projecting a uniform, 2D intensity map onto a segmented scintillator that overlies an AMFPI array. Each of the scintillator and the array is represented by a 2D matrix of equal-size squares surrounded by regularly-spaced gaps. The squares represent the active components, which are the scintillator crystals and the photodiodes for the segmented scintillator and the AMFPI array, respectively, while the gaps represent regions in which there is no signal integration. In the model, the dimensions chosen for the scintillator crystals, photodiodes, and gaps are based on those of the prototype scintillators and the AMFPI array used in the study. (For the scintillator, the gap was set to 0.05 mm.) The projection of an object onto the overlapping matrices representing the scintillator and the AMFPI array was performed in such a way that signal transmitted from the elements to the pixels was integrated only within the active areas, discarding any signal recorded in the gaps. The resulting image was subsequently binned to a pitch of 1.016 mm, corresponding to that of the scintillator. Given an exact factor of 8 in the ratio of the pitch of the scintillator elements to that of the array pixels, ideal registration can be obtained – a configuration referred to as “0 pixel offset”. To simulate misregistration, deliberate shifts away from ideal registration, by multiples of 0.127 mm (configurations referred to as 1 “pixel offset”, “2 pixel offset”, etc), leads to binning of signal across adjacent elements – an undesirable situation referred to as *cross-binning*.

Using this model, spatial resolution degradation was quantified through determination of the MTF and qualitatively demonstrated through reconstructed cone-beam CT images of a resolution phantom. For the MTF determination, an object in the form of a narrow, 5 μm wide line with a Gaussian profile was projected onto the overlapping scintillator and array at an angle of $\sim 2.7^\circ$ relative to their matrices. Projected images of such an object, which represents a radiation slit, were first binned to 1.016 mm and then used to obtain LSFs employing the angled-slit technique. From these LSFs, MTFs were determined for the various configurations. For the CBCT imaging demonstration, an object simulating a resolution phantom with a diameter of 12 cm and a set of 8 line-pair inserts corresponding to resolutions ranging from ~ 1 to 5 lp cm^{-1} was used. For each configuration, 180 projection images were obtained by rotating the object around its own axis in angular steps of 2° . From these images, reconstructed CT images were obtained using analysis techniques similar to that described in section II.B.2

II.C.2 Simulations to model the effect of radiation transport—To compare the empirical performance of the segmented scintillator prototypes to theoretical predictions, Monte Carlo simulations of radiation transport were performed using EGSnrc (Kawrakow and Rogers, 2000) with the EGSnrc C++ class library (Kawrakow, 2005). The simulation methods are similar to those previously reported in (Liu *et al.*, 2012) and only a brief description of these methods follows. The simulations were used to determine values for MTF and DQE which, in the absence of accounting for optical effects, represent theoretical upper limits to performance. Each simulated detector consists of a segmented scintillator covered by a 1 mm Cu plate. The AMFPI array was not included in the simulation since its contribution to radiation transport is negligible. The x-ray source (a 6 MV photon beam representing the spectral output of a Varian LINAC (Sheikh-Bagheri, 1999)) was simulated as a point located 130 cm away from the detector. The scintillator size and thickness, element-to-element pitch, crystal material, and septal wall thickness were chosen to correspond to those of the prototypes – with septal wall material assumed to consist of polystyrene. The simulated scintillators were assumed to be ideal with no misalignment. Radiation deposition in the scintillator crystals was tallied, resulting in image frames from which MTF and NNPS were obtained.

For the MTF simulation, a radiation beam defining a narrow ($62 \times 0.004 \text{ mm}^2$) slit was perpendicularly incident over the center of each detector, at a small ($\sim 2^\circ$) angle with respect to the shorter direction of the scintillator's matrix. The resulting image of the slit was used to extract the LSF employing the angled slit technique. The Fourier transform of the LSF yielded the simulated MTF. For the NNPS simulation, a $123 \times 62 \text{ mm}^2$ radiation beam covering the entire detector was used to generate 600 image frames – with 10 million x-ray histories per frame. Each frame was divided into 3 non-overlapping blocks of 100×14 elements corresponding to the central part of the detector, and each block was used to generate a NPS using the synthesized slit technique (see section II.B.1). NNPS was determined from Eq. 1 using the average NPS and an x-ray fluence of $\sim 1300 \text{ X rays/mm}^2$. DQE was calculated from Eq. 2 using the simulated MTF and NNPS. Note that the resulting values of DQE do not account for the effect of electronic additive noise, making these results independent of dose.

III. RESULTS

III.A Quantification of prototype alignment and description of binning techniques

An example of an image obtained with the LYSO-50 prototype using the radiotherapy beam is shown in Fig. 2a. In this image, the scintillator elements as well as the septal walls are clearly visible – with the relatively brighter septal walls displaying a distinctly higher signal. The availability of over-sampled images such as that of Fig. 2a facilitates numerous possibilities (64 in total) for assigning contiguous groups of 8×8 pixels to the scintillator elements. However, the arbitrary selection of any one of these possibilities may degrade spatial resolution as a result of element-to-element misalignment. To illustrate the degree of misalignment that can occur, a regular grid based on a period corresponding to the nominal pitch of the elements is superimposed on the image of Fig. 2a. While this grid overlaps quite well with the septal walls in the enlarged region shown in Fig. 2b, it shows a significant offset in the enlarged region shown in Fig. 2c – a clear indication of element-to-element misalignment. The offset in Fig. 2c corresponds to a shift of one and two lines of pixels in the vertical and horizontal directions, respectively.

Given the existence of misalignment, it is desirable to identify that choice of 8×8 binning which provides the best achievable registration (which shall be referred to as *optimal*) between scintillator elements and array pixels so as to reduce the amount of cross-binning. Toward this goal, a metric that quantifies the overall alignment of a given scintillator was defined as the percentage overlap of two binary masks: a mask obtained from the corresponding x-ray images (such as that in Fig. 2a) that represents the location of wall segments of individual elements; and a perfect mask representing the location of the wall segments of a matrix of regularly spaced elements with an element-to-element pitch of 1.016 mm. Since misalignment is not necessarily uniform across the surface of a given scintillator, it was necessary to determine this percentage for all possible overlaps between the two masks using 1 pixel shifts in the relative position of the masks in each spatial direction, resulting in 64 values of alignment. Figures 3a and 3b show a plot of these values for prototypes LYSO-115 and LYSO-50, respectively. For each prototype, the maximum value of the metric represents an estimate of the overall alignment. In the case of LYSO-115, a maximum value of ~86% is observed (indicative of a high degree of alignment) while a maximum of ~63% is found for LYSO-50. Images illustrating the overlapping septal wall segments that correspond to these maximum values are shown in Fig. 4. While LYSO-115 (Fig. 4a) exhibits a large fraction of well-aligned elements in both directions, prototype LYSO-50 (Fig. 4b) shows a considerably lower fraction in the horizontal direction – a consequence of a systematic shift that is apparent in Fig. 2c. For BGO-50 and CsI-50, the maximum values of the alignment metric were found to be ~74% and ~71%, respectively – with the results summarized in Table II. Digital binning of the imaging signal from groups of 8×8 array pixels, corresponding to that choice of mask overlap which maximizes the alignment metric, results in improved *digital registration* – creating the best possible match of 8×8 grouped pixels with individual scintillator elements, hereafter referred to as *optimal 8×8 binning*.

As described above, the creation of an alignment metric facilitates identification of optimal 8×8 binning. However, depending on the extent of element-to-element misalignment in a given scintillator, such binning results in varying degrees of effectiveness in improving registration – falling short of complete elimination of misregistration and the resulting cross-binning of signal. This motivates the creation of yet other binning techniques to overcome the problem. These additional techniques employ the oversampled information provided by the high resolution AMFPI array, which (as for the technique described above) allows identification of septal wall locations. For a given segmented scintillator prototype, such information is used to perform an alternative form of binning involving the selection of a subset of contiguous array pixels from *within* the block of pixels defining every element in the scintillator. Two types of such *selective binning* were employed in this study. The first involves selection of a *fixed* number of central contiguous pixels (i.e., 3×3 , 4×4 or 6×6) for every element, referred to as “*inner binning*”. The second type involves selection of all pixels inside every element, excluding only those corresponding to septal wall locations – a technique referred to as “*smart binning*”. For both techniques, cross-binning is thus avoided, resulting in the *elimination* of misregistration. The various binning techniques employed in this paper are summarized in Table III.

III.B Simulated performance results

Figure 5 shows MTF results obtained using the geometrical model for various 8×8 binning configurations. The 0 pixel offset configuration, which corresponds to the ideal case where each group of binned pixels is exactly registered with the overlying scintillator element, results in an MTF that overlaps with the sinc function corresponding to the aperture size defining the active area of the scintillator elements. As the binning offset is increased up to half the element pitch (i.e., with the deliberate introduction of cross-binning which corresponds to poorer registration) MTF is seen to progressively degrade, approaching a performance equivalent to that of a system with a spatial resolution half that of the ideal case. Such degradation of spatial resolution due to misregistration is evident in the reconstructed images of the simulated resolution phantom shown in Fig 6, where resolution gradually degrades with increasing offset. The results of Figs. 5 and 6 illustrate the importance of proper registration of the scintillator elements with the underlying binned pixels.

III.C Empirical performance results

III.C.1 MTF—Figure 7 shows MTF results obtained using various binning techniques for the four prototype segmented scintillators. For optimal 8×8 binning, BGO-50 exhibits the highest MTF while CsI-50 exhibits the lowest. These results, some of which are similar to previously reported results obtained from the same prototype scintillators but with a coarser pitch AMFPI array (Wang *et al.*, 2009), demonstrate the advantage provided by BGO over CsI:Tl by virtue of the higher radiation stopping power (which reduces radiation scatter) and higher refractive index (which restricts light spread) of the former material, leading to superior spatial resolution performance.

For LYSO-50 and LYSO-115, despite the similarity of their simulated radiation MTF results, LYSO-115 exhibits superior empirical MTF with optimal 8×8 binning due to a

combination of better alignment (which reduces cross-binning) and better optical isolation between scintillator elements (which reduces cross-talk). For BGO-50 and LYSO-50, despite their similar simulated radiation MTFs and identical septal wall thicknesses, BGO-50 exhibits superior empirical MTF with optimal 8×8 binning. While this is partially the result of somewhat better alignment, a more important influence is the reduction of cross-talk due to a greater degree of refractive index mismatch between the scintillator material and the glue in the septal walls. (The glue has a refractive index of ~ 1.55 .) For BGO-50, LYSO-50 and LYSO-115, the use of inner 4×4 and smart binning provides significant improvement in MTF compared to optimal 8×8 binning. This is a result of the complete elimination of cross-binning caused by element-to-element misalignment, as well as the exclusion of near-septal wall signal (likely containing photons originating from cross-talk). In the case of BGO-50, inner 4×4 and smart binning both result in MTF performance approaching that of a conventional MV AMFPI, but remaining below the upper limit defined by the simulated radiation MTF. For LYSO-50 and LYSO-115, the improvement with inner 4×4 and smart binning is more modest due to the inherently inferior spatial resolution of these prototypes resulting from smaller refractive index mismatch.

For CsI-50, the empirical MTF obtained with optimal 8×8 binning is much poorer than that of the other prototypes. Given that the alignment of CsI-50 is in the general range of that for the other prototypes, this poor MTF is suspected to be largely a result of a much higher degree of cross-talk. Moreover, the application of neither inner 4×4 nor smart binning provides any noticeable improvement to spatial resolution compared to that obtained with optimal 8×8 binning. This supports the idea that the contribution of misalignment to the empirical MTF for optimal 8×8 binning is relatively small compared to the effect of cross-talk. It also indicates that, unlike for the other prototypes, the exclusion of septal wall signal for CsI-50 does not remove a significant enough portion of optical photons originating from cross talk to improve MTF. The different behavior of the CsI-50 prototype is illustrated by the fact that the signal from septal walls is lower than that from crystals (El-Mohri *et al.*, 2011), whereas the reverse is true for the BGO-50, LYSO-50, and LYSO-115 prototypes. For CsI-50, given that the various binning techniques were equally ineffective in improving spatial resolution, and given that NNPS and DQE results for that prototype have previously been published (Wang et al 2009), further investigations were not performed.

III.C.2 Signal and NNPS—Figure 8 shows signal collected using various binning techniques as a function of radiation dose for BGO-50 and LYSO-50. Results for LYSO-115 are similar to those for LYSO-50 and so are not shown. While the magnitude of the signal for inner 6×6 binning relative to that for inner 3×3 binning is proportional to the ratio of the number of binned pixels for those respective techniques, optimal 8×8 binning results in disproportionately higher signal due to a greater relative signal contribution from the septal walls. Specifically, the inclusion of the signal corresponding to the septal walls in optimal 8×8 binning results, on average, in $\sim 40\%$ and 25% more signal per pixel compared to either inner binning technique for BGO-50 and LYSO-50, respectively. The higher relative signal exhibited by the septal walls of BGO-50 is partially responsible for the greater improvement in spatial resolution for that prototype following the application of inner or smart binning, as

discussed in section III.C.1. Finally, note that LYSO-50 exhibits a sensitivity ~ 3.7 times larger than that of BGO-50 by virtue of a higher light yield (see Table I).

NNPS for BGO-50, LYSO-50, and LYSO-115 are shown in Fig. 9 for various binning techniques at 1 beam pulse. For all prototypes, the magnitude of the NNPS is seen to decrease with increasing number of binned pixels per element (i.e., from inner 4×4 to smart to optimal 8×8 binning), with BGO-50 exhibiting the strongest dependence due to a lower light output. For optimal 8×8 binning, compared to LYSO-50 and LYSO-115, the NNPS for BGO-50 exhibits a less pronounced fall-off as a function of spatial frequency – consistent with the superior spatial resolution demonstrated by that prototype in Fig. 7.

III.C.3 DQE—DQE results for BGO-50, LYSO-50, and LYSO-115 at 1 and 4 beam pulses are presented in Fig. 10 for various binning techniques. For BGO-50 at 1 beam pulse, DQE increases with the number of binned pixels (i.e., from inner 4×4 to smart to optimal 8×8 binning) due to the corresponding increase in signal. Additional increases in signal achieved through use of more radiation (i.e., 4 beam pulses) further increases DQE to levels that are close to the upper limit predicted by radiation transport simulation. This dependence of DQE on signal indicates that performance is limited by electronic additive noise at low dose (i.e., at 1 beam pulse) due to the low light output of the BGO-50 scintillator.

For LYSO-50 and LYSO-115, the higher light output of these scintillators results in input-quantum limited behavior with DQE largely independent of the number of binned pixels and dose. Thus, for both scintillators, even though smart binning improves spatial resolution at the cost of lower signal, it does not degrade DQE compared to the use of optimal 8×8 binning. Conversely, the fact that smart binning does not appear to improve DQE beyond the level achieved with 8×8 binning is an indication that element-to-element misalignment present in these scintillators is not adversely affecting DQE for 8×8 binning. This is likely the result of the deterministic nature of the noise propagation induced by misalignment, which, in turn, results in modulation of the NNPS by the square of MTF (Cunningham, 2000), muting the impact of misalignment on DQE (see Eq. 2). Note that, for both scintillators, the measured results are often found to be higher than the expected upper limit for DQE as predicted by radiation transport simulations – an anomalous behavior that is not presently understood. Finally, as seen in Fig. 10, all prototypes provide DQE values considerably higher than those provided by a conventional MV AMFPI.

III.C.4 CBCT imaging—Qualitative examples of the comparative effects of the various binning techniques are given in Fig. 11, where reconstructed images of the resolution phantom obtained with LYSO-50 at a total dose of 16 cGy are shown. As seen in Fig. 11b, with the use of optimal 8×8 binning it is possible to discern the inserts representing a resolution of 4 lp cm^{-1} (corresponding to a line spacing of $\sim 1.25 \text{ mm}$). However, as seen in Fig. 11a, the use of an 8×8 binning shifted by a single pixel in both directions compared to optimal 8×8 binning degrades spatial resolution. The observed degradation is the result of the deleterious effect of mis-registration that may be exacerbated by the poor alignment of the prototype (see Fig. 2). By comparison, the use of smart binning, shown in Fig. 11c, noticeably improves spatial resolution, as demonstrated by sharper edges and a clearer definition of the inserts up to 5 lp cm^{-1} (corresponding to a line spacing of $\sim 1 \text{ mm}$). Similar

images of the resolution phantom obtained with BGO-50 did not provide adequate image quality to clearly demonstrate the effect of the various binning techniques – a consequence of the low light output of the scintillator.

IV. DISCUSSION

Thick, segmented scintillators offer the advantage of significantly increasing the DQE of MV AMFPs employed in radiotherapy imaging. However, the element-to-element misalignment observed in current prototypes seems likely to persist in larger area devices suitable for clinical applications. This will present a challenge for optimal registration with the underlying AMFP array, and result in degradation of spatial resolution. In this paper, it has been demonstrated how this problem can be overcome through use of an AMFP array with a considerably finer pitch than that of the scintillator, coupled with novel binning techniques (inner and smart binning) to prevent cross-binning between neighboring scintillator elements. In particular, this methodology was found to be effective for prototype scintillators employing BGO and LYSO, but was ineffective in the case of a CsI:TI prototype due to the overwhelming degree of light spread (i.e., crosstalk) across its scintillator elements.

The use of a finer pitch array and inner (or smart) binning introduces the potential for loss of DQE due to a reduction in the signal to noise ratio. This reduction originates from the loss of signal when pixels are discarded during binning, combined with an increase in the magnitude of electronic additive noise relative to the signal of these smaller pixels. In the case of the LYSO prototypes, the light output of the material was sufficiently high that the DQE performance was found to be largely independent of the binning technique and dose. For the BGO prototype, however, the lower light output of the material resulted in degradation of DQE with fewer binned pixels and at low dose.

For optimum performance, it is generally more advantageous to use smart binning since it improves spatial resolution while maximizing binned signal. The application of smart binning not only addresses the problem of misalignment, but also improves the intrinsic resolution of the scintillator by selectively excluding optical photons detected in the vicinity of septal walls. Such selective exclusion removes a portion of photons that originate from neighboring elements on each side of the septal walls – photons that would otherwise be included in 8×8 binning. The relative importance of these excluded photons seems to determine the resulting degree of improvement in spatial resolution. For the prototypes based on BGO and LYSO, a more intense signal is registered under the septal walls than under the crystals, indicating the existence of some form of optical trapping mechanism in the vicinity of the septal walls – a mechanism that traps a portion of optical photons that might otherwise contribute to cross-talk. For the prototype based on CsI:TI, this mechanism appears to be absent, as indicated by a lower signal under the septal walls than under the crystals, resulting in a lesser degree of improvement in spatial resolution with the application of smart binning. The efficacy of smart binning for improving spatial resolution is thus seen to depend on the amount of cross-talk which, in turn, depends on the optical properties of the scintillator material. For example, harder materials (such as BGO and LYSO) facilitate a higher polish (i.e., more mirror-like surfaces), which leads to reduced

cross-talk. In addition, materials (such as BGO) with a high refractive index mismatch between the scintillator crystals and the glue in the septal walls also reduce cross-talk. However, while BGO seems to satisfy many of the desired properties for segmented scintillators, its low light output was observed to preclude input-quantum-limited operation at low dose. For that reason, an alternative material such as CdWO₄, which has approximately a factor of two higher light yield than BGO, as well as a comparable refractive index, mechanical hardness and density, is potentially more suitable for effective application of smart binning, and thus better optimization of the performance of thick, segmented scintillators.

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REFERENCE

- Antonuk LE. Electronic portal imaging devices: a review and historical perspective of contemporary technologies and research. *Phys. Med. Biol.* 2002; 47:R31–R65. [PubMed: 11936185]
- Antonuk LE, Zhao Q, El-Mohri Y, Du H, Wang Y, Street RA, Ho J, Weisfield R, Yao W. An investigation of signal performance enhancements achieved through innovative pixel design across several generations of indirect detection, active matrix, flat-panel arrays. *Medical Physics*. 2009; 36:3322–39. [PubMed: 19673228]
- Barker JL, Garden AS, Ang KK, O'Daniel JC, Wang H, Court LE, Morrison WH, Rosenthal DI, Chao SC, Tucker SL, Mohan R, Dong L. Quantification of volumetric and geometric changes occurring during fractional radiotherapy for head-and-neck cancer using an integrated CT/linear accelerator system. *Int. J. Radiat. Oncol., Biol., Phys.* 2004; 59:960–70. [PubMed: 15234029]
- Cunningham, IA. Handbook of medical imaging. Beutel, J., et al., editors. SPIE; Bellingham, WA: 2000. p. 79-160.
- Dainty, JC.; Shaw, R. Image Science: Principles, Analysis and Evaluation of Photographic Type Imaging Processes. Academic; London: 1974.
- El-Mohri Y, Antonuk LE, Zhao Q, Choroszuca RB, Jiang H, Liu L. Low-dose megavoltage cone-beam CT imaging using thick, segmented scintillators. *Phys. Med. Biol.* 2011; 56:1509–27. [PubMed: 21325709]
- El-Mohri Y, Jee K-W, Antonuk LE, Maolinbay M, Zhao Q. Determination of the detective quantum efficiency of a prototype, megavoltage indirect detection, active matrix flat-panel imager. *Medical Physics*. 2001; 28:2538–50. [PubMed: 11797959]
- Fujita H, Tsai D, Itoh T, Doi K, Morishita J, Ueda K, Ohtsuka A. A simple method for determining the modulation transfer function in digital radiography. *IEEE transactions on Medical Imaging*. 1992; 11:34–9. [PubMed: 18218354]
- Giger ML, Doi K, Metz CE. Investigation of basic imaging properties in digital radiography. 2. Noise Wiener spectrum. *Medical Physics*. 1984; 11:797–805. [PubMed: 6513887]
- Granfors PR, Aufrichtig R, Possin GE, Giambattista BW, Huang ZS, Liu J, Ma B. Performance of a 41 × 41 cm² amorphous silicon flat panel x-ray detector designed for angiographic and R&F imaging applications. *Med Phys*. 2003; 30:2715–26. [PubMed: 14596310]
- Groh BA, Siewerdsen JH, Drake DG, Wong JW, Jaffray DA. A performance comparison of flat-panel imager-based MV and kV cone-beam CT. *Med. Phys.* 2002; 29:967–75. [PubMed: 12094992]
- Huang W, Antonuk LE, Berry J, Maolinbay M, Martelli C, Mody P, Nassif S, Yeakey M. An asynchronous, pipelined, electronic acquisition system for active matrix flat-panel imagers (AMFPIs). *Nucl. Instrum. Meth. Phys. Res. A*. 1999; 431:273–84.

- Jaffray DA, Siewerdsen JH. Cone-beam computed tomography with a flat-panel imager: initial performance characterization. *Med. Phys.* 2000; 27:1311–23. [PubMed: 10902561]
- Jaffray DA, Siewerdsen JH, Wong JW, Martinez AA. Flat-panel cone-beam computed tomography for image-guided radiation therapy. *Int. J. Radiat. Oncol., Biol., Phys.* 2002; 53:1337–49. [PubMed: 12128137]
- Kawrakow, I. Technical Report PIRS-899 National Research Council of Canada. Ottawa, Canada: 2005. egsp: the EGSnrc C++ class library.
- Kawrakow, I.; Rogers, DWO. Technical Report PIRS-701, National Research Council of Canada. Ottawa, Canada: 2000. The EGSnrc Code System: Monte Carlo Simulation of Electron and Photon Transport.
- Krus DJ, Novak WP, Perna L. Precision linear and two-dimensional scintillation crystal arrays for x-ray and gamma ray imaging applications. *Proc. SPIE.* 1999; 3768:183–94.
- Lewis DG, Swindell W, Morton EJ, Evans PM, Xiao ZR. A megavoltage CT scanner for radiotherapy verification. *Physics in Medicine & Biology.* 1992; 37:1985–99. [PubMed: 1438558]
- Liu L, Antonuk LE, Zhao Q, El-Mohri Y, Jiang H. Countering beam divergence effects with focused segmented scintillators for high DQE megavoltage active matrix imagers. *Phys. Med. Biol.* 2012; 57:5343–58. [PubMed: 22854009]
- Maidment AD, Yaffe MJ. Analysis of the spatial-frequency-dependent DQE of optically coupled digital mammography detectors. *Medical Physics.* 1994; 21:721–9. [PubMed: 7935207]
- Marshall NW, Mackenzie A, Honey IE. Quality control measurements for digital x-ray detectors. *Phys. Med. Biol.* 2011; 56:979–99. [PubMed: 21248386]
- Monajemi TT, Tu D, Fallone BG, Rathee S. A bench-top megavoltage fan-beam CT using CdWO₄-photodiode detectors. II. Image performance evaluation. *Med. Phys.* 2006; 33:1090–100. [PubMed: 16696486]
- Morin O, Gillis A, Chen J, Aubin M, Bucci MK, Roach M III, Pouliot J. Megavoltage cone-beam CT: system description and clinical applications. *Med. Dos.* 2006; 31:51–61.
- Mosleh-Shirazi MA, Swindell W, Evans PM. Optimization of the scintillation detector in a combined 3D megavoltage CT scanner and portal imager. *Med. Phys.* 1998; 25:1880–90. [PubMed: 9800695]
- Pouliot J, Bani-Hashemi A, Chen J, Svatos M, Ghelmansarai F, Mitschke M, Aubin M, Xia P, Morin O, Bucci K, Roach M 3rd, Hernandez P, Zheng Z, Hristov D, Verhey L. Low-dose megavoltage cone-beam CT for radiation therapy. *Int. J. Radiat. Oncol., Biol., Phys.* 2005; 61:552–60. [PubMed: 15736320]
- Rathee S, Tu D, Monajemi TT, Rickey DW, Fallone BG. A bench-top megavoltage fan-beam CT using CdWO₄-photodiode detectors. I. System description and detector characterization. *Med. Phys.* 2006; 33:1078–89. [PubMed: 16696485]
- Sawant A, Antonuk LE, El-Mohri Y. Slit design for efficient and accurate MTF measurement at megavoltage x-ray energies. *Medical Physics.* 2007; 34:1535–45. [PubMed: 17555235]
- Sawant A, Antonuk LE, El-Mohri Y, Zhao Q, Li Y, Su Z, Wang Y, Yamamoto J, Du H, Cunningham I, Klugerman M, Shah K. Segmented crystalline scintillators: an initial investigation of high quantum efficiency detectors for megavoltage x-ray imaging. *Medical Physics.* 2005; 32:3067–83. [PubMed: 16279059]
- Sawant A, Antonuk LE, El-Mohri Y, Zhao Q, Wang Y, Li Y, Du H, Perna L. Segmented crystalline scintillators: Empirical and theoretical investigation of a high quantum efficiency EPID based on an initial engineering prototype CsI(Tl) detector. *Medical Physics.* 2006; 33:1053–66. [PubMed: 16696482]
- Seppi EJ, Munro P, Johnsen SW, Shapiro EG, Tognina C, Jones D, Pavkovich JM, Webb C, Molloy I, Partain LD, Colbeth RE. Megavoltage cone-beam computed tomography using a high-efficiency image receptor. *Int. J. Radiat. Oncol., Biol., Phys.* 2003; 55:793–803. [PubMed: 12573767]
- Sheikh-Bagheri, D. Department of Physics. Carleton University; 1999. Monte Carlo Study of Photon Beams from Medical Linear Accelerators: Optimization, Benchmark and Spectra.
- Siewerdsen JH, Antonuk LE, El-Mohri Y, Yorkston J, Huang W, Cunningham IA. Signal, noise power spectrum, and detective quantum efficiency of indirect-detection flat-panel imagers for diagnostic radiology. *Medical Physics.* 1998; 25:614–28. [PubMed: 9608470]

- Sillanpaa J, Chang J, Mageras G, Yorke E, De Arruda F, Rosenzweig K, Munro P, Seppi E, Pavkovich J, Amols H. Low-dose megavoltage cone-beam computed tomography for lung tumors using a high-efficiency image receptor. *Med. Phys.* 2006; 33:3489–97. [PubMed: 17022245]
- Wang Y, Antonuk LE, El-Mohri Y, Zhao Q, Sawant A, Du H. Monte Carlo investigations of megavoltage cone-beam CT using thick, segmented scintillating detectors for soft tissue visualization. *Medical Physics.* 2008; 35:145–58. [PubMed: 18293571]
- Wang Y, Antonuk LE, Zhao Q, El-Mohri Y, Perna L. High-DQE EPIDs based on thick, segmented BGO and CsI:Tl scintillators: Performance evaluation at extremely low dose. *Medical Physics.* 2009; 36:5707–18. [PubMed: 20095283]
- Wang Y, El-Mohri Y, Antonuk LE, Zhao Q. Monte Carlo investigations of the effect of beam divergence on thick, segmented crystalline scintillators for radiotherapy imaging. *Phys. Med. Biol.* 2010; 55:3659–73. [PubMed: 20526032]
- Yin FF, Guan H, Lu W. A technique for on-board CT reconstruction using both kilovoltage and megavoltage beam projections for 3D treatment verification. *Medical Physics.* 2005; 32:2819–26. [PubMed: 16266096]

Figure 1. Photograph of the cone-beam CT bench-top system used in the study, illustrating a close-up of the spatial resolution phantom mounted above one of the segmented scintillator prototypes and the underlying AMFPI array.

Figure 2.

(a) X-ray flood image obtained with an MV AMFPI incorporating the LYSO-50 prototype scintillator using a 6 MV photon beam at a dose corresponding to one beam pulse. The scintillator was coupled to an AMFPI array having a pixel pitch of 0.127 mm. Note that the septal walls, which appear as white grid lines, correspond to a higher signal registered by some of the array pixels. (b) and (c) are enlargements of different parts of this image, and the superimposed black dashed lines correspond to a common, regular grid having a period of 8 pixels, with a pitch of 1.016 mm.

Figure 3.

Three-dimensional plots showing values of the alignment metric for (a) LYSO-115 and (b) LYSO-50. These values correspond to all possible overlaps between the mask of septal wall segments obtained from a flood image of the prototype and a mask of segments corresponding to a perfectly regular, 1.016 mm pitch grid.

Figure 4.

Binary masks illustrating the degree of element-to-element alignment for prototypes (a) LYSO-115 and (b) LYSO-50 corresponding to the maximum value of the alignment metric for each detector, ~86% and 63%, respectively. The images used to generate these masks were derived from x-ray flood images such as that shown in Fig. 2a. Note that the masks represent those wall segments of elements in the scintillators whose positions coincide with those in a perfect grid. Thus, missing segments correspond to misalignment of the septal walls of the scintillator with respect to that perfect grid.

Figure 5.

MTF obtained from the simulated geometry of a segmented scintillator with a 1.016 mm element pitch and 0.05 mm septal walls coupled to an AMFPI array with a 0.127 mm pixel pitch and 80% optical fill factor. The various MTF curves correspond to different pixel offsets of the array with respect to the scintillator prior to application of 8×8 binning to match the size of the scintillator elements. For comparison, a sinc function based on an aperture size corresponding to the active area of the scintillator elements is also included.

Figure 6.

Reconstructed images of a simulated resolution phantom. The images correspond to configurations: (a) 0 pixel offset, (b) 1 pixel offset, (c) 2 pixel offset, and (d) 4 pixel offset – representing increasing offset of the array with respect to the scintillator in both the horizontal and vertical directions. For each image, the line-pair inserts correspond to line spacing distances of (clockwise from the top right) ~4.9, 3.2, 2.3, 1.8, 1.5, 1.3, 1.1 and 1.0 mm. Note that the insets above the reconstructed images in (a) and (d), which are used to enhance presentation, are enlargements of the line-pair insert corresponding to a line spacing distance of ~1.1 mm. (The judicious placement of the insets necessitated anticlockwise arrangement of the images.) Also note that the wavy faint lines appearing across the images are artifacts associated with the reconstruction process.

Figure 7.

Measured MTF results obtained at 6 MV using the (a) BGO-50, (b) LYSO-50, (c) LYSO-115, and (d) CsI-50 prototype segmented scintillators. Results are shown for optimal 8×8 binning, inner 4×4 binning, and smart binning. The results are compared to those obtained from previously published data of a conventional MV AMFPI (El-Mohri *et al* 2001). (This previously published data was binned in a 2×2 format to match the pitch of the prototypes.) The results are also compared to upper limits obtained from radiation transport simulations. Note that, for the BGO-50 prototype, the MTFs for inner 4×4 and smart binning closely overlap.

Figure 8.

Measured signal obtained at 6 MV for the (a) BGO-50 and (b) LYSO-50 prototypes. Results are shown for optimal 8×8 binning, inner 6×6 binning and inner 3×3 binning. The lines running through the data points are linear fits, the slopes of which are reported in units of 10^6 electrons per beam pulse.

Figure 9. Measured NNPS results obtained at 6 MV and 1 beam pulse for the (a) BGO-50, (b) LYSO-50, and (c) LYSO-115 prototypes. Results are shown for optimal 8x8 binning, inner 4x4 binning and smart binning.

Figure 10.

Measured DQE results obtained at 6 MV and 1 beam pulse for the (a) BGO-50, (b) LYSO-50, and (c) LYSO-115 prototypes. Results are shown for optimal 8x8 binning, inner 4x4 binning, and smart binning. For purposes of comparison, results obtained using 4 beam pulses for smart binning, as well as from simulations of radiation transport and from a conventional MV AMFPI (El-Mohri et al 2001), are also shown. (The previously published data from the conventional MV AMFPI was binned in a 2x2 format to match the pitch of the prototypes.)

Figure 11.

Reconstructed images of the resolution phantom obtained with the LYSO-50 prototype. The images were obtained using (a) 8×8 binning with a single pixel offset in the horizontal and vertical directions compared to optimal 8×8 binning, (b) optimal 8×8 binning and (c) smart binning. Note that the gray scale for these images has been inverted in order to enhance presentation. The line-pair inserts present in the images represent spatial resolutions of (clockwise from the top) 1, 2, 3, 4, 5, 6, 7, 8, 9, 10 and 11 lp/cm, corresponding to line spacing distances of ~5.00, 2.50, 1.66, 1.25, 1.00, 0.83, 0.71, 0.62, 0.55, 0.50 and 0.45 mm, respectively.

Table I

Summary of the design specifications and properties of the 1.016 mm pitch segmented scintillator prototypes employed in this study. Specifications reported in the last three columns were obtained from Saint-Gobain Crystals.

Prototype Designation	Septal Wall Thickness (mm)	Thickness (mm)	Density (g/cm ³)	Refractive Index	Light Yield (Photons/keV)
BGO-50	0.05	11.3	7.13	2.15	9
CsI-50	0.05	11.4	4.51	1.80	54
LYSO-50	0.05	10.3	7.10	1.81	32
LYSO-115	0.115	10.2	7.10	1.81	32

Table II

Maximum values of the alignment metric for the segmented scintillator prototypes examined in this study.

Prototype Designation	Alignment metric (%)
BGO-50	74
CsI-50	71
LYSO-50	63
LYSO-115	86

Table III

Summary of pixel binning techniques employed in this study. The second and third columns contain, for each technique, the specific binning combinations employed and a brief description, respectively.

Binning Technique	Pixel binning combinations	Description
Optimal 8×8	8×8	Corresponds to that specific grouping of pixels which maximizes the alignment metric.
Inner n×n	3×3 4×4 6×6	Selective binning of a fixed number of central contiguous pixels within each element.
Smart	Up to 7×7	Selective binning of a variable number of central contiguous pixels within each element.