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# A new approach in the design of electronic portal imaging devices for portal dosimetry in radiotherapy

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ABSTRACT

A CCD-based EPID using new crystal-assembly X-ray (CAX) converters is investigated for radiotherapy dosimetry. The proposed EPID design consists in replacing the common phosphor X-ray converters of current CCD-based EPIDs with high-stopping-power CAX converters. A Test Imaging Device (TID), consisting of a 30-mm-thick CAX converter made of Bismuth Germanate (BGO), coupled to a highly sensitive CCD camera, was used to evaluate the accessible imaging and dosimetric performance of the proposed design. The system response to dose and its dependence on photon beam energy were investigated. The effects of ghosting, dose rate, field size and phantom thickness were evaluated as well. The same measurements were also performed with our clinically used aSi-EPID so that comparisons of performance could be directly inferred. The TID displayed no detectable ghosting or sensitivity to dose rate. Its response to MU exposure was found to be linear within about ±1%. The level of glare induced in the TID and the aSi-EPID were equivalent. The TID resolution was higher than that of the aSi-EPID on the axis, but was found to decrease with off-axis distance. Finally, the image quality, assessed on the basis of signal-to-noise ratio in low dose radiographs of the larynx of a patient, was higher for the TID. The imaging performance accessible with the TID proved to be satisfying and its dosimetric capability was found to be superior to that of the current aSi-EPID.



## Introduction

The electronic portal imaging devices (EPIDs) were initially introduced to replace radiographic films for verification of patient positioning [1,2]. Their use for radiotherapy dose measurement has increasingly been investigated and developed ever since [3].

The detective quantum efficiency (DQE) [4] is a widely used measure of the performance of imaging devices [1,5,6]. DQE is expressed as the square of the ratio of the output signal-to-noise ratio (SNR) to the input SNR as a function of spatial frequency $f$ (see Eq. (1)). The higher the DQE, the better is transferred the information contained in the incident X-ray image. The X-ray quantum efficiency (QE) represents the fraction of incident X-ray photons that actually interact with the imaging device. Since $DQE \le QE \le 1$ [4], the imaging device should always have high QE. However, the QEs of most portal imaging devices commercially available do no exceed a few percent, and their DQEs reaches only about 1% in case of the best performing systems [1,6]

$$\mathrm{DQE}(f) = \left[\frac{\mathrm{RSB}_{\mathrm{out}}(f)}{\mathrm{RSB}_{\mathrm{in}}(f)}\right]^2 \tag{1}$$

The amorphous Silicon electronic portal imaging devices (aSi-EPIDs) are the most common EPIDs commercially available today. They display advantageous characteristics for use as portal dosimeters [3]. However, even if amorphous silicon p-i-n diodes are very resistant to radiation [7], the imaging properties of aSi-EPIDs alter with time due to the radiation damage to the peripheral readout electronics [6]. Quality control measurements should be performed on a regular basis and if changes in the dosimetric properties are detected, the detector must be recalibrated to ensure the needed accuracy in the verification process [8]. The recalibration procedure consumes valuable staff and machine time. In addition, the intrinsic ghosting of aSi-EPIDs [9] directly impacts the duration of the procedure because pauses (a few minutes typically) must be made between EPID irradiations to ensure that the ghosting effects do not affect the measurements.

CCD-based EPIDs are also highly suitable for portal dosimetry [10,11] and display excellent long-term stability of the electronics, especially when cooled CCDs are used [11]. However, the optical

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scatter (glare) inside the optical system induces large field size dependence. This effect needs to be removed from the images, for instance by using cross-talk deconvolution techniques [10].

In the commercially available CCD-based or aSi-EPIDs, the X-ray-to-light converter is a phosphor (usually Gadolinium Oxysuphide) bonded to a metal plate (usually a 1-mm-thick Copper sheet). The QE of such converters, at therapy energies, is only about 2–4% [2]. In addition, it has been shown that increasing the phosphor thickness beyond about 400 mg/cm$^2$ brings little gain in light output while spatial resolution keeps declining due to the spreading of the light photons and the dose deposition inside the phosphor [12]. In case of the CCD-based EPIDs, however, one solution consists in replacing the metal/phosphor converter with a thick, transparent, monolithic scintillator to considerably increase the QE of the system without drastically reducing the system resolution: such imaging systems were evaluated [13] and proved to produce high quality portal imaging (although, to our knowledge, their evaluation for dose measurement has not been reported yet). When coupling thick monolithic scintillators (up to a few centimeters) to an optical system through reasonably small angular aperture – typically less than 10° – the voxel (elementary volume in the scintillator viewed by a CCD pixel) remains small enough to ensure good resolution [14,15]. Since standard commercial objectives are generally less than 80 mm in diameter, when the objective is located further than 450 mm from the scintillator, the induced angular aperture at the object plane is indeed less than 10° in angle. Unfortunately, industrial manufacturing techniques of the best suited crystals for multi-MeV X-ray imaging (i.e. those having both high density and good light yield, and which can be manufactured with very few or no visible defects in the bulk) do not currently allow production of plates as large as 300–400 mm. For instance, in case of Bismuth Germanate $Bi_4Ge_3O_{12}$ (BGO), manufacturers are only able to produce blocks of high quality crystal with uniform light yield and with no major defects (such as microscopic cracks or small bubbles) up to approximately 80 × 250 mm$^2$ with a maximum thickness of about 35 mm. In case of Cesium-doped Lutetium Yttrium Orthosilicate $Lu_{2(1-x)}Y_{2x}SiO_5$:Ce (LYSO), the maximum size is even smaller (approximately 70 × 100 mm$^2$ at the present time, in thicknesses lower than 30 mm). Cesium Iodide (CsI) and Sodium Iodine (NaI) crystals can be manufactured in larger sizes (typically up to 200 × 200 mm$^2$ and 400 × 400 mm$^2$ respectively), but these crystals have lower density and are rather hygroscopic. In addition, according to our supplier, the number of defects present in these crystals is noticeably higher (particularly with CsI) which makes them poorer candidates than BGO or LYSO for imaging applications such as those evaluated in the present study. For instance, in Sawant et al. [13] the author reports about 20–30 flaws, 0.1–4 mm in width, randomly spread over the volume of their 12-mm thick, 230 × 230 mm$^2$ in dimension, CsI(Tl) crystal. These defects scatter the light produced within the scintillator and thus produce visible specks in the final image. An alternative solution is to use 2D crystal arrays (known as segmented scintillators) instead of monolithic crystals [16] but, their cost is higher and the image quality is not as good as with monolithic crystals [14].

Finally, to overcome the size limitation in the manufacturing of monolithic crystal scintillators, a specific technique of assembling crystal plates has been developed so that the joints between plates have minimum effect on the resulting images (such X-ray converters are now commercially available [15]). These new crystal-assembly X-ray (CAX) converters present small apparent joint thicknesses at the plate junctions (typically about 200 μm) and are optimized so that little light gets diffused at the joints and at the edges. In particular, the surface located on the X-ray source side and all edge surfaces are covered with a black (absorbing) paint or glue layer in order to minimize optical scatter. CCD-based radiographic imagers using CAX converters have been used for several years in shock physics experiments [14].

In order to derive the primary conclusions on the potential imaging and dosimetric capabilities of CCD-based EPIDs if equipped with a CAX converter, we built and investigated the performance of a Test Imaging Device (called TID in the present study), consisting of a 30-mm-thick BGO CAX converter coupled to a highly sensitive, low-noise, thermoelectrically cooled CCD camera. The system response to dose and its dependence on photon beam energy were investigated. The effects of ghosting, dose rate, field size and phantom thickness were evaluated as well. The image quality was assessed on the basis of signal-to-noise ratios in the visualization of a specific anatomic feature in low dose (2 MU exposures) patient radiographs. The same measurements were also performed with our clinically used aSi-EPID so that comparisons of performance could be directly inferred. In our assessment, specific attention was given to the ability of the proposed EPID design to lead to reduced work-load and frequency of periodic calibrations for radiotherapy.

## Materials and methods

### *Linear accelerator and equipment*

All measurements were performed on an Elekta linear accelerator, model Precise (Elekta, Stockholm, Sweden). This accelerator can produce 6, 10 and 18-MV photon beams. The aSi-EPID was an Elekta *i*ViewGT set at a fixed source-detector distance of 160 cm its imaging size is 41 × 41 cm$^2$. It is made of 1024 × 1024, 400 μm square pixels. The *i*ViewGT EPID contains a 1-mm-thick copper plate to provide build-up and absorb scattered radiation and a 0.54-mm-thick phosphor screen made of terbium-doped gadolinium oxysulphide ($Gd_2O_2S$:Tb) to convert the incident radiation into optical photons. The dosimetric properties of *i*ViewGT EPIDs have already been extensively investigated by several teams, e.g. Ref. [17]. The measurements were performed about 32 months after installation of the EPID. At the time of the measurements, the EPID had mainly been used to produce localization images, i.e. with typically 2–3 cGy exposures, prior to delivery of the main treatment doses (during which the EPID is always retracted). A calibrated 0.3 cm$^3$ thimble-type PTW ionization chamber (model TW31013, PTW, Freiburg, Germany) and a Wellhofer electrometer (model DOSE 1, Wellhofer, Schwarzenbruck, Germany) were used for the dose measurements. Water dose measurements were performed using adequate build-up solid water slab thickness (5 cm at 6 MV photon beam energy, and 10 cm at 10 and 18 MV), as well as 10 cm of solid water slab at the back of the ionization chamber to produce backscatter.

### *Test imaging device (TID)*

A 3D view of the TID is provided in Fig. 1. It includes a highly sensitive, low-noise, thermoelectrically cooled, 16-bit CCD camera equipped with a photographic lens and a mirror used to place the CCD camera out of the direct X-ray beam.

The scintillator, used to convert the X-ray image into visible light, was a 30-mm-thick, 165 × 165 mm$^2$ dimension, CAX converter made of two BGO crystal plates (see photograph in Fig. 2) and manufactured by Saint-Gobain Crystals (Saint-Gobain Cristaux, Saint-Pierre-les-Nemours, France). A 6-mm-thick Aluminum cover protects the scintillator. There is a 23 mm air gap between the scintillator and the Aluminum cover. The cover also provides filtering of low-energy scattered X-rays (and some build-up while, ideally, it should be in contact with the scintillator). The CCD camera was manufactured by Princeton Instruments (Trenton, New Jersey, USA), model PIXIS 2048B. The CCD sensor, 27.6 × 27.6 mm$^2$