Dosimetry in radiotherapy using a-Si EPIDs: Systems, methods, and applications focusing on 3D patient dose estimation

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Abstract. An overview is provided of the use of amorphous silicon electronic portal imaging devices (EPIDs) for dosimetric purposes in radiation therapy, focusing on 3D patient dose estimation. EPIDs were originally developed to provide on-treatment radiological imaging to assist with patient setup, but there has also been a natural interest in using them as dosimeters since they use the megavoltage therapy beam to form images. The current generation of clinically available EPID technology, amorphous-silicon (a-Si) flat panel imagers, possess many characteristics that make them much better suited to dosimetric applications than earlier EPID technologies. Features such as linearity with dose/dose rate, high spatial resolution, real-time capability, minimal optical glare, and digital operation combine with the convenience of a compact, retractable detector system directly mounted on the linear accelerator to provide a system that is well-suited to dosimetric applications. This review will discuss clinically available a-Si EPID systems, highlighting dosimetric characteristics and remaining limitations. Methods for using EPIDs in dosimetry applications will be discussed. Dosimetric applications using a-Si EPIDs to estimate three-dimensional dose in the patient during treatment will be overviewed. Clinics throughout the world are implementing increasingly complex treatments such as dynamic intensity modulated radiation therapy and volumetric modulated arc therapy, as well as specialized treatment techniques using large doses per fraction and short treatment courses (ie. hypofractionation and stereotactic radiosurgery). These factors drive the continued strong interest in using EPIDs as dosimeters for patient treatment verification.

1. Introduction
Radiation treatment delivery is becoming more complex, emphasized with the introduction of techniques such as dynamic intensity modulated radiation therapy (IMRT) and rotational IMRT. Rotational IMRT, commercially available as either RapidArc™ (Varian Medical Systems, Palo Alto, CA) or VMAT™ (Elekta AB, Stockholm, Sweden), is especially complex, delivering patient treatment while the gantry rotates, and involves simultaneous modulation of radiation aperture, dose rate, gantry speed, and potentially collimator and couch speeds. In addition to these new delivery technologies, there is increased clinical interest in aggressive treatment regimens that deliver the therapeutic dose in larger per-fraction doses and in fewer fractions compared to standard treatments. These techniques, including hypofractionation and stereotactic radiosurgery, place greater importance on the dosimetric
accuracy of each individual treatment fraction delivered. In response to these developments, quality assurance bodies throughout the world acknowledge that patient treatment verification, preferably through in vivo patient dosimetry, is highly desirable for optimal patient safety during radiation treatment [1-4]. Electronic portal imaging devices (EPIDs) provide a means for this verification.

EPIDs were originally developed to improve the accuracy of patient setup by generating radiographs of the patient during treatment delivery. The image is formed by x-rays created by the megavoltage photon source in the linear accelerator (i.e. linac). Previously, several different technologies have been utilized to capture these images, including the scanning liquid ionization chamber array (commercialized as PortalVision by Varian Medical Systems), and camera-based systems viewing a phosphor scintillating screen. Camera-based systems used either analogue video cameras (Beamview plus by Siemens Healthcare) or CCD cameras (SRI-100 by Elekta AB). By today’s standards, these first generation EPID designs produced somewhat poor quality radiological images and possessed some useful but very limited dosimetric characteristics. Beginning in the early 2000s and continuing throughout the decade, the current generation of amorphous-silicon (a-Si) EPID technology eventually became commercially available from three major linac manufacturers. Current a-Si EPID technology not only generates better quality images, but also possesses more useful dosimetric characteristics that will be discussed later in this review.

Most dosimetric applications have been investigated using the imagers in an ‘integrating’ acquisition mode where the image signal is integrated over an entire irradiation (i.e. during a treatment beam delivery) to produce a single image. However, recent studies have examined the dosimetric characteristics of the imagers operated in a ‘movie’ acquisition mode, where the image signal is read out at equal time intervals throughout an irradiation to produce a series of images. This is largely motivated by the time-dependent nature of rotational IMRT. Ultimately, the widespread availability of a-Si EPIDs have made them the subject of strong research activity into a broad range of dose applications including patient treatment verification.

There are a variety of ways to employ EPIDs as patient dosimeters, ranging from pre-treatment verification to full 3D dose estimation of the patient treatment from images gathered during treatment delivery. These methods have been described in detail by van Elmpt [5] and will only be briefly outlined in this review. Application examples for 3D patient dose estimation will be focused on here, since the rewards of daily 3D patient dose estimation are great: (1) mapping under- or over-estimates of patient dose onto the anatomy, thus providing clinically relevant feedback, (2) availability of cumulative dose over the entire treatment course, (3) improved treatment through adaptive radiotherapy approaches, and (4) production of a medical/legal record of delivered patient dose.

2. a-Si EPID systems and dosimetric characteristics

This section will review how a-Si EPID systems form images and also discuss system characteristics with respect to measuring dose.

2.1 a-Si EPID systems

The modern a-Si EPID is based on thin film semiconductor technology. The current generation of commercially available EPIDs use amorphous silicon (a-Si) deposited on a glass substrate to form arrays of photodiodes and field-effect transistors (FETs). An array of pixels is formed with each pixel consisting of a photodiode and an FET. Pixel pitches are currently 0.39×0.39 mm² (Varian) and 0.40×0.40 mm² (Elekta and Siemens), but with varying active areas of 40×30 cm² (Varian) and 41×41 cm² (Elekta and Siemens), corresponding to 1024×768 pixels and 1024×1024 pixels, respectively. The pixel arrays detect radiation indirectly, through the use of a scintillating screen (the standard phosphor used is gadolinium oxysulfide doped with terbium, Gd₂O₂S:Tb), which converts deposited radiation energy into optical photons that are then detected by the photodiodes. To improve detector efficiency, an additional sheet of metal is placed directly upstream of the scintillator, to help convert incident photons to electrons that then deposit dose in the scintillator. Typical metal layers in commercial EPIDs are 1 mm of copper (Varian and Elekta), although some users may place additional
buildup on the detector especially if performing dosimetry with high-energy therapy beams above 6 MV. Antonuk et al. helped pioneer the development of the modern a-Si EPID systems for radiotherapy, and early publications by that group contain detailed discussions of the design and operation of the detector components for the interested reader [6-8].

2.2. Dosimetric characteristics of a-Si EPIDs
Desirable characteristics of any dosimeter include linearity (with dose and dose rate), reproducibility, high spatial resolution, no dead time, and real-time readout. Researchers have demonstrated many of these characteristics for a-Si EPIDs. For the PortalVision aS500/1000 (Varian) operated in integrating mode, Greer and Popescu’s 2003 investigation [9] contains many useful results. A small dead time effect identified in that work was subsequently removed by the manufacturer through a software upgrade in the acquisition computer (from IAS2 to IAS3). Dosimetric properties of aS500/1000 EPIDs operated in movie mode were investigated by McCurdy and Greer [10]. For the iViewGT EPID system (Elekta), McDermott et al [11] and Winkler et al [12] document many dosimetric performance studies of the system operated in integrating mode. Long term reproducibility of the Varian system has been demonstrated to be <1% (all pixels) over a three year period [13] and of the Elekta system was demonstrated to be <0.5% (all pixels) over nearly two years [14]. The small pixel pitches of these systems (~0.4 × 0.4 mm²) provide a high spatial resolution, vastly superior to most ionization chambers and also superior to computed tomography data sets where voxel sizes are typically about 1×1×2 mm³.

Several investigators have observed image lag and image ghosting effects with a-Si EPIDs [11, 12, 15, 16]. Image lag is due to trapped charge in the photodiode which, when read out in subsequent frames, results in the EPID signal being offset. Image ghosting refers to the change in individual pixel gains due to the trapped charge modifying the electric field strength in the photodiode. These effects have been shown to result in a relative under response of 4-11%, and this may have a direct impact on dosimetric applications of EPIDs [17]. However, these issues are primarily limited to short irradiation times (i.e. low number of monitor units) typically below those of routine clinical use, and also can be corrected for if desired [11, 12].

2.3. Challenges to using EPIDs for patient dosimetry
As useful for dosimetry as EPIDs have proven to be, there are still several challenges to applying them in clinical practice. First is the image ghosting/lag issue as described above. Second, the a-Si EPID is known to over respond to low energy photons (below about 0.5 MeV) as compared to a water equivalent detector [18, 19] due to the increased photoelectric effect in the copper/phosphor screen. Third, the a-Si EPID imaging units are fairly thin, cassette-like systems weighing a few kilograms, but this is enough material to cause a ‘self-scatter’ signal. That is, photon scatter generated within the EPID itself (not just primary photons from the linear accelerator head and/or scattered photons originating in the patient) will contribute to image signal [20]. Fourth, the optical photons scattering within the scintillator layer contribute to an optical glare effect (i.e. a blurring of the deposited dose image pattern). This effect is much smaller in a-Si EPIDs as compared to the previous generation of camera-based systems, and has been characterized [21]. Fifth, photon scatter will be generated by the treatment beam interacting with the patient, creating an additional, complex source of mostly lower energy photons incident on the EPID [19]. Sixth, for the Varian a-Si EPID, the imaging unit is mounted on a robotic arm that has been shown to contribute additional signal to the image from increased backscattered photons. The backscatter signal contribution is known to be asymmetrical, field size dependent, and field location dependent [22]. Seventh, for clinical use, the EPID systems are extended outwards from the main linear accelerator unit, into the treatment beam. When extended, the mounting systems are subject to gravitational forces which may cause mechanical flexion. This introduces small shifts in the EPID image location (intended position versus real position) as a function of gantry angle, of upwards of 1 mm (Varian) [23] and 4 mm (Elekta) [24]. Eighth, for some EPID dosimetry applications in rotational IMRT delivery, the gantry angle needs to be accurately known. It has been demonstrated [25] that gantry angle uncertainty associated with the acquired EPID...
images on Varian a-Si EPIDs typically exceeds the AAPM TG142 tolerance of 1° [26]. Ninth, for movie acquisition mode in Varian a-Si EPIDs, it has been shown that a small amount of image dose is missing from the total acquisition [10]. More recent work has shown this to be the result of incomplete image frame acquisition at the beginning of an irradiation [25].

The impact of these issues often depends on the specific EPID application. However, when using a-Si EPIDs for dosimetry, a variety of approaches have been shown to mitigate these challenges. Often empirical, correction-based techniques have been employed, which are attractive due to their simplicity. Others have shown that model-based approaches can be successful, trading off an increase in complexity to achieve robustness over a wide range of operating conditions.

3. Methods for EPID dosimetry
The extensive review of EPID dosimetry carried out in 2008 by van Elmpt et al. [5] provides a detailed cataloguing of the wide variety of methods found in the literature. The brief description that follows here is based on that work.

EPID dosimetry can be categorised as either ‘pre-treatment verification’, where the EPID measures treatment beams without the patient present, or ‘treatment verification’, where the EPID measures treatment beams during patient treatment. Further sub-categorisation can be applied depending on whether or not the treatment beams have passed through an attenuating medium before being measured by the EPID. If the treatment beams have not passed through an attenuating medium to reach the EPID, the method is referred to as ‘non-transmission dosimetry’ (or ‘non-transit dosimetry’). Otherwise, the method is described as ‘transmission dosimetry’ (or ‘transit dosimetry’).

In general, dosimetry applications for EPIDs compare the dose estimated via an EPID measurement with an expected independently calculated dose. Therefore, a third layer of sub-categorisation can be employed, based on where the dose comparison is being made. There are two locations where the comparison can be made. The dose comparison may be made at the detector level (i.e. comparing a measured image to an expected image), typically in 0D (point comparison) or 2D (image comparison). Alternatively, the dose comparison can be made at the patient/phantom (i.e. inferring a dose in the patient/phantom from the measured EPID image and comparing to an expected dose in the patient/phantom). This comparison is usually made in 0D (point comparison), 2D (plane comparison), or 3D (volumetric comparison).

The special case where patient dose (0D, 2D, or 3D) is estimated from EPID measurements during patient treatment (verbosely described by the above nomenclature as an ‘in-patient, transmission dosimetry, treatment verification method’) is simply described as ‘in vivo patient dosimetry’.

Much work has been done investigating a-Si EPID dosimetric applications in all of these categories and the reader is referred to the van Elmpt et al. [5] review paper for a detailed summary of the EPID dosimetry literature up to 2008. The remainder of the discussion in this work will focus on 3D in vivo patient dosimetry applications of a-Si EPIDs.

4. Techniques for 3D in vivo patient dosimetry using EPIDs
Arguably the most valuable a-Si EPID dosimetric application is 3D in vivo patient dose reconstruction, where the delivered patient dose is estimated in 3D from measured EPID transmission images of the treatment beam(s). Even before the current generation of a-Si EPID technology, there was strong interest in using EPIDs for in vivo patient dose verification. Many groups investigated in vivo EPID dosimetry methods using earlier EPID technologies such as the scanning liquid ionization chamber design [27-29] or the camera-based designs [30-33]. A few of these techniques were modified and tailored for use with a-Si EPIDs. In particular the methods of Boellard et al. [27] and Louwe et al. [28] were adapted by Wendling et al. [34, 35], while the methods of Hansen et al. [32] and Spies et al. [33] were customized by Partridge et al. 2002 [36]. Other approaches estimating 3D patient dose include Jarry et al. [37] and Van Uytven et al. [38]. Although each of these efforts are technically distinct, they share the same general framework: (1) patient scatter is estimated and subtracted from the measured EPID image, resulting in an estimate of primary fluence entering the EPID, (2) the estimate
of primary fluence is backprojected to a plane above the patient, and (3) the estimate of incident
primary fluence is used as input to a patient dose calculation algorithm. The main techniques in the
literature for utilizing a-Si EPIDs to recover 3D dose in the patient are summarized below.

The approach described by Partridge et al. [36] used a calibration technique to convert an EPID
image into a map of patient thicknesses. A series of pre-generated, Monte Carlo kernels (as a function
of thickness) were used in an iterative fashion to estimate the scattered radiation generated in the
patient. Once the scattered radiation component was subtracted from the image, the remaining
estimate of primary fluence was backprojected to a plane above the patient accounting for attenuation
and the inverse square effect. This primary fluence estimate was then passed as input to a
convolution/superposition style dose algorithm to calculate patient dose in 3D. One phantom was
tested using a 5-field IMRT plan, demonstrating a 5.5% agreement in absolute dose.

Jarry et al. [37] estimated the dose signal in the EPID due to patient scatter by employing a Monte
Carlo simulation. This scatter signal was subtracted from the measured EPID image to estimate the
EPID dose due to the primary beam, which was assumed to be proportional to primary fluence in the
EPID. The simulation also tracked photon energy spectra in small rectangular regions distributed
throughout the image. Primary photons that were scored in the simulation were then back-projected
through the patient CT data set towards an ideal point source (ie. an infinitesimal focal spot). This
created a phase-space file above the patient that was then used as a radiation source to simulate the
particle transport (in a forward fashion) through the CT data set. For several phantom test situations,
including 3DCRT and IMRT fields, an absolute dose agreement of 5% was found.

Another approach has been recently proposed by van Uytven et al. [38]. This method used a
detailed physical model to predict EPID images with the patient present. The model-based algorithm
predicting a-Si EPID images was developed by Chytyk et al. [39-41], based on earlier work by that
group [42-44]. The approach uses a detailed head fluence model of the linac, accounting for focal and
extra-focal sources, multi-leaf collimator, and energy spectra. It also explicitly handles patient scatter
(including energy spectra), energy response and self-scatter of the EPID, optical glare, and asymmetric
backscatter from the support arm. The model is used as part of an iterative algorithm that removes the
patient scatter from the EPID image to estimate primary fluence entering the EPID, including energy
spectra details. The iterative algorithm is similar to that of McNutt et al. [27], modified to take
advantage of the information available in the extra-focal portion of the head fluence model. Once
estimated, the primary fluence is back-projected above the patient, accounting for patient attenuation
and the inverse square law, and then utilized by an in-house developed collapsed-cone convolution
style patient dose calculation algorithm. Validation was performed on slab and anthropomorphic
phantoms for simple square fields, dynamic IMRT, and rotational IMRT for several clinical treatment
plans including prostate, lung, and head and neck sites. Agreement with the Eclipse treatment
planning system (Varian) planned dose was observed to be within 2%/3 mm using absolute dose.

In 2009, Wendling et al. [34] extended to 3D that group’s previous work reconstructing a 2D plane
of dose within the patient [35]. A correction-based approach was taken, accounting for the dose
response of the EPID as well as the self-scatter of the EPID detector. Scatter-to-primary ratios were
measured for a range of homogeneous slab thicknesses and then used to estimate the scatter from the
patient to the EPID. Patient attenuation and the inverse square law were handled. The spectral effects
of beam hardening were accounted for through a depth-dependent attenuation correction factor. An
in-house developed patient dose calculation algorithm was used, where measured depth-dependent
patient scatter kernels were employed to estimate scatter dose contributions within the patient. Several
clinically realistic treatments were tested including a prostate, a rectum, and a head and neck IMRT
plan. Homogeneous phantoms were used with a 3%/3 mm gamma comparison criteria, comparing
high dose voxels (defined as voxels within the 50% isodose surface). The comparison demonstrated
that the percentage of pixels in agreement (i.e. with gamma<1) was 99.9%, 98.1, and 96.6% for the
prostate, rectum, and head and neck treatments, respectively. This work was recently validated for
rotational IMRT [24] where tests included two prostate treatments, a lung SBRT, and a head and neck
treatment. Tests run with an homogeneous phantom demonstrated differences in dose at isocentre of
In 2011, this work was further extended to make use of CT imaging data to calculate beam transmission [45]. This is attractive since it avoids the need to acquire a set of ‘open field’ EPID images (i.e. treatment field images but without any patient or phantom present), previously required by this method to generate a beam transmission factor image.

5. Clinical experience using 3D in vivo patient dosimetry with EPIDs

Of the in vivo techniques described in the previous section, there is currently only one that has been clinically implemented on a large scale. The methods developed at the Antoni van Leeuwenhoek Hospital of the Netherlands Cancer Institute [28, 29, 34, 35] have been used clinically since 2005. A retrospective analysis of nearly five years of clinical data collected was presented by Mans et al. [46]. The value of an in vivo EPID dosimetry program was clearly demonstrated, with 17 serious errors caught over 4337 patient plans analysed between January 2005 and August of 2009.

6. Summary

This paper presents a brief overview of a-Si EPIDs in the context of dosimetric applications focusing on 3D patient dosimetry. Clinically available a-Si EPID systems were discussed, highlighting dosimetric characteristics and remaining challenges to routine use. Available methods to use the EPID for dosimetry applications were outlined. A summary of the use of a-Si EPIDs during treatment to estimate 3D dose in the patient was presented.

Today’s modern radiotherapy clinics are implementing more complex treatments such as dynamic IMRT and rotational IMRT, and there is increasing demand for specialized treatment techniques that incorporate large doses per fraction and short treatment courses (i.e. hypofractionation and stereotactic radiosurgery). These factors strongly motivate researchers to continue investigating EPIDs for in vivo patient treatment verification.

7. Acknowledgements

The author gratefully acknowledges the contributions of our research group team members, past and present, including Eric van Uytven, Tim van Beek, Pete McCowan, Ganiyu Asuni, Krista Chytyk, and Stephen Pistorius. Our Australian collaborators in Newcastle led by Peter Greer and including Pejman Rowshanfarzad, Mahsheed Sabet, Henry Woodruff, Todsaporn Fuangrod, Brian King, and Maura Monville, are also acknowledged for their many contributions.

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