Dose–response and ghosting effects of an amorphous silicon electronic portal imaging device


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The purpose of this study was to investigate the dose–response characteristics, including ghosting effects, of an amorphous silicon-based electronic portal imaging device (a-Si EPID) under clinical conditions. EPID measurements were performed using one prototype and two commercial a-Si detectors on two linear accelerators: one with 4 and 6 MV and the other with 8 and 18 MV x-ray beams. First, the EPID signal and ionization chamber measurements in a mini-phantom were compared to determine the amount of buildup required for EPID dosimetry. Subsequently, EPID signal characteristics were studied as a function of dose per pulse, pulse repetition frequency (PRF) and total dose, as well as the effects of ghosting. There was an over-response of the EPID signal compared to the ionization chamber of up to 18%, with no additional buildup layer over an air gap range of 10 to 60 cm. The addition of a 2.5 mm thick copper plate sufficiently reduced this over-response to within 1% at clinically relevant patient–detector air gaps (>40 cm). The response of the EPIDs varied by up to 8% over a large range of dose per pulse values, PRF values and number of monitor units. The EPID response showed an under-response at shorter beam times due to ghosting effects, which depended on the number of exposure frames for a fixed frame acquisition rate. With an appropriate build-up layer and corrections for dose per pulse, PRF and ghosting, the variation in the a-Si EPID response can be reduced to well within ±1%.

Key words: portal dosimetry, amorphous silicon EPID, radiation dosimetry, dose response, flat-panel imagers

I. INTRODUCTION

Electronic portal imaging devices (EPIDs) were originally designed and developed for the purpose of geometric verification of patient set-up during treatment. However, their use has been increasingly extended to also obtaining dosimetric information of the radiation treatment, either by pre-treatment verification or by means of in vivo dosimetry. Despite the obvious advantages of an on-line, 2-D EPID dosimetry system, extensive clinical use has been limited for a number of reasons. The use of EPIDs in the clinic for set-up verification is still limited worldwide, and there has been little support from vendors to invest extensively in dosimetric applications. Verification of the dose in two or three dimensions in the patient is complicated, which is especially significant if portal dosimetry is to be used as an independent check of the treatment planning system. Furthermore, most departments will have limited resources for the development of software required for portal dosimetry.

There are various possible approaches to portal dosimetry. One is to calculate the dose at the plane of the detector behind the phantom or patient. This usually requires development of in-house algorithms to predict the portal dose image (PDI), since this option is not yet widely available in commercial treatment planning systems. An alternative approach is to take the PDI and predict the dose in a plane in the phantom or patient, since it is often more interesting to know and verify the dose in the target volume than at the EPID plane. This approach has been applied using an algorithm to back-project the PDI to the mid-plane or to reconstruct the patient dose based on convolution/superposition methods, back-projecting the fluence measured with the EPID.

Applications of portal dosimetry have been reported mainly for the type of EPIDs based on a liquid-filled (Li–Fi) ionization chamber matrix,1–3,5–7 the CCD camera-based EPID8–13 and a solid-state-based EPID.14 These studies compared the PDI with various combinations of film and ionization chamber measurements. The accuracy in dose prediction of 3% for most of these studies has usually been limited to point doses on the central beam axis, low dose gradient regions and/or the use of homogeneous phantoms.

More recently, amorphous silicon (a-Si)-based EPIDs have been developed with excellent image quality, relatively high optical transfer efficiency, large imaging area and high resistance to radiation damage.15,16 Moreover, the response of the a-Si EPID has an excellent long-term signal reproducibility of 0.5% (1 SD), as observed in our department.17 Use of the a-Si EPID has also been extended to dose determination in recent years.18–21 A reported advantage of the a-Si EPID is its linear dose–response relationship, whereby a simple calibration factor may be applied to convert the EPID signal to the absolute dose. A number of these studies have reported a linear signal better than 1%, for various commercial types of a-Si panels.

In addition to quantifying the EPID dose–response rela-
tionship, it is important to determine the amount of buildup layer required, as in any dosimetric system applied to high energy photon beams. EPIDs are typically designed today with an intrinsic 1 mm copper (Cu) plate covering the phosphor layer, used to absorb low energy, scattered radiation which would otherwise reduce the image contrast. This Cu plate also provides some buildup to convert primary x-rays into Compton electrons.22 Two reasons for using additional buildup material for portal dosimetry are (1) to ensure measurements are made beyond the dose maximum and (2) to attenuate scattered radiation from the patient. The latter is especially relevant at small patient–detector air gaps (< 40 cm), in which case the scatter contribution from the irradiated volume (of a phantom or patient) to the EPID dose may be considerable.23,24 Yeboah and Pistorius24 investigated the dependence of the EPID dose–response on scatter from a phantom using Monte Carlo calculations for a copper-phosphor portal imaging screen. They found a broad peak in the photon spectra at low energies (50–100 keV) and observed a high sensitivity of the imager for low energy photons. Previous authors have reported on the need for additional buildup layer material such as polyethylene (PS),19,25 stainless steel22 and Cu23,26 for EPID dosimetry applications. As suggested by Partridge et al.,23 a sharp rise in the mass attenuation coefficient of Cu below 500 keV indicates that this material should preferentially filter out low energy photons.

Other reported phenomena in imaging are ghosting effects, namely image lag and change in sensitivity or gain. These may be distinguished as separate effects,27 reportedly primary due to the trapped charge in the photodiodes.28,29 Image lag is a signal delay, so charge generated in one image frame is read out in subsequent frames, adding an offset to the signal in subsequent recorded frames or images.28 Previous reports have studied image lag by measuring the residual signal in the dark field, lasting in the order of minutes following beam off.27,29–31 The extent of the image lag has been found to be highly dependent on the number of frames read out, or acquired, as opposed to an explicit dependence on the time of irradiation. Therefore one would expect a change in the influence of ghosting on the response for alternative or irregular frame acquisition rates. If the frame acquisition rate is fixed, the two parameters, beam time and number of frames, can be said to be interchangeable. Another type of ghosting, while also related to charge trapping, has been associated with a change in gain (or sensitivity). During exposure, the charge stored in deep trapping states alters the electric field strength within the photodiode bulk and interface layers. This will consequently change the sensitivity of the a-Si layer.26,31 Gain ghosting is multiplicative, influencing read out of both the signal at the time of exposure and any delayed signal. Properties of, and corrections for, both image lag and change in gain have been investigated for normal portal imaging applications of a-Si detectors in the above mentioned studies. However, the effect has not yet been reported in terms of a-Si portal dosimetry, where both types of ghosting will influence the measured dose–response relationship.

To use the EPID for dosimetry, ideally the absolute response of the EPID should be determined, i.e., the signal of the EPID per unit dose from photons, yielding an energy dependent dose–response curve. This is impractical, however, because the output of the accelerator is not monoenergetic. More importantly, the purpose of this study was to calibrate the measured portal signal against dose in air, approximated by the dose measured with an ionization-chamber in a mini-phantom at the EPID plane. This calibration would then later be used with a back projection algorithm to determine the dose within the phantom or patient. This work extends previous a-Si EPID investigations to include all relevant parameters that change under clinical conditions, such as dose, dose per pulse and pulse repetition frequency (PRF). Therefore the objectives of this study were to investigate the amount and type of material required as a buildup layer for dosimetry and the dose–response characteristics, including ghosting effects, of the a-Si EPID.

II. METHODS AND MATERIALS

The type of EPID used throughout all experiments is an amorphous silicon flat panel-type imager (Elekta iViewGT). It has a 41 × 41 cm² detection area (1024×1024 pixels), a touch guard, a 1 mm Cu buildup layer, a phosphor screen and a hydrogenated a-Si:H photodiode array. Further details regarding similar types of imagers can be found in an overview by Antonuk.32

An EPID image is defined as the stored signal—this may be the signal integrated or averaged over all or a specific number of frames. A frame is defined as the signal from one readout of the entire panel. For this study, an acquisition mode was used whereby a frame is taken every 285 ms during acquisition and 2 pre- and 2 post-beam frames are added to ensure that dose from partially irradiated frames at beam-on and beam-off are included. An average of all frames is then stored as a raw image, *I* _raw_ , or alternatively, it is also possible to store all individual frames, as required for parts of this study. All images are processed to correct for individual pixel sensitivity and offset. The processed image *I* _proc_ is then

\[
I_{\text{proc}} = \frac{I_{\text{raw}} - I_{\text{dark}\_\text{dyn}}}{I_{\text{flat}} - I_{\text{dark}}},
\]

where *I* _flat_ is an open flood-field image encompassing the sensitive area of the detector and *I* _dark_ is a dark field image (i.e., acquired without irradiation) which serves as an offset correction. An arbitrary scaling factor of 8192 is included to be able to store the signal data in a 16-bit format. The calibration images (*I* _flat_ and *I* _dark_) are typically acquired when the EPID is installed or any changes in the set-up are made. *I* _dark\_dyn_ is a new dark field acquired prior to image acquisition. This updated dark field compensates for any dependence on ambient temperature or radiation history in the signal.

Reference dose measurements were performed with an ion chamber (Semiflex 0.125 cm³, PTW Freiburg) and electrometer (Keithley #617). The reference dose (*D* _ref_ ) was
measured at the dose maximum in a polymethylmetacrylate (PMMA) mini-phantom, with a diameter of 4.0 cm for 4, 6 and 8 MV and 5.0 cm for 18 MV. The effective point of measurement was at the depth of dose maximum for each energy. The mini-phantom was placed at the same source–surface distance (SSD) as the imager touch guard surface (with a touch guard approximately 3 cm above the scintillator screen) such that the effective point of measurement for both detectors was reproducible and similar. For some measurements it was necessary to compare reference and EPID dose-rate signals during irradiation, in these cases a p-type diode (Scanditronix Medical AB, Uppsala, Sweden) positioned on the EPID was used as a reference detector. All ionization chamber and diode measurements were relative, except those concerned with the dose per pulse, whereby the absolute dose to water was determined in the mini-phantom (with corrections for measuring in a PMMA medium). The EPID signal (\(S_{\text{EPID}}\)) was calculated by multiplying the average pixel value of 20×20 central pixels from \(I_{\text{proc}}\) by the number of frames acquired (stored in the image file header). The EPID response (\(R_{\text{EPID}}\)) was then defined as the ratio of the EPID signal and the reference dose:

\[
R_{\text{EPID}} = \frac{S_{\text{EPID}}}{D_{\text{ref}}} = \frac{I_{\text{proc}} \times (\# \text{ frames})}{D_{\text{ref}}},
\]

Measurements were made using two detectors (based on the same design) mounted on two linear accelerators, with photon beams of 4 and 6 MV (Elekta SL-15i, with EPID A) and 8 and 18 MV (Elekta SL-20i, with EPID B), respectively. A number of measurements was repeated using a prototype a-Si flat panel device (RID 1640 AF2, Perkin Elmer Optoelectronics, EPID X). This imager has the advantage of mobility and is similar in design to the commercially available EPIDs. However, EPID X has a different type of amplifier, producing 1/3 of the output signal of EPIDs A or B. When attached to the linear accelerator, SSD of all EPIDs (to the surface of the touch guard) was fixed at 156.9 cm. EPID X (which did not have a touch guard) was placed at 161.0 cm when the SSD was not varied, to ensure the sensitive areas of both panels were at a comparable SSD.

### A. Buildup layer and air gap

In this part of the study, \(R_{\text{EPID}}\) was measured over the maximum possible air gap range of 10–60 cm, defined as the distance between the table exit surface on which the phantom is placed and the surface of the EPID (Fig. 1). This was performed for two different buildup layer materials, Cu and polystyrene (PS). To compare the influence of these two buildup materials, the thickness of each layer was chosen according to approximate equivalent density thicknesses: 22.0 mm of PS and 2.5 mm Cu. A comparison of the two buildup layer materials of these particular thicknesses were chosen, as they were similar to the layer thicknesses used in other studies measuring the dose-response of an a-Si EPID. The additional layer (polystyrene or copper) was positioned on top of the imager touch guard. A polystyrene slab-geometry phantom was used in all cases, with a thickness of 15 cm for the 4 and 6 MV beams and 35 cm for the 8 and 18 MV beams, which are typical patient thicknesses for these energies. The EPID response was measured with a 15×15 cm² field, 100 monitor units (MUs) at a PRF of 400 Hz. The air gap was varied by changing the height of the table on which the phantom was placed.

In addition to determining the influence of the material used, the required thickness of the buildup layer was also studied. The response, \(R_{\text{EPID}}\), was measured as a function of the air gap with additional buildup layers of 1.0, 2.5 (as above) and 5.0 mm Cu for the 4 and 6 MV photon beams (EPID A). For 8 and 18 MV (EPID B), the response was measured with only 2.5 and 5.0 mm Cu layers, since at least 2.5 mm Cu buildup was required for measuring beyond the dose maximum.

### B. Dose–response relationships

The EPID response was determined by varying dose delivery parameters at the position of the detector in various ways. Physical properties of the beam that were modified included variation in the dose per pulse, PRF and total dose. All measurements were made with open fields and an additional 5.0 mm Cu buildup layer.

First, \(R_{\text{EPID}}\) was measured at different SSDs ranging from 85 to 334 cm, to vary the dose per pulse from 0.08 to 1.40 cGy/pulse. To allow such a large SSD range, the gantry was rotated 90 degrees and the EPID was detached. The detector was then positioned at successive source–surface distances, keeping the detector plane perpendicular to the beam direction at each measurement position. The SSD range was limited by the distance from the linac head to the opposing wall of the treatment room. To avoid changes in the EPID lateral scatter dose component, the field size was adjusted, maintaining a constant effective field size at the sensitive layer of the detector of 20×20 cm². All beams were delivered with 100 MUs at the maximum PRF for each energy (200 Hz for 4 and 18 MV, 400 Hz for 6 and 8 MV). In this part of the study, EPID B was used at both machines to compare the dose per pulse dependence of all four energies with EPID X.

Second, \(R_{\text{EPID}}\) was measured by varying the dose-rate setting at the treatment machine, effectively changing the beam PRF from 13 to 400 Hz for each energy, except 4 and 18 MV.
which have a maximum PRF of 200 Hz. The change in response was measured in two ways—first delivering a series with a varying number of MUs, keeping the beam time constant at 50 s, followed by a second series delivering a constant dose of 100 MUs, which varied the beam time.

Finally, $R_{\text{EPID}}$ was measured by varying the total dose delivered using a fixed PRF over a range of 5 to 1000 MUs at maximum dose-rate setting for each energy. Dose-response measurements were performed with EPID A for 4 and 6 MV, EPID B for 8 and 18 MV and EPID X for all energies, unless otherwise specified.

### C. Ghosting

Two sets of measurements were made using EPID A with 6 MV photon beams delivered at PRFs of 400, 200, 50 and 25 Hz. In the first set, the beam time was kept constant (12 s) and the total dose (number of MUs) was correspondingly varied. In the second set, the total dose was kept constant (100 MUs) and the beam time was varied. Instead of storing an average image for each beam, the signal from each frame was recorded during and after irradiation. The decay of the signal was measured over time following beam-off.

Prompted by results from studying the signal decay, another set of measurements was performed to investigate the cumulative effect of ghosting over all frames, both during and after irradiation. Images were acquired for a series of beams ranging from 5 to 1000 MUs, at 2 PRFs (400 and 200 Hz), for 8 and 18 MV photon beams with EPID B. All frames were stored (every 285 ms) while simultaneously recording the relative dose rate with a diode as a reference detector. Measurements were continued following beam-off, until the EPID signal dropped below 0.1% of the maximum signal during exposure.

The results for $R_{\text{EPID}}$ as a function of PRF (for varying beam time) and number of MUs were expressed as a function of beam time and combined to determine a beam time dependency relationship for the EPID response. A fit was made of the data set to then use as a “ghosting correction” for response measurements as a function of PRF and number of MUs. The time of each irradiation was determined from the number of frames (read every 0.285 s) acquired for each image.

### III. RESULTS

The reproducibility of the ionization chamber, diode and EPID signal readings, was ±0.5% (1 SD). In each set of response measurements, $R_{\text{EPID}}$ was normalized to the maximum value, except for the PRF measurements, where results were normalized to 200 Hz, the maximum PRF common to all energies.

#### A. Buildup layer and air gap

Figure 2 presents data for the variation in $R_{\text{EPID}}$ over an air gap range of 10 to 60 cm with two different buildup materials, copper (Cu) and polystyrene (PS), for 4 and 6 MV beams. The over-response at smaller air gaps (relative to the reference detector distance of 60 cm) can be attributed to a change in the number of photons scattered from the phantom to the EPID. This over-response is reduced from 16% to 12% for 4 MV and from 12% to 8% for 6 MV when a Cu buildup layer is used compared with an equivalent density thickness of PS.

$R_{\text{EPID}}$ is presented in Fig. 3 over a range of air gaps for different thicknesses of the Cu buildup layer. The response curves increase at small air gaps for 4, 6 and 8 MV beams. This over-response was minimized by using an additional 5.0 mm Cu buildup layer, especially at lower beam energies. Results for the minimum clinically relevant air gap of 40 cm are given in Table I. At this air gap, the greatest reduction in the over-response was found using the thickest Cu layer (5.0 mm) and the lowest beam energy (4 MV), where the over-response was reduced from 2.2% to 0.1%. At 18 MV, the response remains within 1% over the entire air gap range, independent of the Cu buildup layer thickness. The response for an air gap of at least 40 cm was within the limits of accuracy of 1% for all energies with an additional 5.0 mm Cu buildup layer, and less than 1.1% with 2.5 mm Cu.

#### B. Dose–response relationships

For EPID B (measured at 4, 6, 8 and 18 MV), $R_{\text{EPID}}$ is not constant as a function of the dose per pulse, with a change in response up to 8% relative to SSD=100 cm, over the dose per pulse range [Fig. 4(a)]. For measurements with EPID X [Fig. 4(b)], the variation in response over the investigated range of dose per pulse values is within 4%. The response curves show no clear dependence on beam energy. This implies that corrections of the EPID response are needed based on the dose per pulse, which would require knowledge of the thickness of the attenuating medium (e.g., phantom, patient or wedge) for a fixed source-EPID distance. The gradient of the response curves for both EPIDs is steeper at low dose per pulse values (<0.5 cGy/pulse to the reference detector). This corresponds to a reduction in measured dose of over 70%.
The EPID response as a function of PRF is given in Fig. 5 for 13 to 400 Hz. For the commercial EPIDs (A and B), \( R_{\text{EPID}} \) falls by 6% with constant beam time [Fig. 5(a)] and 5% for a constant number of MUs [Fig. 5(b)]. Both curves show a sharper decrease in response at lower PRF values, independent of beam energy. Results for the prototype EPID (X) show less variation in \( R_{\text{EPID}} \) over the same range, with the response remaining within 2% for a constant beam time [Fig. 5(c)] and a constant number of MUs [Fig. 5(d)]. Since the PRF is always known prior to treatment, it is a simple correction to apply if beams are delivered at a PRF other than that at which the EPID is calibrated.

Finally, \( R_{\text{EPID}} \) was also found to vary as a function of number of monitor units for all four energies and both EPID-types, as shown in Fig. 6. Relative to the maximum dose (1000 MUs), the under-response was up to 5% over the entire dose range for the commercial EPID and 6% for the prototype EPID. The EPID response curve is very similar for both EPID types. The response varied by 3% for beams of 5 to 100 MUs, which was the range of MUs where the response gradient was steepest. This suggested either a beam time or dose dependence of the EPID response, which prompted further investigation.

C. Ghosting

The signal of the dark field following irradiation is presented in Fig. 7. All data is normalized to the maximum irradiated signal. The series in which the beam time was kept constant [12 s, Fig. 7(a)], the decay was very similar for all curves. In the following series for irradiations of equal dose [100 MUs, Fig. 7(b)], the decay in the image signal was faster for shorter beam times. This suggests that the decay rate of the EPID signal depends primarily on beam time, and not on dose or PRF.

Figure 8(a) shows the dose rate signal as measured by the EPID and diode, acquired simultaneously during and following irradiation. Results are given for 1000 MUs delivered with an 8 MV photon beam at maximum PRF (400 Hz). The signal of each frame is normalized to the signal at 100 s, at which the signals have reached a stable value. Comparing EPID and diode dose rates, the EPID takes approximately 40
s to reach to within 0.5% of its maximum value, whereas the diode signal is within this range after only 8 s. The three curves in Fig. 8(b) represent an ideal, constant response, the measured response ($R_{\text{EPID}}$), and a corrected measured response, $\bar{R}_{\text{EPID} + \text{lag}}$. This corrected response includes the integrated image lag signal measured after beam-off (as shown in Fig. 7). The measured response varies up to 5%, while the corrected response is only improved by at most 1.5%, with up to 3.5% remaining below the ideal case. Therefore the post-beam signal is not enough to account for the slower rise in EPID signal, compared with the reference detector at beam-on [Fig. 8(a)]. This suggests that the ghosting effect is more than only image lag, and an alternative correction method is required based on the irradiation time.

To determine a ghosting correction factor, $G(t)$, an exponential fit was made for the normalized $R_{\text{EPID}}$ as a function of beam-on time (Fig. 9). The function was accurate for all energies and dose rates measured to within 0.5% for beam-on times greater than 2 s, corresponding to 5 MUs at a maximum PRF value of 400 Hz. The correction function applied took the form of a triple-exponential,

$$G(t) = A_0 - A_1 \exp(-r_1t) - A_2 \exp(-r_2t) - A_3 \exp(-r_3t),$$

where $A_0$ to $A_3$ were fitted coefficients (1.000, 234.3, 0.036, 0.026) and $r_1$ to $r_3$ were the fitted decay rates (7.8, 0.46, 0.034 s$^{-1}$) of the combined ghosting effects. Modeling the curve with up to 3 time constants was necessary to match measured data to within 0.5%.

The corrected response of the EPID, $R_{\text{EPID} \cdot G}$, was then determined by the EPID signal ($S_{\text{EPID}}$), the reference dose ($D_{\text{ref}}$) and the ghosting correction factor $G(t)$, based on the exposure time determined from the number of frames acquired:

$$R_{\text{EPID} \cdot G} = \frac{S_{\text{EPID}}}{D_{\text{ref}} \times G(t)}.$$  \hspace{1cm} (4)

The ghosting correction was applied to all EPID A and B response measurements involving varying exposure times (examples are given in Fig. 10). For all energies, $R_{\text{EPID} \cdot G}$ as a function of the number of MUs was effectively linear within ±1% of the measured dose [4 and 18 MV data are shown in Fig. 10(a)]. For the response as a function of PRF, the dependence on beam time is removed and both curves representing $R_{\text{EPID} \cdot G}$ overlap [Fig. 10(b)], indicating a true PRF dependence. The effect of the correction was similar for all energies. Without the correction, the under-response at lower PRF values is partly compensated by the increased response due to ghosting at longer beam times.

IV. DISCUSSION

Properties of the a-Si EPID were investigated in this study for the purpose of understanding the EPID response behavior, prior to calibration for patient treatment evaluation. Whether the dose is to be compared at the portal plane, or a back-projection method is applied, the dose–response characteristics and ghosting effects should be well understood for EPID dosimetry.

Applying an adequate buildup layer for a-Si EPID dosimetry applications is important for three reasons: (1) to absorb low-energy electrons before reaching the EPID sensitive layer which would reduce image quality; (2) to ensure electron equilibrium at the sensitive layer of the detector during dosimetry; and (3) to minimize the additional photons scattered from the patient and measured by the EPID. For each of these issues, the thickness of buildup required depends on the energy applied, but not necessarily in the same way. Measurements from this study highlighted the benefits of using Cu instead of PS buildup for the absorption of patient scatter at small air gaps. At lower, scattered photon energies, the photoelectric effect dominates. Its cross section is dependent on the atomic number ($\tau \propto Z^3$), which becomes negligible at energies above 100 keV. The inclusion of a higher $Z$ material absorbing layer such as copper meant more scattered photons were absorbed than with the equivalent density.
thickness of polystyrene. It is also worth noting that the copper plate was more convenient than polystyrene as buildup layer material because it occupied less volume (limited by the presence of the touch guard) within the panel for the same effective path length.

There was an apparent, minor trend for the measurements presented in Fig. 3 of the air gap which gave the minimum response to decrease with increasing energy, i.e., the minimum response was at the largest air gap for 4 and 6 MV, and at mid-range air gaps for 8 MV and 18 MV. This could be due to two opposing effects. At smaller air gaps, the phantom is farther from the source and the dose to the phantom is reduced by the $1/r^2$ law. The secondary photons are scattered over a broader angle, so fewer low energy photons per unit phantom volume reach the center of the detector. On the other hand, the total scattering volume is larger and the exit plane is closer to the EPID. This increases the scatter to primary ratio of photons reaching the EPID. The rate at which each of these effects increases or decreases the amount of scatter from the phantom detected by the EPID depends on the beam energy and Cu buildup layer thickness. However, detailed Monte Carlo calculations modeling this particular experiment would be required to fully understand the observed results.

With a 5.0 mm additional Cu buildup layer, the over-response of the EPID due to patient scatter was almost completely eliminated, especially for lower energies where the over-response was generally higher. However, adding 2.5 mm or 5.0 mm thick Cu layers (area = $41 \times 41$ cm$^2$, Cu density = 8.9 g/cm$^3$) increases the load on the EPID-arm by either 3.7 kg or 7.5 kg, respectively. For 8 and 18 MV, 2.5 mm Cu was sufficient buildup thickness, and the weight was considered acceptable for clinical use. Additional fatigue calculations would be required to justify the extra load on the EPID-arm. Another potential constraint to the buildup layer

Fig. 5. The EPID response as a function of the PRF for 4 MV (triangles), 6 MV (circles), 8 MV (squares) and 18 MV (diamonds). EPIDs A and B with the same beam time (50 s) are given in (a), the same number of MUs (100 MUs) in (b). The corresponding curves for EPID X are given in (c) and (d). Results are normalized at 200 Hz, the maximum PRF for 4 MV.

Fig. 6. The EPID response as a function of dose for a varying number of monitor units with EPIDs A, B and X, for 4 MV (triangles), 6 MV (circles), 8 MV (squares) and 18 MV (diamonds). The maximum possible PRF was used for each energy.
was reduced image quality. While the intrinsic buildup layer of 1 mm Cu is used to reduce blurring due to low energy scattered electrons, adding too much Cu may generate an excess amount of scatter and have the opposite effect. Partridge et al. indicated that there is no significant effect on the spatial resolution of this type of panel by adding 3 to 4 mm Cu.27 Line spread function measurements made to study the effect of adding 5.0 mm Cu on image quality at our institution showed negligible deterioration (data not shown). Qualitative checks involving blind tests to distinguish bony anatomy in a Rando-Alderson head phantom were also performed by a radiation technologist. Though the images made with additional Cu buildup layer could be distinguished, it was found the slight reduction in image quality was not significant for patient treatment set-up verification.

The EPID response was different for both EPID types when the dose per pulse and the PRF were varied. This may be related to differences in design of the EPIDs used for this study, primarily in the types of amplifiers each type uses.

Wischmann et al. have attributed the amplifier as a major source of nonlinearity in the EPID signal, though some corrections to the gain are applied to both EPIDs, this might be one possible explanation.29 For both the dose per pulse and PRF, the dose per frame is varied, however, we have no physical explanation for the observed EPID response variation. While they are linear by design as long as the photodiodes are not saturated, it is possible that various amplifiers will have different sensitivities over the same signal level range. These results emphasize the need for a unique investigation of the dose–response characteristics for any type of EPID used for dosimetric applications.

While dependence of the EPID response on PRF may only amount to a few percent, it is important to be aware of the characteristic curves for each imager applied for dosimetric applications. Corrections for PRF would rarely be required for the EPIDs used in this study, since beams deliv-
ered at a PRF other than the maximum (at which the EPID is calibrated) are rare in the clinic, and the PRF would have to be reduced by a factor of 3 (which is very rare) before the EPID response will change by more than 2% (Fig. 5).

The dose per pulse dependence with varying SSD was recently investigated by Greer and Popescu\(^\text{21}\) and found to be linear within 1%, this was consistent with the results of this study, however, it was over a smaller range, and a detector from a different manufacturer was used. The dose per pulse dependence of the EPID signal is significant for dosimetry because not only is it dependent on SSD, which is usually fixed, but also on the attenuation of the beam (i.e., the thickness of the phantom, patient or wedge). According to the response curve determined in this study (Fig. 4), the signal response drops significantly at dose per pulse values below 0.5 cGy/pulse. This is similar to the attenuation of an 18 MV photon beam by a 40 cm thick phantom (lateral field) or by a 60° wedge. Therefore in most clinical situations, the dose per pulse dependence is closer to within 3% for both types of EPID (excepting beams with thick wedges). However, the variation of dose per pulse based on attenuator thickness would still need to be determined for a full calibration of the EPID. This is because of the spectral change induced when the dose per pulse is reduced by an attenuating medium, especially for higher Z materials such as that of wedges.

The dose–response as a function of the number of MUs was similar for both EPID types. In clinical cases, most treatment prescriptions lie within the range of 20–300 MUs, the steeper part of the response curve (Fig. 6). IMRT fields are an exception to this, with step-and-shoot segments down to 3 MUs not uncommon. If beams are prescribed with segments ranging from 5 to 50 MUs, for example, the EPID response will vary by 3% if no corrections for ghosting effects are applied. One solution would be to continue reading out the charge in the dark field following exposure and include this in the integrated EPID signal. However, it is not often clinically practical to continue acquiring images after irradiation (especially between segmented fields given in quick succession), and in any case, our measurements showed that the method does not sufficiently account for the signal missed in the first 40 s of irradiation. An alternative solution proposed in this report involves determining a fit of the response curve as a function of beam time. Since previous studies have shown that ghosting effects depend on the number frames acquired\(^\text{27,28}\) and in all our studies the frame acquisition rate is fixed, the time and number of frame parameters may be considered equivalent. This is then applied as a universal correction for ghosting effects of all images acquired, independent of the beam energy, PRF setting or number of monitor units. This suggests that the solution is especially ideal for verifying segmented IMRT fields, despite the fact that it is not derived from a theoretical or physical basis. Most importantly, the pragmatic approach is a simple means of ensuring the calculated dose based on EPID measurements,
once corrected for PRF and ghosting, is approximately linear with the reference dose within 1%.

V. CONCLUSIONS

The response of the a-Si EPID was found to vary by up to 18% over an air gap range of 10 to 60 cm. This variation was reduced to within 1% for clinically relevant air gaps (>40 cm) with an additional 2.5 mm Cu buildup layer on the EPID during dosimetric applications. The measured EPID response was significantly dependent on the type of imager used, even between imagers based on a similar design. For the commercial EPID used in this study, the signal for varying PRF and dose per pulse is nonlinear over the range measured, with the maximum variation in response up to 8%. For clinically relevant ranges, a small correction would be required for these settings at the time of treatment. Nonlinearities up to 6% also persist for total dose measurements (varying MUs) for both EPID-types over the applied energy range. This was explained by the combined ghosting effects of image lag and changing sensitivity. Ghosting effects are found to depend on the number of acquired frames, not on the dose or PRF, within the range measured. Modeling the change in response as a function of beam time is a successful, pragmatic correction to obtain a constant dose-response to within ±1% up to 1000 MUs.

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