Comparison of forward- and back-projection in vivo EPID dosimetry for VMAT treatment of the prostate

James L Bedford, Ian M Hanson and Vibeke N Hansen

Abstract
In the forward-projection method of portal dosimetry for volumetric modulated arc therapy (VMAT), the integrated signal at the electronic portal imaging device (EPID) is predicted at the time of treatment planning, against which the measured integrated image is compared. In the back-projection method, the measured signal at each gantry angle is back-projected through the patient CT scan to give a measure of total dose to the patient. This study aims to investigate the practical agreement between the two types of EPID dosimetry for prostate radiotherapy. The AutoBeam treatment planning system produced VMAT plans together with corresponding predicted portal images, and a total of 46 sets of gantry-resolved portal images were acquired in 13 patients using an iViewGT portal imager. For the forward-projection method, each acquisition of gantry-resolved images was combined into a single integrated image and compared with the predicted image. For the back-projection method, iViewDose was used to calculate the dose distribution in the patient for comparison with the planned dose. A gamma index for 3% and 3 mm was used for both methods. The results were investigated by delivering the same plans to a phantom and repeating some of the deliveries with deliberately introduced errors. The strongest agreement between forward- and back-projection methods is seen in the isocentric intensity/dose difference, with moderate agreement in the mean gamma. The strongest correlation is observed within a given patient, with less correlation between patients, the latter representing the accuracy of prediction of the two methods. The error study shows that each of the two methods has its own distinct sensitivity to errors, but that overall the response is similar. The forward- and back-projection EPID dosimetry methods show moderate agreement in this series of prostate VMAT patients, indicating that both methods can contribute to the verification of dose delivered to the patient.

1. Introduction

The electronic portal imaging device (EPID) (Greer and Popescu 2003, Louwe et al 2004, McDermott et al 2004, Winkler et al 2007) is used widely in radiotherapy for both pre-treatment and in vivo dosimetric verification (Wendling et al 2006, McDermott et al 2007, see also reviews by van Elmpt 2008 and Mijnheer et al 2013). Of these two situations, in vivo EPID dosimetry is the more challenging due to the requirement to model the patient in the process of relating the internal dose and the external EPID signal. These two classes of verification can be further divided into whether they relate to intensity-modulated radiation therapy (IMRT) (Wendling et al 2009, Millin et al 2016) or volumetric modulated arc therapy (VMAT) (Mans et al 2010a, Fidanzio et al 2014, McCowan et al 2015, McCowan and McCurdy 2016), the former referring to fixed gantry angles with either segmental (step-and-shoot) or sliding window delivery, and the latter referring to intensity modulation delivered with the gantry rotating continuously.

A further subdivision of applications in EPID dosimetry relates to whether forward-projection or back-projection is used in the technique. In the forward-projection method, the beam geometry and planned dose distribution are used to predict at the time of treatment planning the EPID signal that is expected to occur during a fraction of treatment (van Zijtveld et al 2009, Fogliata et al 2011, Berry et al 2012, Chytyk-Praznik et al 2013,
To do this, the dose or fluence at the planning target volume (PTV) is projected to the plane of the EPID and the response function of the EPID panel applied to give the expected EPID signal. After treatment delivery, the predicted and measured EPID signals are compared to give an indication of the accuracy of the treatment. Conversely, the back-projection method is applied after measurement (van Elmpt et al 2007, Wendling et al 2009, Slosarek et al 2010, Mans et al 2010b, McCowan et al 2015, 2017, McCowan and McCurdy 2016, Van Uytven et al 2015). The measured EPID signal at each gantry position is projected back to the PTV to give the actual dose that the patient received, either at a point (Nijsten et al 2007, Slosarek et al 2010, Francois et al 2011), or over a volume (Mans et al 2010b). Both methods involve the use of a scatter model to incorporate the presence of photon scatter in the patient and the EPID panel.

The advantage of the forward-projection method is that it does not necessarily require the measured images to be labelled with gantry angle. For example, in VMAT verification, the predicted EPID images can be summed over all control points of the arc to give an integrated image. The measured images can likewise be summed over the whole arc and the predicted and measured images compared. However, methods are now available for time-resolved forward-projection, which may give improved sensitivity to treatment errors (McCurdy and Greer 2009, Persoon et al 2016, Schyns et al 2016). Real-time error detection has also been demonstrated using this method (Fuangrod et al 2013). The back-projection method requires the gantry angle of each image to be known, so that the EPID signal can be back-projected at the correct orientation into the patient. The major advantage of back-projection is that the method gives a 3D distribution of reconstructed dose in the patient, which is easier to interpret than a difference in an EPID signal (figure 1). Online (Spreeuw et al 2016) and 4D (Yoon et al 2016) versions of this approach have also been demonstrated.

This study compares the forward- and back-projection methods of EPID dosimetry for the case of VMAT treatment of prostate and seminal vesicles. Both methods are available clinically at this centre and it has therefore been possible to carry out a unique comparison of techniques in which the same treatment plans and measured images are used for both arms of the study. Several studies have evaluated the sensitivity and specificity of EPID dosimetry methods using phantom measurements (Bedford et al 2014, Schyns et al 2016) and reviews of patient data (Mans et al 2010b, Persoon et al 2016), but there is still scope to better understand the relative performance of the different methods in identifying specific types of errors. This study therefore begins by investigating the relative performance of forward- and back-projection methods in a cohort of clinical patients and then continues to compare the two methods in a phantom scenario, where errors are introduced deliberately into the treatment plans.

2. Methods and materials

2.1. Patients and treatment plans

Thirteen patients were considered in this study. In nine patients, three gold seeds were implanted into the prostate to aid daily image guidance. Patients were CT scanned with 1.5 mm slice thickness on a Brilliance Big Bore scanner (Philips Healthcare, Cleveland, OH) and the scans were transferred to Pinnacle3 (v9.10, Philips Radiation Oncology Systems, Madison, WI) and the relevant volumes were delineated. Three PTVs were created following the protocol for Conventional versus Hypofractionated High-dose intensity-modulated radiotherapy.
for Prostate cancer (CHHiP) (South et al 2008, Dearnaley et al 2012, Tree et al 2013), the innermost (PTV1) consisting of the prostate only, with a margin of 3 mm in all directions except posteriorly, where the margin was 0 mm. The second PTV (PTV2) consisted of an expansion of 3 mm around the first PTV, with the first PTV excluded. The third PTV (PTV3) consisted of an isotropic expansion of 6 mm on the prostate plus seminal vesicles, and the other two PTVs were excluded. Patients without gold seed fiducials were treated with the same set of PTVs, but with margins of 5 mm and 10 mm instead of the 3 mm and 6 mm stated above, and with the rectum specifically excluded from PTV1. The prescribed dose was 60 Gy in 20 fractions (11 patients) or 36 Gy in 6 fractions (2 patients) (table 1).

The dose constraints for planning were those of the CHHiP trial, which were principally that 99% of each PTV should receive >95% of the prescribed dose to that PTV, rectum $V_{50\text{Gy}} < 60\%$, rectum $V_{65\text{Gy}} < 30\%$, bladder $V_{50\text{Gy}} < 50\%$, bladder $V_{60\text{Gy}} < 25\%$, and femoral heads $V_{50\text{Gy}} < 50\%$.

The AutoBeam (v5.7) inverse treatment planning system (Bedford 2009, 2013) was then used to generate VMAT treatment plans for the 6 MV flattened beam of a Versa HD accelerator (Elekta AB, Stockholm, Sweden) (Bedford et al 2013). The plans consisted of a single anticlockwise arc, from gantry angle 120° to 240° in accord with previous beam orientation studies (Bedford et al 1999, Khoo et al 2003) (2/13 patients), or 179° to 181° (11/13 patients), with control points spaced at 2° of gantry angle. Collimator angle was fixed at 2° throughout the arc, so as to spread out any interleaf leakage that might be present. The control points were divided up into groups consisting of 20° of gantry angle. Optimisation then proceeded by optimising one fluence map at the central gantry angle of each control point group, sequencing, and then aperture optimisation. Aperture optimisation (Earl et al 2003) was performed using iterative least squares with a conversion from dose to leaf position so as to adjust leaf positions in response to deviation of dose from the prescribed dose distribution (Xing and Chen 1996, Bedford 2013). During aperture optimisation, the delivery constraints of the Agility multileaf collimator were taken into account. Additional constraints of a minimum gantry speed of 2.5° s⁻¹ and a maximum change in dose rate between adjacent control points of a factor of 4.0 were imposed to ensure prompt and efficient delivery, as well as a minimum segment aperture width of 15 mm to ensure reliable dosimetry. Optimisation was based on a fast convolution dose calculation (Bedford 2002) on a 4 mm × 4 mm grid. Note that the complexity of the treatment plan affected the accuracy with which portal dosimetry could be carried out.

The treatment plans were transferred to Pinnacle3 for final dose calculation using collapsed cone convolution on a 2.5 mm × 2.5 mm × 2.5 mm grid. The difference between the monitor units calculated by AutoBeam and Pinnacle3 was less than 1.0% in 10/13 cases and less than 1.6% in the remaining cases. Patients were treated with daily cone-beam CT imaging.

2.2. Forward-projection EPID dosimetry

After production of the optimised treatment plan, AutoBeam produced an integrated predicted portal image (Bedford et al 2014) with 512 × 512 pixels of 0.8 mm × 0.8 mm size, at distance 1.6 m from the source. For each control point of the arc, the dose at the isocentre plane was first converted into in-aqua dose by deconvolving scatter and reconvolving under unit density. A series of rays were then cast from the source to each pixel i of the imaging panel for each control point, j, of the arc (figure 2). The dose was then projected to the plane of the portal imager to give the in-field component of the portal image:

$$I_{ij}^{\text{in-field}} = d_{ij} \mu a f (s_j) c_i \exp (-\mu (s_j) r_{ij}^m)$$

(1)

where $d_{ij}$ is the dose at the intersection of ray i in segment j with the isocentre plane, $a$ is a factor consisting of the source-to-isocentre distance (1.0 m) divided by the source-to-imager distance (1.6 m) and $f (s_j)$ is the output factor for the imaging panel as a function of field size $s_j$.

The term $c_i$ is a pixel-specific calibration factor for the panel, $\mu$ is an attenuation coefficient and $r_{ij}$ is the equivalent path length from the isocentre plane to the imaging panel along ray i. The exponent $m = 0.6$ corrects for beam hardening and for forward scattered photons produced along the length of the path from isocentre plane to imaging plane (Bedford et al 2014). In the interests of speed, ray tracing was carried out to every fifth pixel (i.e. reducing the resolution to 2.5 mm at isocentre) and then bilinear interpolation was used to calculate the intermediate values.

### Table 1. Summary of PTVs and prescribed doses used in the study for patients with implanted gold seed fiducials.

| Structure    | Volume                | Prescribed dose (Gy)          |
|--------------|-----------------------|------------------------------|
|              |                       | 60 Gy in 20 fractions | 36 Gy in 6 fractions |
| PTV1         | Prostate + 3 mm (0 mm posteriorly) | 60.0 | 36.0 |
| PTV2         | PTV1 + 3 mm           | 57.6 | 34.6 |
| PTV3         | Prostate + seminal vesicles + 6 mm | 48.6 | 28.8 |
The out-of-field scatter component of image intensity was calculated as:

\[ I_{\text{out-of-field}} = d_j a^\sigma g(r_j)(s_j - s_0)H(s_j - s_0) \]  

where \( d_j \) is the mean in-aqua dose in the isocentric plane and \( g(r_j) \) is the out-of-field scatter factor, as a function of \( r_j \), the mean radiological distance from the isocentre plane to the imager plane along the cast rays. \( H \) is the Heaviside step function and \( s_0 \) is a threshold field size, below which no scatter is observed.

The total image for control point \( j \) was then calculated according to:

\[ I_{ij} = (I_{\text{in-field}} + I_{\text{out-of-field}}) \otimes \sum_{n=1}^{2} k_n \]  

where \( k_n \) is a Gaussian scatter kernel. The first kernel, \( k_1 \), had a magnitude of 1.0 and a standard deviation of 2 mm, while the second kernel, \( k_2 \), had a magnitude of 0.2 and a standard deviation of 40 mm. These values were chosen empirically, based on the penumbra and out-of-field regions when fitting the model to measured data for square fields. After determining \( I_{ij} \) for all segments \( j \), the integrated image was calculated according to:

\[ I_i = \sum_j I_{ij}. \]  

Couch attenuation was not included in this model. At the time of patient treatment, images of 1024 × 1024 pixels of 0.4 mm × 0.4 mm size (0.25 mm × 0.25 mm at isocentre) were acquired using an iViewGT portal imager (Elekta) in continuous movie mode at a frequency of 4 Hz. The set of images was summed using in-house software in an offline procedure to give an integrated image for the treatment, which was then compared with the forward-projection image using OmniPro GammaRay (IBA, Schwarzenbruck, Germany).

2.3. Back-projection EPID dosimetry

The iViewDose package (Elekta), based on the method of Wendling et al (2006, 2009) and Mans et al (2010a) was used for back-projection EPID dosimetry. In summary, the method corrected the pixel values of the \( j \)th acquired image (as opposed to control point in the previous section) for sensitivity to calculate the dose, \( d_{ij}^{\text{EPID}} \), at the imager plane, then calculated the primary portal dose, \( p_{ij}^{\text{EPID}} \), at the image plane by deconvolving the scatter from the EPID itself and correcting for penumbra over-sharpness:

\[ p_{ij}^{\text{EPID}} = d_{ij}^{\text{EPID}} \otimes -1 k_B^A \otimes k_b^B, \]

where \( k_A^A \) and \( k_B^B \) are convolution kernels, the former being similar in form to a delta function and the latter being Gaussian in form. Scatter from the patient was then subtracted, proportional to the integrated value of \( p_{ij}^{\text{EPID}} \) over the EPID. The primary dose in the patient at physical distance \( x \) from the source was then calculated according to:
\[ p_{ij}(x) = p_{ij}^{\text{EPIK}} \left( \frac{x}{x_{\text{EPIK}}} \right)^{-2} z_{ij}(x). \]  

(6)

The attenuation coefficient, \( z_{ij}(x) \), was calculated from the transmission of primary radiation, \( t_{ij}(x) \), between the depth \( x \) and the exit of the beam, and the total transmission, \( t_{ij}^{\text{TOT}} \), over the entire path of the beam through the patient:

\[ z_{ij}(x) = \frac{t_{ij}(x)}{t_{ij}^{\text{TOT}}}. \]  

(7)

both of these transmissions being calculated in the VMAT implementation from the equivalent path length \( r \) (Mans et al 2010a):

\[ t_{ij}(x) = \exp \left( -\mu r_{ij} + \sigma r_{ij}^2 \right), \]  

(8)

where \( \mu \) is the linear attenuation coefficient and \( \sigma \) is a correction for beam hardening. Finally, the total dose at depth \( x \) in the patient is calculated by adding the patient-specific scatter component to the primary dose:

\[ d_{ij}(x) = p_{ij}(x) + \left[ p_{ij}(x) \cdot \text{SPR} \right] \otimes k_{c}^{\text{C}}. \]  

(9)

in which SPR is the scatter to primary ratio as a function of patient thickness and \( k_{c}^{\text{C}} \) is a scatter kernel giving field-size dependence to the SPR. The process is repeated for all images, \( j \), acquired during the arc. A simple geometric couch model was included in this method. Practical implementation of this method has been discussed by Hanson et al (2014). Of particular note for the present study was an automatic shift of the measured images applied by the software so that the irradiated region aligned with a simple forward-projected image for the corresponding control point. This was intended to remove any panel shift or panel sag effects, but had implications for detection of errors in MLC leaf positioning.

Comparing these equations with those in section 2.2, it will be seen that the basic concept of both the forward- and back-projection methods is the projection of radiation between the patient and the image planes, with attenuation correction and inverse square correction, together with scatter corrections. Although the details of the methods vary, the core principles are very similar.

### 2.4. Clinical study

In this study, the same set of measured images as was used for comparison with the forward-predicted images, was also used for back-projection. At least one set of images was acquired for each patient in the series, with additional images on further treatment fractions being acquired in some cases. Acquisition of further images was either because the EPID dosimetry results for one fraction were unsatisfactory, or because images were required for other research studies. Although both of the forward- and back-projection methods described above were in routine clinical use, the back-projection method was used in this cohort of patients for making clinical decisions, as it was available in a bespoke commercial product. All patients consented to the use of their images for research.

Forward-projection images were compared using an intensity threshold of 10% and back-projection dose distributions were compared using a dose threshold of 50%. In both cases, gamma for 3% of global dose and 3 mm was calculated and evaluated. In the forward-projection method, global dose was taken to be a representative image intensity in the high-intensity part of the image, and in the back-projection method, it was defined as the prescribed dose. The percentage gamma less than unity and the mean gamma were evaluated. For the forward-projection method, the tolerances were based on a study of deliberately introduced errors (Bedford et al 2014), while the tolerances for the back-projection method were those recommended by the supplier, based largely on the experience of Mans et al (2010a, 2010b) and other work by the same group. In addition, the difference in isocentric signal was noted. For the forward-projection method, this was the difference in measured and predicted fluence on the central axis of the beam, and for the back-projection method, this was the difference in measured and calculated dose at the isocentre.

### 2.5. Phantom study

Each of the 13 treatment plans was also delivered to a Solid Water phantom (Radiation Measurements, Inc., Middleton, WI) of dimensions 300 mm wide by 300 mm long by 200 mm high, with the isocentre located at the centre of the phantom. Portal images were acquired and EPID dosimetry was carried out using both forward- and back-projection methods. The purpose of this step was to establish the intrinsic accuracy of the two techniques in the absence of any anatomical errors.
In three cases, deliberate errors were then introduced into the delivery of the treatment plan as follows:

1. An increase in monitor units at all control points of the arc, from 2% to 10% in 2% steps.
2. An increase in multileaf collimator opening on all leaves at all control points of the arc, from 2 mm to 10 mm in 2 mm steps.
3. A shift in multileaf collimator opening on all leaves at all control points of the arc, from 2 mm to 10 mm in 2 mm steps.
4. Replacement of a 50 mm slab of the phantom with two 300 mm long (superior–inferior direction) by 150 mm wide (lateral direction) by 50 mm high (anterior–posterior direction) slabs of polymethylmethacrylate, which were moved apart laterally from 10 mm to 50 mm in 10 mm steps, so as to create a medially located air gap. The replacement slab and air gap were situated 20 mm above the couch top, so that the uppermost (most anterior) edge was 30 mm below (posterior) to the isocentre (figure 3).

Both methods were then used to analyse the images, using the treatment plan and predicted images for the correct plan as the baseline.

3. Results

3.1. Clinical study

All plans meet the clinical dose constraints for the CHHiP trial. A typical treatment plan, together with the first few segments to give an indication of the complexity of the plan, is shown in figure 4. This figure also shows the dose-volume histograms in the absence of errors and with a selection of the simulated delivery errors used in the phantom study. Monitor unit errors make a predictable change to the dose delivered to the patient, while an increase in aperture size increases the delivered dose due to the increase in output factor, and also significantly increases rectal dose due to the larger irradiated volume. A shift in all apertures has a rather smaller impact on the treatment plan, due to the convolution of the shift with the rotational treatment, so that the error has a blurring effect on the dose distribution. Air in the rectal region has a similarly small impact on the delivered dose distribution. Note that these effects of the simulated errors on the delivered dose distribution are shown here for one patient only, and may be dependent on the exact nature of the errors and may also be different for other treatment sites.

Figure 5 shows the predicted integrated image, the corresponding measured integrated image and the resulting gamma distribution. The results of the back-projection approach are shown in figure 6 for the same patient and day of treatment as figure 5. In this case, the result is a 3D dose distribution based on the measured portal images, which can be compared with the planned dose distribution.

The results of the two different approaches are compared in figure 7 in terms of percentage gamma less than unity, mean gamma and the difference in isocentric fluence/dose between the two methods. In examining these results, a distinction is made between the equality of the two methods and the correlation, the former referring to the numerical agreement of the two methods, and the latter referring to the linearity of the relationship between the two methods. For example, one could conceive of a situation in figure 7 in which a set of measurements was exactly in a straight line (not necessarily at 45°) far from the line of equality, and another set of measurements scattered rather non-linearly around the line of equality. The former would have a high correlation coefficient, being on a straight line, but the latter would have higher equality. The Pearson product moment correlation coefficient, \( r \), is presented for completeness in table 2. The correlation coefficient is calculated for each patient separately, giving a measure of linearity within each patient, but is also calculated overall for the cohort of patients, this latter being based on the mean measurement for each patient (Altman and Bland 1983, Bland and Altman 1986, 1994, 1995a, 1995b, 1999). Figure 8 shows distributions of differences between the two different methods.
The current clinical tolerances for the comparison of measured and calculated results are also shown in figure 7. A two-stage evaluation is carried out, in which results lying within the tighter of the two tolerances are considered to pass, while results lying outside of the two tolerances are considered to fail, therefore requiring further evaluation, either by repeating the image acquisition on a further fraction of treatment, studying the cone-beam CT images, or carrying out phantom measurements. In between the two tolerance levels, results may be accepted after evaluation of the cone-beam CT images. All patients in this cohort pass the looser clinical tolerance, but some of the patients require multiple images before acceptance. This is discussed further below.

The outlying cases in figure 7 are investigated in figure 9. This shows the results of both EPID dosimetry methods, together with corresponding cone-beam CT images, for the two marked points in figure 7. For point A, forward-projection does not give good gamma agreement, but back-projection does. This effect is clearly not related to the prediction model as the gamma agreement for forward-projection is much better on other days of treatment. This result indicates that the particular positioning of the patient and the internal anatomy has a greater effect on the forward-projection method than on the back-projection method. For point B, a large pocket of gas in the rectum is present at treatment but not at planning. This has a larger effect on the back-projection method than on the forward-projection method, so that the percentage of the patient with gamma less than unity drops more rapidly for the back-projection method.

3.2. Phantom study
Figure 10 shows the percentage gamma less than unity, mean gamma and the difference in isocentric fluence/dose between the two methods for the treatment plans delivered to a phantom in the absence of errors. The methods do not give exactly the same results as each other, but the differences are less than in the patient cases, where there is anatomical variation also. In general, the forward-projection method has a larger spread of results, but the isocentric fluence difference is centred around zero (0.7% ± 2.0% (Mean ± 1 SD)), whereas the back-projection method produces a more consistent result, but with a systematic offset in isocentric dose (3.0% ± 1.0%), which is thought to be due to the use of a single beam model for multiple treatment units, as described by Hanson et al (2014). These plots represent the intrinsic accuracy of the portal dosimetry models used in the forward- and back-projection methods.
Figure 11 shows the percentage gamma less than unity and the difference in measured and predicted isocentric fluence/dose for the four types of introduced error. For monitor unit errors, the difference in measured and predicted isocentric fluence/dose behaves exactly in accord with the introduced error. This translates to a change in gamma less than unity, with the 3D gamma used in back-projection responding somewhat stronger than the 2D gamma used in the forward-projection case. For an error consisting of too large a field size, a similar result is seen, with the dual effects of increased field extent and increased fluence due to the secondary effect of increased output factor causing a rapid increase in isocentric difference and a rapid decrease in percentage of area/volume having gamma less than unity. An error in field positioning has a rather different effect. The isocentric fluence/dose is not grossly affected by this type of error, as expected. However, the percentage of area/volume with gamma less than unity is substantially reduced in the forward-projection method but not at all reduced in the back-projection method. This is a known limitation of the automatic image shift applied by the back-projection software, which has the consequence of removing the measured image shift. Referring to figure 4, this type of error has a small impact on the quality of the treatment plan, so is not critical. An aperture shift of 10 mm reduces the PTV coverage by 5–15%, which is clinically very significant, but the likelihood of such a large shift is very small. Finally, air gaps have a simple effect on the isocentric fluence/dose and a similar effect to a monitor unit change on the gamma result, with slightly more response in the back-projection method. This behaviour can be considered an over-response in both forward- and back-projection methods, as from figure 4, it can be seen that the impact of this type of error on the treatment plan is small. In summary, there is a simple relationship between the forward- and back-projection methods, with slightly greater response in the gamma for the back-projection method, except in the case of a shift, where the forward-projection method is the only method to respond at all.

Note that this behaviour can be traced with points A and B in figure 7. Point A is a shift effect, so only forward-projection responds, whereas point B is an air gap effect, so back-projection responds slightly more than forward-projection. Note also that for the high values of proportion of gamma less than unity typically encountered in the clinical cases, the results for both forward- and back-projection are similar, as the errors are not significant enough for the differences to become marked.

4. Discussion

These results show that both forward- and back-projection EPID dosimetry methods produce similar results for the patient cohort considered. Figure 7 shows that both methods produce results clustered around the
line of equality, so it is clear that both methods broadly agree. However, looking closer at the results, it can be seen that there is a very definite trend within each individual patient, but less of a trend in the overall cohort. The differences between patients can be understood by considering the uncertainties in the process. Both the forward- and back-projection methods use different calculation models, so it is expected that the two methods should give slightly different results for each patient. Moreover, the forward approach is based on the dose in the isocentric plane as calculated by AutoBeam, whereas the back-projection approach compares reconstructed dose with that calculated by Pinnacle3, so there is an additional source of uncertainty introduced by this difference. Dose differences between AutoBeam and Pinnacle are in the order of 1%, so differences in results in this study of the corresponding magnitude are expected. Finally, the forward- and back-projection methods are inherently different in their operation, with the forward-projection method being a comparison of 2D images with a 2D gamma method and associated percentage threshold, whereas the back-projection method is based on comparison of 3D dose distributions using a 3D gamma method. These observations are borne out by the results of the phantom study in the absence of errors, where the intrinsic accuracy of each projection model can be seen.

Within each patient, the above sources of error are approximately constant, leaving the variation in the images to provide differences in the metrics. These differences in images from day to day may be caused by differences in daily setup and overall changes in patient status with the course of treatment. The error study shows that monitor unit and aperture size errors have a large impact on the treatment plan and can be detected by both forward- and back-projection methods. Spatial shifts are simulated in the error study by an aperture shift, but may include a simple panel positioning error. This type of error is only detected by the forward-projection method, but only has a relatively small impact on the delivered dose distribution. Note, however, that a difference in patient position compared to the time of planning may also occur, and this type of error results in a shift of the dose distribution relative to the patient, which is likely to be of greater impact than the blurring of the dose distribution produced by an aperture shift. Pockets of rectal gas are detected by both forward- and back-projection methods, but these have a relatively low impact on the treatment plan and may therefore be considered as false positives. Spatial shifts and rectal filling are thought to be the main source of the intra-patient trends in figure 7. A further example of rectal filling is given in figure 12, where four gamma maps for four treatments in the same patient are shown, in which the rectal filling clearly influences the results. Variation in anatomy has been shown in other studies to influence the results of EPID dosimetry (Persoon et al 2015, Ricketts et al 2016, McCowan et al 2017).
The differences between the inter-patient variation of the results and the intra-patient variation are reflected in the Pearson correlation coefficient (table 2). Within each patient, the correlation between forward- and back-projection methods has a very high \( r \) value, leading to a relatively high mean \( r \) over the set of patients. This is particularly seen with the difference in isocentric dose/fluence between plan and measurement. Meanwhile, when the set of measurements is taken in aggregate, the \( r \) value is low. It is thought that the isocentric dose/fluence exhibits higher correlation than either of the gamma metrics because the isocentric dose and fluence on the central axis are coupled by a relatively simple relationship. On the contrary, the gamma distribution can change rapidly with dose and fluence, particularly if the dose difference reaches 3% or if there is a small spatial offset in planned and delivered dose.

In terms of simplicity and ease of implementation, the forward-projection approach is the most straightforward, because the calculation can be performed at the time of treatment planning and the acquired images do not require gantry angle to be recorded, the method being based instead on the sum of the acquired images. Compared to the forward-projection method, the benefit of the back-projection method is its ease of interpretation. Since the method produces a 3D dose distribution from the acquired images, comparison with the original planned dose distribution is possible in relation to the patient anatomy.

Recent research has shown that if gantry angle information is available, the sensitivity and specificity of the forward-projection method to errors can be improved using time-resolved images. Persoon et al (2015) have studied a large cohort of lung cancer patients, and find that most cases of atelectasis can be detected with integrated forward-projection, but other clinically relevant changes are not captured. No significant correlations are found between changes in dose-volume metrics and gamma fail rates. However, when using time-resolved planar EPID dosimetry results in terms of (a) percentage gamma (3% and 3 mm) less than unity, (b) mean gamma (3% and 3 mm) and (c) measured—predicted isocentric fluence/dose. The different colours represent different patients in the series. The dashed lines represent equality between the two methods and are only intended to guide the eye. The feint dotted lines represent the current clinical tolerances for the different methods. Points A and B are identified for further discussion.

Figure 7.
Table 2. Pearson product moment correlation coefficient ($r$) for the three comparison measures. The correlation coefficients of the individual measurements for each patient are shown, together with their mean, and the overall correlation of the cohort as a whole. Those patients for whom less than three measurements are available are marked as not applicable, and have been excluded from the mean $r$ but included in the overall $r$. Items marked with an asterisk ($*$) are statistically significant at the 0.05 level.

| Patient | Percent gamma < 1 | Mean gamma | Isocentre fluence/dose difference |
|---------|------------------|------------|----------------------------------|
| 1       | $-0.42$          | $-0.90*$   | $0.95*$                          |
| 2       | $0.68*$          | $0.85*$    | $0.98*$                          |
| 3       | $0.99*$          | $0.98*$    | $0.99*$                          |
| 4       | N/A              | N/A        | N/A                              |
| 5       | $0.90$           | $0.98*$    | $1.00*$                          |
| 6       | $0.99$           | $0.72$     | $1.00*$                          |
| 7       | N/A              | N/A        | N/A                              |
| 8       | $1.00$           | $1.00$     | $1.00$                           |
| 9       | $1.00$           | $1.00$     | $1.00$                           |
| 10      | $1.00*$          | $0.99$     | $0.99$                           |
| 11      | $-0.80$          | $0.19$     | $0.94$                           |
| 12      | $0.94$           | $0.97$     | $1.00*$                          |
| 13      | $1.00*$          | $0.98$     | $0.96$                           |

| Individual mean $R$ | 0.59 | 0.64 | 0.98 |
| Overall $R$         | 0.23 | 0.19 | 0.59*|

Figure 8. Histograms of number of images versus the difference between forward- and back-projection EPID dosimetry results, for (a) percentage gamma (3% and 3 mm) less than unity, (b) mean gamma (3% and 3 mm) and (c) measured—predicted isocentric dose. Each marker on the vertical axis represents one acquired image ($n = 46$).
Figure 9. Gamma maps for images A and B identified in figure 7. The results for forward-projection are shown on the left, results for back-projection on the appropriate planning CT are shown in the centre, and the corresponding cone-beam CT images are shown on the right. PTV1, bladder and rectum from the planning CT scan are shown overlaid on the cone-beam CT images.

Figure 10. Scatter plots comparing forward- and back-projection EPID dosimetry results in a phantom in the absence of anatomical errors, in terms of (a) percentage gamma less than unity, (b) mean gamma and (c) measured—predicted isocentric fluence/dose. The different colours represent the same plans as in figure 7.
Figure 11. Scatter plots comparing forward- and back-projection EPID dosimetry results in terms of (left) percentage gamma less than unity, and (right) measured—predicted isocentric fluence/dose, for phantom irradiations with (a) monitor unit errors, (b) aperture size errors, (c) aperture position errors and (d) air cavity errors. The different colours represent the same patients as figure 7. The dashed lines represent equality between the two methods and are only intended to guide the eye. The faint dotted lines represent the current clinical tolerances for the different methods.
imaging (Persoon et al 2016), it is possible to demonstrate improved sensitivity to anatomical changes. Schyns et al (2016) also find in a phantom context that time-resolved imaging is superior to time-integrated imaging, but they still find that it is difficult to interpret gamma fail rates in terms of changes in dose-volume metrics. The present study does not verify this directly, but as the back-projection method is time-resolved, the comparison undertaken here may allow some comparison. The back-projection method generally shows slightly higher sensitivity to errors than the forward-projection method in this study. Further work is clearly needed in this area.

With both methods, the projection is currently based on the planning CT rather than a treatment (cone-beam) CT, so that images and doses in the presence of rectal changes during treatment may not be perfectly reconstructed. As this study shows that most discrepancies are anatomy-related, EPID dosimetry based on the cone-beam CT is likely to improve the agreement between measurements and predictions for both the forward- and back-projection methods. This is also expected to be a promising area of future work.

5. Conclusion

Both forward-projection and back-projection EPID dosimetry methods have been compared for a cohort of prostate VMAT patients and found to give comparable results. Forward-projection using integrated images is relatively straightforward to implement, whereas back-projection has the advantage of providing a measured dose distribution for comparison with planned dose, and is therefore relatively easy to interpret. Both cases show similar numerical results in this study, indicating that both methods are equally valuable in ensuring the overall safety of clinical treatments. However, patient setup is not addressed by EPID dosimetry and other
complementary methods are therefore needed to identify possible sources of error such as spatial shifts or bowel gas. These other methods may include cone-beam CT or evaluation of the individual orientation-specific EPID images.

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