Original Research Article

Extending in aqua portal dosimetry with dose inhomogeneity conversion maps for accurate patient dose reconstruction in external beam radiotherapy

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Background and purpose: In aqua dosimetry with electronic portal imaging devices (EPIDs) allows for dosimetric treatment verification in external beam radiotherapy by comparing EPID-reconstructed dose distributions (EPID_IA) with dose distributions calculated with the treatment planning system in water-equivalent geometries. The main drawback of the method is the inability to estimate the dose delivered to the patient. In this study, an extension to the method is presented to allow for patient dose reconstruction in the presence of inhomogeneities.

Materials and methods: EPID_IA dose distributions were converted into patient dose distributions (EPID_IA_MC) by applying a 3D dose inhomogeneity conversion, defined as the ratio between patient and water-filled patient dose distributions computed using Monte Carlo calculations. EPID_IA_MC was evaluated against dose distributions calculated with a collapsed cone convolution superposition (CCCS) algorithm and with a GPU-based Monte Carlo dose calculation platform (GPUMCD) using non-transit EPID measurements of 25 plans. In vivo EPID measurements of 20 plans were also analyzed.

Results: In the evaluation of EPID_IA_MC, the average γ-mean values (2%/2mm, 50% isodose volume) were 0.70 ± 0.14 (1SD) against CCCS and 0.66 ± 0.10 (1SD) against GPUMCD, respectively. Percentage differences in median dose to the planning target volume were within 3.9% and 2.7%, respectively. The number of in vivo dosimetric alerts with EPID_IA_MC was comparable to EPID_IA.

Conclusions: EPID_IA_MC accommodates accurate patient dose reconstruction for treatment disease sites with significant tissue inhomogeneities within a simple EPID-based direct dose back-projection algorithm, and helps to improve the clinical interpretation of both pre-treatment and in vivo dosimetry results.

1. Introduction

Modern external beam radiation therapy demands dosimetric methods to perform patient-specific quality assurance (QA). Electronic portal imaging devices (EPIDs) show dosimetric characteristics [1–4] that have made them suitable for both pre-treatment and in vivo dosimetric verification [5–9]. Direct back-projection EPID dosimetry systems use the measured EPID signal to reconstruct dose within the patient model using dose-deposition kernels or other empirically based approaches [10,11]. Our simple direct back-projection algorithm allows for accurate dose reconstruction only in water-equivalent material. In such case, both transit and non-transit EPID dosimetry have proven to be equivalent in dosimetric terms to conventional detector arrays [12]. For dosimetric verification in the presence of inhomogeneities, the in aqua method has been developed [13]. In this approach, the algorithm uses the EPID signal to reconstruct dose in a patient-shaped water-equivalent geometry, the so-called water-filled patient. The treatment is verified by comparing EPID-reconstructed dose distributions with dose distributions calculated with the treatment planning system (TPS) in the water-filled patient geometry. The in aqua conversion was initially validated with inhomogeneous anthropomorphic phantom measurements in the original paper. The performance of the in aqua EPID dosimetry method

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against TPS dose calculations has been extensively evaluated for non-
transit EPID dosimetry [14,15] and for in vivo EPID dosimetry [16].
Despite the fact that in aqua EPID dosimetry has proven to be useful in
detecting variations in patient position and patient anatomy, there are
drawbacks inherent to this methodology. First, water-filled patient TPS
dose distributions need to be calculated and exported only for EPID
dosimetry purposes. This represents additional work for the planning
department which, depending on the TPS, may not be easy to streamline.
Furthermore, in aqua EPID-reconstructed dose distributions do not
estimate the actual dose delivered to the patient but the dose delivered
to a water-filled patient. Finally, detected deviations cannot easily be
related to clinically relevant comparison metrics such as patient dose-
volume histograms.

The purpose of this work was to extend in aqua EPID dosimetry by
incorporating 3D dose inhomogeneity conversion maps, accommoda-
ting accurate patient dose reconstruction for treatment disease sites with
significant tissue inhomogeneities within a single EPID-based direct
back-projection algorithm. Monte Carlo (MC) dose calculations [17,18]
were chosen for the determination of the conversion maps included in
this study due to the high accuracy of MC codes in inhomogeneous media [19,20]. The validity of the proposed method was evaluated by
comparing EPID-reconstructed dose distributions to reference dose dis-
tributions calculated with a collapsed cone convolution superposition
algorithm and with a GPU-based MC dose calculation platform. The perfor-
ance of the method for in vivo treatment verification was also
investigated.

2. Materials and methods

2.1. EPID measurements and equipment

EPID measurements were made on VersaHD linear accelerators
(Elekta, Crawley, UK) equipped with a PerkinElmer XRD 1642 AP
amorphous silicon EPID. EPID dose reconstructions were performed
with in-house developed clinical software. The EPID dosimetry software
reconstructed 3D patient dose distributions for intensity modulated ra-
diation therapy (IMRT) plans and for volumetric modulated arc therapy
(VMAT) plans [21]. For IMRT beams, the algorithm used the accumu-
lated image acquired between beam-on and beam-off to reconstruct the
dose distribution of the delivered IMRT beam. For VMAT arcs, the algo-
rithm used the sum of all frames acquired within a certain gantry-
angle range to reconstruct partial dose distributions, which were sum-
med together to obtain the dose distribution of the delivered VMAT arc.
EPID-reconstructed dose distributions of all the beams or arcs were
summed together to obtain the EPID-reconstructed dose distribution of
the delivered fraction. The grid size for EPID dose reconstructions was
2 mm. Treatment plans were generated with Pinnacle V9.16 (Philips
Medical Systems, Eindhoven, The Netherlands). This study was
approved by our institutional review board (IRBd20-336).

2.2. Reference dose calculations

Reference dose distributions were calculated with Pinnacle and with
a research version of the GPU-based MC dose calculation platform
(GPU-MCD) developed by Elekta for the Monaco TPS. Pinnacle employs a
collapsed cone convolution superposition (CCCS) algorithm [22]. Pa-

tient and water-filled patient dose distributions were calculated by
turning the inhomogeneity correction on and off in Pinnacle, respec-
tively. Mimicking clinical practice, the grid size for dose calculations
was 4 mm for prostate, breast, and lung, 3 mm for head-and-neck and 2
mm for lung stereotactic body radiotherapy (SBRT). GPU-MCD employs a
GPU-based MC dose calculation algorithm [23]. The MC beam models
and machine files required by the algorithm were provided by Elekta.
The grid size for dose calculation was 2 mm for all treatment disease
sites. For water-filled patient dose calculations, a density override of 1 g/

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2.4. EPID dosimetry without correction for inhomogeneities (EPID_NC)

The parameters of the back-projection algorithm were determined during the commissioning process using absolute dose measurements and EPID measurements made behind water-equivalent phantoms. Strictly speaking, if no corrections (NC) are applied for tissue inhomogeneities, the reconstructed dose distributions are accurate only for reconstructions in water-equivalent material and, in transit mode, only if the EPID measurements were made behind water-equivalent phantoms. EPID_NC is used in our clinic for pre-treatment and in vivo dosimetric verification of homogeneous treatment disease sites, such as prostate, liver or whole brain.

2.5. In aqua EPID dosimetry (EPID_IA)

The primary portal dose distribution $P_{EPID}$ was converted into the in aqua primary portal dose distribution $P_{EPID_IA}$, the equivalent primary portal dose distribution that would be measured behind a water-filled patient by:

$$P_{EPID_IA} = P_{EPID} \cdot IAC = P_{EPID} \cdot \frac{T_{CT_water}}{T_{CT_patient}}$$

The 2D in aqua conversion $IAC$ was defined as the ratio between two primary portal transmission images calculated from the water-patient and patient CT scan, respectively. For non-transit dosimetry, the in aqua primary portal dose distribution estimates the transit primary portal dose distribution that would be measured behind the water-filled patient in the absence of anatomical changes:

$$P_{EPID_IA, non-transit} = P_{EPID, non-transit} \cdot \frac{T_{CT_water}}{T_{CT_patient}}$$

For transit EPID dosimetry, the effect of the in aqua conversion is to ‘remove’ the effect that patient inhomogeneities have on the primary portal dose distribution.

$$P_{EPID_IA, transit} = P_{EPID, transit} \cdot \frac{T_{CT_water}}{T_{CT_patient}}$$

The in aqua conversion is accurate only if the CT scan perfectly accounts for the anatomy of the patient during the delivery. Otherwise, as often is the case with transit in vivo measurements, the in aqua conversion helps to determine a change in the delivery from the planned delivery. The algorithm used Eq. (6) for non-transit in aqua EPID dosimetry and Eq. (7) for transit in aqua EPID dosimetry. EPID_IA is used in our clinic for pre-treatment and in vivo dosimetric verification of treatment disease sites involving large tissue inhomogeneities, such as lung.

2.6. In aqua EPID dosimetry with Monte Carlo dose inhomogeneity conversion (EPID_IA_MC)

For each IMRT beam or VMAT arc, EPID_IA dose distributions $D_{EPID_IA}$ calculated in the water-filled patient were converted into patient dose distributions $D_{EPID_IA, MC}$ by applying a voxel-by-voxel 3D dose inhomogeneity conversion $DIC_{MC}$:

$$D_{EPID_IA, MC} = D_{EPID_IA} \cdot DIC_{MC} = D_{MC, patient}$$

The conversion was defined as the ratio between patient and water-filled patient MC dose distributions. $D_{EPID_IA, MC}$ was used for $DIC_{MC}$ calculations using a dose grid size of 2 mm [27,28]. The CT number to mass density conversion map was set equal to the one used in our TPS. For water-filled patient dose calculations, a density override of 1 g/cm$^3$ was applied to the external contour of the patient. The nominal dose computation uncertainty in SciMoCa can be selected from six options: Extra Fast (4%), Fast (2%), Fine (1%), Extra Fine...
(0.5%), Commissioning (0.25%) and Ultimate (0.1%). Such nominal uncertainty is defined as the average uncertainty for voxels with a dose larger than 70% of the maximum dose. In a preliminary investigation with three patient cases, an uncertainty of 0.5% showed the best compromise between accuracy and computation times. Calculations were performed in an Oracle Linux 7.6 server with an Intel Xeon E5-2697A v4 CPU (64 logical cores @ 2.60 GHz) and a 125 GB RAM. The average calculation time was $2.2 \pm 1.8$ (1SD) minutes. 6 MV (FF) and 10 MV (FF and FFF) photon beams were included in this study. MC beam models based on Elekta’s standard beam data for Versa HD linacs were used for dose calculations.

### 2.7. Alderson phantom measurements

An inhomogeneous anthropomorphic (Alderson) phantom was used to mimic in vivo patient dose reconstruction as much as possible. The Alderson phantom was irradiated using a double arc VMAT lung plan. Non-transit and transit EPID measurements were performed to reconstruct EPID_NC, EPID_IA and EPID_IA_MC dose distributions.

### 2.8. Evaluation of EPID_IA_MC dosimetry

Non-transit in air EPID measurements for 20 double arc VMAT plans (5 prostate, 5 head-and-neck, 5 lung and 5 lung SBRT) and for 5 IMRT breast plans were used to reconstruct EPID_NC, EPID_IA and EPID_IA_MC dose distributions. In air non-transit EPID dose reconstructions to the patient anatomy were used for method evaluation because they are exempt of patient related deviations.

### 2.9. In vivo treatment verification with EPID_IA_MC dosimetry

In vivo EPID measurements for plans of three treatment disease sites involving large inhomogeneities were evaluated: 18 VMAT lung fractions (9 plans), 16 VMAT lung SBRT fractions (8 plans) and 9 IMRT breast fractions (3 plans). IMRT dose reconstruction for breast plans was performed using the primary portal dose transmission calculated from EPID measurements as given in Eq. (4). In vivo dosimetry systems act as binary classifiers where plans are classified either as fail (alerted) or pass (non-alerted). The performance of EPID_IA and EPID_IA_MC regarding alert classification was compared by calculating the percentage of passing plans with varying alert threshold values for $\gamma$-pass rate results.
Table 1
Comparison between non-transit EPID-reconstructed and reference 3D dose distributions using 2L2 γ-analysis and ΔPTV_{D50}. EPID_IA distributions were compared to reference dose distributions calculated in the water-filled patient. 20 double arc VMAT plans (5 prostate, 5 head-and-neck, 5 lung and 5 lung SBRT) and 5 IMRT breast plans were analyzed. Results are presented as average ± (1SD). The last row presents the average results for all treatment disease sites together with the range displayed between parenthesis.

|                | EPID_NC | EPID_IA | EPID_IA_MC | ΔPTV_{D50} (%) | EPID_NC | EPID_IA | EPID_IA_MC |
|----------------|---------|---------|------------|----------------|---------|---------|------------|
| **vs CCCS**    |         |         |            |                |         |         |            |
| Prostate       | 0.53 ± 0.12  | 0.57 ± 0.04 | 0.53 ± 0.05 | 1.0 ± 0.9     | -0.5 ± 1.1 | -0.1 ± 0.9 |
| Head-and-neck  | 0.70 ± 0.08  | 0.71 ± 0.11 | 0.75 ± 0.10 | -0.8 ± 0.8    | -1.2 ± 0.8 | -1.5 ± 0.6 |
| Breast         | 1.48 ± 1.03  | 0.72 ± 0.11 | 0.78 ± 0.12 | 1.5 ± 2.7     | -0.6 ± 1.4 | -0.5 ± 1.5 |
| Lung           | 2.33 ± 0.95  | 0.71 ± 0.04 | 0.83 ± 0.07 | 6.7 ± 6.4     | 0.8 ± 1.0 | 0.6 ± 1.8 |
| Lung SBRT      | 4.17 ± 0.91  | 0.59 ± 0.04 | 0.61 ± 0.09 | 32.5 ± 9.5    | 1.3 ± 0.9 | -0.1 ± 3.0 |
| Total          | 0.66 ± 0.10 (0.52, 0.86) | 0.70 ± 0.14 (0.49, 0.93) |         | 0.0 ± 1.5 (2.2, 2.8) | -0.3 ± 1.8 (-3.6, 3.9) |
| **vs GPUMCD**  |         |         |            |                |         |         |            |
| Prostate       | 0.56 ± 0.10  | 0.54 ± 0.06 | 0.55 ± 0.05 | 0.9 ± 0.9     | -0.5 ± 0.9 | -0.1 ± 1.3 |
| Head-and-neck  | 0.70 ± 0.04  | 0.69 ± 0.05 | 0.72 ± 0.05 | 0.1 ± 0.9     | -0.7 ± 0.6 | -0.6 ± 0.6 |
| Breast         | 1.53 ± 0.96  | 0.72 ± 0.09 | 0.72 ± 0.08 | 2.3 ± 2.5     | 0.0 ± 0.9 | -0.1 ± 1.3 |
| Lung           | 1.88 ± 0.81  | 0.64 ± 0.10 | 0.72 ± 0.08 | 6.0 ± 5.7     | -0.2 ± 0.5 | -0.7 ± 0.6 |
| Lung SBRT      | 3.76 ± 1.30  | 0.60 ± 0.03 | 0.57 ± 0.03 | 29.0 ± 15.8   | 1.0 ± 1.1 | 0.6 ± 1.1 |
| Total          | 0.64 ± 0.09 (0.49, 0.82) | 0.66 ± 0.10 (0.49, 0.83) |         | 0.0 ± 1.1 (1.7, 2.8) | -0.3 ± 1.2 (2.7, 2.0) |

EPID_NC = EPID dosimetry with No Correction.
EPID_IA = In Aqua EPID dosimetry.
EPID_IA_MC = In Aqua EPID dosimetry with Monte Carlo based dose inhomogeneity correction maps.
CCCS = Collapse Cone Convolution Superposition algorithm.
GPUMCD = GPU-based Monte Carlo Dose calculation algorithm.

Fig. 3. (a) TPS-calculated dose distributions for five plans of different treatment disease sites, EPID-reconstructed left–right dose profiles through the center of the PTV compared to dose profiles calculated with (b) a collapsed cone convolution superposition algorithm (CCCS) and with (c) a GPU-based Monte Carlo dose calculation algorithm (GPUMCD). EPID dose reconstructions were performed with non-transit EPID measurements in three back-projection modes: without corrections (EPID_NC), in aqua (EPID_IA) and in aqua combined with Monte Carlo dose inhomogeneities conversion maps determined with ScMoCa dose calculations (EPID_IA_MC). CCCS_PATIENT and CCCS_WATER refer to dose calculations performed with CCCS for the patient and water-filled patient geometries, respectively. GPUMCD_PATIENT and GPUMCD_WATER refer to dose calculations performed with GPUMCD for the patient and water-filled patient geometries, respectively.
3. Results

3.1. Alderson phantom measurements

In the 2L2 γ-comparison against CCCS-calculated patient dose distributions, γ-mean values of 0.58 and 0.56 were found for transit and non-transit EPID_IA_MC dosimetry, respectively. ΔPTV\textsubscript{D50} results were within 1.5%. As can be seen in the dose profiles presented in Fig. 2, the transit and non-transit EPID_IA_MC dosimetry distributions were nearly identical and agreed well with the patient reference dose distribution. Compared with each other, the 2L2 γ-mean value was 0.3 and the ΔPTV\textsubscript{D50} value was 0.1%.

3.2. Evaluation of EPID_IA_MC dosimetry

In the comparison of EPID_IA with reference water-filled patient dose calculations, 2L2 γ-mean values were lower or equal to 0.86 and ΔPTV\textsubscript{D50} deviations were within 2.8% (see Table 1). In the comparison of EPID_IA_MC with reference patient dose calculations, 2L2 γ-mean values were lower or equal to 0.93 and ΔPTV\textsubscript{D50} deviations were within 3.9%. The EPID_IA_MC agreement was better against GPUMCD than against CCCS. The differences between EPID_IA_MC and EPID_IA results are explained by the extra discrepancies between the algorithm employed in the calculation of the DIC\textsubscript{MC} conversion (SciMoCa) and in the computation of the reference dose distribution (CCCS and GPUMCD). These differences were small for prostate and head-and-neck but they became larger for treatment disease sites with significant tissue inhomogeneities. For lung, the average increase in 2L2 γ-mean values was 0.12 ± 0.07(1SD) against CCCS and 0.08 ± 0.03(1SD) against GPUMCD. For lung SBRT, the average differences in ΔPTV\textsubscript{D50} results were −1.4% ± 3.1%(1SD) and −0.4% ± 1.0%(1SD), respectively. For illustration purposes, Fig. 3 exhibits results for arbitrarily selected plans of five treatment disease sites. For completeness EPID_NC results are also displayed.

3.3. In vivo treatment verification with EPID_IA_MC dosimetry

For the in vivo treatment verification cases, the EPID_IA_MC agreement was again better against GPUMCD than against CCCS (see Table 2). Differences between EPID_IA_MC and EPID_IA results were also observed. For all treatment disease sites together, the average decrease in 3G2 γ-pass rate values was 2.5% ± 2.5%(1SD) against CCCS and 0.8% ± 2.2%(1SD) against GPUMCD. These differences explain the somewhat lower percentage of passing plans with EPID_IA_MC than with EPID_IA, see Fig. 4.

4. Discussion

In this study, we have extended the in \textit{aqua} EPID dosimetry method by incorporating a MC-based dose inhomogeneity conversion map. This was illustrated with the examples of Fig. 3 where EPID_IA water-filled patient dose distributions were converted into EPID_IA_MC patient dose distributions that can be compared with reference patient dose calculations, therefore improving the clinical interpretation of results.

In this study, non-transit EPID_IA_MC dosimetry was evaluated against dose calculations performed with CCCS and GPUMCD algo-
rithms (see Table 1). The largest 2L2 γ-mean value was 0.93. This sug-
gests that, for the worst case, an average dose difference of ~ 2% is
expected in low-dose gradient regions or a distance-to-agreement of 2
mm in steep gradient region which is considered adequate for the pur-
pose of EPID-based pre-treatment verification. A similar evaluation of
transit EPID_IA_MC dosimetry with EPID measurements made behind
anthropomorphic phantoms was deemed unnecessary for this study. The
reason is that transit and non-transit EPID_IA dosimetry produce nearly
identical results in such case. The underlying principle of non-transit
EPID dosimetry is that \( \Phi_{\text{EPID}}^{\text{virtual}} \) correctly estimates \( \Phi_{\text{EPID}}^{\text{virtual}} \) in the absence of anatomical changes. Note how, in that situation, Eq. (7)
does not apply to EPID_IA_MC would lead to a decrease in the number of passing plans compared to EPID_IA. Fig. 4
exhibited how the extent of the decrease depends on the algorithms
employed for the calculation of the DIC.\({ }^{\text{MC}}\) conversion and for the
reference dose distribution. The more similar these algorithms are to
each other, the more closely the EPID_IA and EPID_IA_MC verification results are related.

In indirect EPID back-projection methods, the EPID signal is back-
projected through the patient model to determine the incident fluence
to the patient. This fluence, together with a patient model, is fed into a
conventional forward dose engine to calculate the contribution of each
EPID image (frame) to the total patient dose distribution. Conventional forward dose engines are used
to back-project the transit and in-vivo dosimetry in radiotherapy using empirically
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