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Virtual patient 3D dose reconstruction using in air EPID measurements and a back-projection algorithm for IMRT and VMAT treatments

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ABSTRACT

Purpose: At our institute, a transit back-projection algorithm is used clinically to reconstruct *in vivo* patient and in phantom 3D dose distributions using EPID measurements behind a patient or a polystyrene slab phantom, respectively. In this study, an extension to this algorithm is presented whereby in air EPID measurements are used in combination with CT data to reconstruct 'virtual' 3D dose distributions. By combining virtual and *in vivo* patient verification data for the same treatment, patient-related errors can be separated from machine, planning and model errors.

Methods and materials: The virtual back-projection algorithm is described and verified against the transit algorithm with measurements made behind a slab phantom, against dose measurements made with an ionization chamber and with the OCTAVIUS 4D system, as well as against TPS patient data. Virtual and *in vivo* patient dose verification results are also compared.

Results: Virtual dose reconstructions agree within 1% with ionization chamber measurements. The average γ-pass rate values (3% global dose/3 mm) in the 3D dose comparison with the OCTAVIUS 4D system and the TPS patient data are 98.5 ± 1.9%(1SD) and 97.1 ± 2.9%(1SD), respectively. For virtual patient dose reconstructions, the differences with the TPS in median dose to the PTV remain within 4%.

Conclusions: Virtual patient dose reconstruction makes pre-treatment verification based on deviations of DVH parameters feasible and eliminates the need for phantom positioning and re-planning. Virtual patient dose reconstructions have additional value in the inspection of *in vivo* deviations, particularly in situations where CBCT data is not available (or not conclusive).

1. Introduction

The introduction of intensity modulated RT (IMRT) and volumetric modulated arc therapy (VMAT) was accompanied by additional verification of individual patient treatments, as recommended by several organizations [1]. These patient specific checks are generally performed pre-treatment by using phantom-detector array combinations, film or portal dosimetry. Although pre-treatment verification using in phantom dosimetry helps in determining whether the treatment can be delivered as expected to a stable anatomy, its usefulness is known to be limited in detecting clinically relevant errors in the delivery of the treatment to the patient [2,3]. Furthermore, several studies have shown that γ-pass rate has insufficient predictive power for pre-treatment dose verification [4–6] and have indicated the necessity to assess the delivered dose to a patient using other metrics such as dose-volume histograms (DVHs) of the target volume and organs at risk. Alert criteria based on deviations in DVH parameters have in addition a clearer relation with treatment planning data, while being easier to interpret by most radiotherapy staff than γ-evaluation data.

At the Netherlands Cancer Institute (NKI), pre-treatment verification of clinical plans is performed using in phantom EPID dosimetry with a slab polystyrene phantom [7]. Pre-treatment verification based on deviations of DVH parameters related to the patient anatomy is, by definition, not feasible with in phantom dosimetry. A second disadvantage of our in phantom EPID-based dosimetry method, which also applies to other pre-treatment dose verification methods, is that it requires additional clinical time due to the need for phantom re-planning and positioning. The main purpose of this study is to use in air EPID measurements to reconstruct dose distributions within the patient anatomy. This will eliminate the need for phantom positioning and re-planning and will make pre-treatment verification based on deviations of DVH parameters feasible.

Non-transit EPID dosimetry has been in use for some time in IMRT pre-treatment verification [8–14]. For patient dose reconstruction in 3D, in air EPID measurements are generally used to derive the actual fluence delivered by the accelerator, which may

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then be used as input to a dose engine that generates the patient 3D dose distribution. van Zijtveld et al. [15] used their clinical treatment planning system (TPS) for the dose engine of their reconstruction method and the system was clinically evaluated for 17 IMRT treatments of different sites. The use of the same TPS for dose reconstruction and patient treatment planning may hide, however, potential inaccuracies in the TPS dose calculation algorithm. Aiming at a TPS-independent dose calculation, van Elmpt et al. [16] used an in-house developed Monte Carlo dose engine and their model was used in the pre-treatment verification of 9 lung cancer patients treated with a 3D conformal technique and 5 head-and-neck cancer patients treated with a step-and-shoot 7-field IMRT technique. A similar approach is followed by a commercially available system, Dosimetry Check (Math Resolutions, Columbia, MD, US), where the fluence is entered into an independent pencil beam type dose calculation algorithm [17], [18]. Recently, a GPU-accelerated collapsed cone convolution technique was explored for dose reconstruction in phantom or in CT simulation data sets using in air EPID measurements by Zhu et al. [19]. Results for the verification of a head-and-neck, two lung and one prostate IMRT plan showed good agreement with TPS calculated dose distributions using 3%/3 mm criteria.

In this study, a novel approach is taken. Rather than estimating the energy fluence delivered by the accelerator and then using it as input to a dose engine for a forward dose calculation, the use of an extension of our transit back-projection algorithm was investigated. This new extension to the algorithm allowed us to use in air EPID measurements in combination with planning CT data to calculate patient 3D dose distributions, i.e., virtual patient dose distributions. These virtual patient dose distributions can be used for both IMRT and VMAT pre-treatment verification.

A unique advantage of using the same algorithm for virtual and *in vivo* reconstruction would be that patient-related errors may be separated from machine, planning and model errors by combining virtual and *in vivo* patient verification data for the same treatment. This circumvents a common limitation of *in vivo* dose verification methods which is the inability to discriminate changes in the measured *in vivo* dose distribution due to the variation in the fluence incident on the patient from changes due to anatomical variations within the patient. In this study, we illustrate the usefulness of this approach with some clinical examples. It should be noted that, in principle, the methods mentioned earlier [16,17,19] are also capable of combining pre-treatment with *in vivo* 3D dose verification results for the same patient. However, to the best of our knowledge these groups have not yet published results of those types of studies.

Finally, this is one of the first published studies[1] presenting patient-specific pre-treatment VMAT verification results in 3D using in air EPID measurements.

Another practical advantage of this approach would be that no extra commissioning work is required prior to the introduction of patient virtual dose reconstruction in our clinic.

In this study, we present the modifications made to our algorithm to allow for virtual dose reconstruction and the results of the assessment of its accuracy.

2. Materials and methods

2.1. Virtual patient dose reconstruction algorithm

Our back-projection algorithm requires the primary portal dose distribution, i.e. the dose component at the EPID level which results from radiation coming directly from the radiation head of the accelerator.

For *in vivo* patient dose reconstruction, the transit algorithm uses *in vivo* EPID measurements to determine the primary portal dose distribution behind the patient. For in phantom dose reconstruction, the transit algorithm uses EPID images acquired behind the phantom to determine the primary portal dose distribution behind the phantom.

For virtual patient dose reconstruction in this study, the new virtual algorithm uses in air EPID measurements in combination with the CT data of the patient to predict the primary portal dose distribution behind the patient. In a similar way, the new virtual algorithm can use in air EPID measurements in combination with the CT data of the phantom to predict the primary portal dose distribution behind the phantom. Fig. 1 shows schematically these four situations.

The algorithm then uses this primary portal dose distribution to reconstruct the dose distribution in any plane parallel to the EPID [20,21]. By iterating this 2D reconstruction in multiple planes, the dose distribution in 3D can be obtained.

In the following, we use the notation X_{ij} for a quantity X at pixel *ij* of the EPID. The line from the target of the linac to this pixel *ij* constitutes the geometrical back-projection line *ij*. Note that the coordinate system for dose reconstruction is fixed to the gantry.

In our algorithm, the *in vivo* primary portal dose distribution is calculated from the pixel values $PV_{ij}^{EPID,patient}$ of EPID images measured behind the patient by correcting for the sensitivity matrix S_{ij}, applying the EPID dose response D_r, removing the scatter within the EPID by deconvolving with kernel K_{ij}^{EPID1} and blurring the EPID signal with convolution kernel K_{ij}^{EPID2}, correcting for the couch attenuation C_{ij} and removing the scatter from the patient to the EPID $SC_{ij}^{patient,EPID}$:

$$Pr_{ij}^{EPID,in\ vivo} = ((PV_{ij}^{EPID,patient} \cdot S_{ij} \cdot D_r) \otimes^{-1} K_{ij}^{EPID1} \otimes K_{ij}^{EPID2}) \cdot C_{ij} \cdot SC_{ij}^{patient,EPID} \quad (1)$$

If the treatment is delivered without the patient in the beam, the in air primary portal dose distribution can be calculated in a similar way, but without the corrections for the couch attenuation and the patient scatter:

$$Pr_{ij}^{EPID,in\ air} = ((PV_{ij}^{EPID,in\ air} \cdot S_{ij} \cdot D_r) \otimes^{-1} K_{ij}^{EPID1} \otimes K_{ij}^{EPID2}) \quad (2)$$

The primary transmission at the EPID level $T_{ij}^{primary,CT}$ can be calculated using the radiological thickness of the patient t_{ij}^{radiol}, the linear attenuation coefficient of water for a specific beam energy μ and the beam hardening coefficient σ [22]. The radiological thickness is determined by ray-tracing through the patient's planning CT scan, i.e., by calculating a Digitally Reconstructed Radiograph (DRR). The numerical values for μ and σ are determined during the commissioning process of the portal dosimetry system fitting the equation

$$T_{ij}^{primary,CT} = e^{\left(-\mu \cdot t_{ij}^{radiol} + \sigma \cdot (t_{ij}^{radiol})^2\right)} \quad (3)$$

against the ratio of Eqs. (1) and (2) for a set of measurements behind phantoms of different thicknesses.

In this study, the algorithm was adapted to use $Pr_{ij}^{EPID,inair}$ and $T_{ij}^{primary,CT}$ to estimate the *virtual* primary portal dose distribution by combining Eqs. (2) and (3):

$$Pr_{ij}^{EPID,virtual} = Pr_{ij}^{EPID,in\ air} \cdot T_{ij}^{primary,CT} \quad (4)$$

The rest of the algorithm remains the same regardless of whether Eqs. (1) or (4) is used to determine the primary portal dose distribution. In air EPID measurements contain information

[1] In a recent paper by McCowan et al. (Phys. Med. Biol. 62: 1600–1612, 2017), which was published after submission of our paper, a similar study has been described.