3D dose reconstruction in external beam radiotherapy using portal imaging

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Abstract – The verification of the dose delivered to a patient during radiation therapy is a particularly important step for ensuring treatment quality. This step, known as in vivo dosimetry can be realized using a bidimensional dosimeter named Electronic Portal Imaging Device (EPID). However, few methods enable the determination of an accurate and fully personalized 3D dose distribution for the most current state of the art irradiation techniques. In this work, we propose a 3D dose reconstruction method based on Monte Carlo simulations allowing an accurate and fully personalized dose estimation from EPID device.

Index terms - Radiotherapy, Signal Processing

I. INTRODUCTION

Several methods of EPID-based 3D dose reconstruction are described in the literature. Some of them are based on the back projection through the voxelized patient of the primary contribution of the dose, extracted from the 2D EPID images. The dose deposited in each voxel may then be calculated by convolving the primary signal entering the voxel with a dose deposition kernel [1] or using a MC dose calculation algorithm [2]. However, such back projection methods do not take into account all the physical and geometrical distortions. In a more recently proposed 3D dose reconstruction approach [3], the cumulative dose in the EPID and in a patient slice are related by a linear system. As the dose signal acquired by the EPID is known for an irradiation, the resolution of the linear system simply relies on patient and EPID impulse response function (IRF) estimations. This resolution leads to a 3D dose distribution by repeating this resolution step for each patient slice and for each beam angle. IRFs are estimated using Monte Carlo (MC) simulations; all physical and geometrical effects are then taken into account. Nevertheless, considering realistic clinical applications and current irradiation techniques such as Volumetric Modulated Arc Therapy (VMAT), IRF method is not suitable for clinical practice. Indeed, considering the resolution needs and the large number of beam positions, estimating and solving such a linear system will be computationally demanding. In addition, the number of the necessary IRFs will be too high to be handled even if a computing cluster is used.

Relying on Yeo's work [3], we propose to improve and adapt the IRF method, aiming at its application in a clinical context. Preliminary results on simulated data validate the proposed approach.

II. MATERIALS AND METHODS

II.1. IRF method

The linear accelerator beam is divided into N rectangular impulsional sources ("beamlets"). Supposing that the dose delivery system is linear, the cumulated dose delivered to the i^{th} voxel of a patient slice $P(i)$ can be expressed as a linear combination of the dose emitted from the set of N beamlets $(S(k))_{1\leq k\leq N}$:

$$P(i) = \sum_{j=1}^{N} R_P(i,j)\, S(j)$$

For similar reasons, the cumulated dose received by the i^{th} pixel of the EPID $E(i)$ can also be expressed as:

$$E(i) = \sum_{j=1}^{N} R_E(i,j)\, S(j)$$

Considering a portal region of interest composed of m pixels and a portion of a patient slice containing n voxels, work assumptions finally become:

$$\underline{P} = \boldsymbol{R_P}\underline{S} \quad (1) \qquad \text{With,} \quad \underline{P} = [P(1)\, P(2) \ldots P(n)]^T$$
$$\underline{E} = \boldsymbol{R_E}\underline{S} \quad (2) \qquad \underline{E} = [E(1)\, E(2) \ldots E(m)]^T$$

From equations (1) and (2), a relation between the distribution of the cumulated dose deposited in a patient slice $\underline{P}$ and the distribution of the cumulated dose deposited in the EPID $\underline{E}$ may be deduced:

$$\underline{P} = \boldsymbol{R_P}\boldsymbol{R_E}^{+}\,\underline{E} \qquad (3)$$

With $\boldsymbol{R_E}^{+}$ the pseudo-inverse of the matrix $\boldsymbol{R_E}$.

$\boldsymbol{R_P}$ and $\boldsymbol{R_E}$ are the global response matrices of the patient slice and of the EPID to a set of impulsions. These matrices are also known as "influence matrices" or "kernel matrices". Each column represents a response of a

patient slice/of the EPID to a beamlet. To perform a highly accurate dose reconstruction, these global response matrices are estimated using MC simulations.

II.3. **Sparse IRF method**

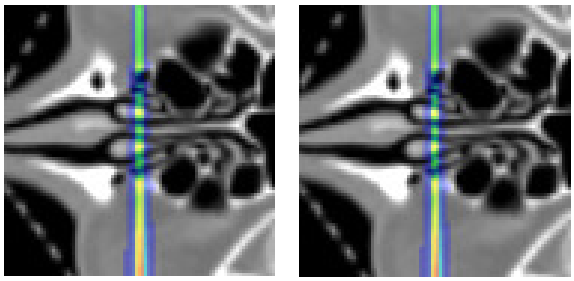
Considering a large number of beamlets, the main drawback of the IRF method is $\boldsymbol{R_P}$ and $\boldsymbol{R_E}$ sizes**.**The second issue is the time required to estimate such a linear system using Monte Carlo simulations (MCS). Here, we deal with the first issue related to the size of the global response matrices. Because the beamlet is very thin, the associated dose is deposed along a line through the patient. Consequently, the matrices $\boldsymbol{R_P}$ and $\boldsymbol{R_E}$ are large but sparse. This sparsity is exploited to reduce the data storage space of the IRF method. First, a threshold is defined to extract, from the background, the dose delivered by the beamlet. The MCS dose calculation uncertainty is used to choose this threshold. Subsequently, this value is used to compress the data using a sparse format. A Column Compressed Storage (CCS) format is used for its efficiency. Finally, $\boldsymbol{R_P}$ and $\boldsymbol{R_E}$ are built for each angle beam specified within the treatment planning. These data are precalculated before the treatment. Each EPID acquisition will be used to reconstruct the patient dose on-the-fly thanks to corresponding pair of matrices $\boldsymbol{R_P}$ and $\boldsymbol{R_E}$. The pseudo-inverse matrix vector multiplication ($\boldsymbol{R_E}^+ \underline{E}$) required in equation (3) is solved using a conjugate gradient algorithm developed in the sparse domain (CCS). In addition, to assure the method application in a clinical context, reconstruction is performed using Graphics Processing Units (GPU) and the cuSparse library.

II.2. **Evaluation study**

Estimation of global response matrices $\boldsymbol{R_P}$ and $\boldsymbol{R_E}$ is performed for a TrueBeam Novalis linear accelerator, an EPID aS1000 (1024×768 pixels) and a head and neck patient CT image ($512\times512\times294$ voxels, resolution: $1.26\times1.26\times2\ mm^3$) using GATE Monte Carlo simulation platform [4]. The beamlet to isocenter distance was set to 49cm. A number of 31×31 beamlets with a size of $2\ mm^2$ at the isocenter was chosen for the simulation. Dose depositions were recorded within the phantom for a region-of-interest (ROI) around the isocenter ($70\times70\times70$ voxels of $1\ mm^3$). Similarly, EPID images were recorded for a ROI of 255×255 pixels with a size of $0.392\times0.392\ mm^2$.

III. **RESULTS**

The dose distribution within the patient was reconstructed from the simulated EPID image of the central beamlet using a GPU GTX590 (computing time: 3s). In the considered compressed domain, $\boldsymbol{R_P}$ and $\boldsymbol{R_E}$ sizes were 144 MB and 75 MB while $\boldsymbol{R_P}$ and $\boldsymbol{R_E}$ sizes reached 921 MB and 175 MB in classical data storage format.



(a) (b)

Figure 1: Dose image of one beamlet crossing the nose cavity obtained by (a) Monte Carlo simulation and (b) the EPID reconstruction.

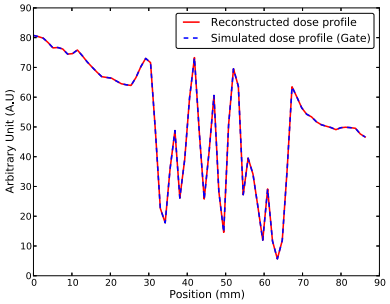


Figure 2: Comparison of the simulated and reconstructed dose profiles

Simulated and reconstructed dose distributions were compared in Fig. 1 and Fig. 2 using one slice of the phantom. Their profiles were also compared. Results show a perfect agreement between simulated and reconstructed dose.

IV. **CONCLUSION**

We proposed a sparse version of the IRF method which enables an accurate dose reconstruction. In addition, the reconstruction time is compatible with a clinical context. Further works will be planned to reduce the time to estimate the global matrices by using GPU for the MCS. Clinical studies will also be investigated considering different pseudo-continuous irradiation technique such as VMAT.

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