Dynamic 3D measurement of modulated radiotherapy: a scintillator-based approach

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Abstract. With the rise of high-conformity dynamic radiotherapy, such as volumetric modulated arc therapy and robotic radiosurgery, the temporal dimension of dose measurement is becoming increasingly important. It must be possible to tell both ‘where’ and ‘when’ a discrepancy occurs between the plan and its delivery. A 3D scintillation-based dosimetry system could be ideal for such a thorough, end-to-end verification; however, the challenge lies in retrieving the volumetric information of the light-emitting volume. This paper discusses the motivation, from an optics point of view, of using the images acquired with a plenoptic camera, or light field imager, of an irradiated plastic scintillator volume to reconstruct the delivered 3D dose distribution. Current work focuses on the optimization of the optical design as well as the data processing that is involved in the ongoing development of a clinically viable, second generation dosimetry system.

1. Introduction
Three-dimensional mapping of photon-based radiation therapy (RT) dose delivery is increasingly relevant in the era of small tumor targets and high conformity treatments [1]. To assure treatment quality, image guidance can be used to verify that a patient is in the right position and that his or her anatomy has not undergone important changes. However, patient imaging is only a subset of the whole treatment quality monitoring. High conformity RT delivered dynamically can often be strenuous for the treatment machine. It is therefore important to have instruments capable of verifying that a linear accelerator can deliver treatments exactly as planned. Detectors used for this purpose usually share similar properties: a large number of sampling points to map rapidly planar or volumetric doses, a good spatial resolution to track high dose gradients, and a compatibility with some type of phantom to measure dose as it would be received by the patient. An ideal detector would be capable of testing the whole treatment chain from end to end (i.e., from planning to delivery) with high accuracy and high precision. Moreover, the temporal capability of such a detector would enable an identification of the source of any potential deviations. Dosimetric gels have been used with success for this end-to-end task, partly because they are one of the only dosimeters offering complete 3D dose mapping [2]. However, gels can be labor-intensive and they do not yet offer dynamic measurements. Scintillation-based 3D dosimetry has so far shown promising results. Scintillators are water-equivalent and emit light spontaneously (within a few tens of nanoseconds). Feasibility of 3D tomographic reconstruction of scintillation measurements has been demonstrated [3], but it initially required a long acquisition time and a rotating setup. Liquid scintillator-based systems for proton therapy yielded promising results [4], but they exploit the unique physics of proton dose deposition, which cannot be used with
photon-based treatments. To image a dynamic treatment, it is preferable to have a static acquisition system to avoid interplay effects. Acquiring a large number of projections required for tomographic reconstruction without rotation is not trivial, but it is possible [5]. The conception of an ideal acquisition system is challenging at best. There is an intrinsic tug-of-war between optimizing the system’s spatial resolution, its temporal resolution and its light collection efficiency. Furthermore, the optimal tomographic reconstruction strategy greatly depends on the number of projections and their angles, which are determined by the system’s optical design.

While relatively new in medical physics, reconstructing 3D light-emitting objects is actively pursued in combustion research. Several methods for the reconstruction of light patterns from flames and chemiluminescence processes have been proposed [6, 7]. There are several similarities between the reconstruction of a flame and the reconstruction of the light field emitted by a scintillator. Thus, it is possible to use advances in flame/chemoluminescence reconstruction to help optimize a new 3D dosimeter. However, to conceive a system that is clinically viable, both the optical design and data processing (i.e., reconstruction algorithms) must be carefully planned.

2. A 3D scintillator-based prototype using a single plenoptic camera

Over the last decade, plenoptic cameras, or light field imagers, have been brought forth as a solution for “3D imaging” of solid objects, namely for digital refocussing, depth measurements and post-acquisition point-of-view adjustment. In essence, the plenoptic camera is simply a conventional digital camera to which a microlens array is added in front of its active sensor (CCD or CMOS). This additional optical component allows a measurement of an imaged light-field, i.e., the 4D function describing the angular distribution of light incident on each pixel of the camera’s detector. The microlenses subdivide the rays incident on the camera’s main lens, thus generating an array of micro-images on the sensor. Consequently, the signal detected on a single pixel results from the integration of rays incident from within a finite range of angles, which allows for a partial recovery of the angular information lost in the case of a conventional camera. The amount of angular information recovered depends mainly on the placement of the microlens array with respect to the main lens. When coincident with the main lens’ focal plane, maximum angular resolution can be attained at the cost of low spatial resolution (plenoptic 1.0). A spatio-angular trade-off is possible when the microlens array is offset from this plane, thus relaying the image formed by the main lens onto the sensor (plenoptic 2.0). A further alteration of the plenoptic 2.0, or focused plenoptic camera, is that of interleaving microlenses of different focal lengths within the microlens array, resulting in what is known as the multifocus plenoptic camera [8, 9]. This type of plenoptic camera serves to increase the usable depth-of-field (DOF), meaning that at least one microlens type is focused for any given imaged point in object space. Usually, increasing the DOF of a camera means reducing the size of the aperture stop of the system, thus involving systems operating at higher f-numbers (f/#). However, the build of the multifocus plenoptic camera allows the system to be used at higher speeds, i.e. lower f/#’s, thus increasing the light collection efficiency of the system and the spatial resolution of the system [8].

In the current 3D scintillation-based dosimetry system, the directionally-weighted information of the scintillation light field acquired by a multifocus plenoptic camera forms the basic input to an iterative tomographic reconstruction algorithm: based on paraxial optics, light rays are propagated from the sensor pixels to the voxelized plastic scintillator volume, first intersecting the microlens array then the main lens. In its current proof-of-concept state [10], the prototype, shown in Fig. 1., uses a plastic scintillator volume of 10×10×10 cm³ (490 nm emission peak) both as the radiosensitive and phantom material. The cubic scintillator is embedded within cylindrical acrylic plates, allowing for a better angular uniformity during dynamic irradiations (Fig. 1c). The scintillator is imaged with a static, 3-focal length polychromatic multifocus plenoptic camera, used with a 50-mm main lens operated at a f/# of 2.8. The current acquisition rate is of 1 Hz, set as a compromise between the light collection efficiency and temporal resolution of the system. Sampled planes from an absolute 3D dose reconstruction of a lung SBRT treatment are shown in Fig. 2.
Towards a second generation prototype: optimizing the system’s optics

Improving the dose reconstructions relies on a better knowledge and subsequent optimization of all optical parameters involved in the ray tracing, namely the diameter and focal lengths of the plenoptic camera’s main lens and microlenses, as well as the spacing between each component. These parameters influence not only the accuracy of the ray tracing based on the recovered angular information, but they affect the amount of vignetting present in separate micro-images, which limits the number of pixels usable for the reconstruction. Since these optical parameters are difficult to alter and study in the lab, simulations were performed using Zemax (OptiStudio™ 15, Zemax, LLC), a ray tracing and optical design software. Baseline parameters of the current system were first determined by studying the plenoptic camera with an optical microscope and an optical test instrument (OptiSpheric®, TRIOPTICS). These measured values were used as input parameters to reproduce the optical system in Zemax. Simulated images of targets of varying spatial resolution at different camera-to-target distances were compared with plenoptic images acquired in the lab. Using a sequential ray tracing mode with paraxial lenses, and as shown in Fig. 3, it was possible to model appropriately the change in spatial resolution as a function of depth without any lens vignetting effects or noise introduced by the optical system. These simulations form the basis needed for studying the effect of varying different plenoptic parameters.

It must be highlighted that the limit to using the plenoptic camera as a solution for 3D imaging is the lack of spatial resolution along the camera’s optical axis. From a tomography point of view, it is as if the plenoptic camera acquisition provides a partial sinogram, which needs to be compensated for using some perpendicular acquisition of the radiation beam’s eye view (BEV). Images acquired with the electronic portal imaging device (EPID) during dynamic deliveries were first used to provide a constraint on the depth to which the rays were backprojected into the scintillating volume [10]. This solution was initially motivated by the ease-of-use and availability of the EPID, and had the potential to be better exploited in a future prototype. However, in optimizing a 3D dosimetric system for dynamic modulated radiotherapy, the ideal is to provide a fully independent system, making the EPID-
based solution imperfect in this perspective. In continuing the optical simulations and optimization of the plenoptic system, it will be important to understand fully the limit of using a plenoptic camera for 3D imaging of translucent media. Knowing this limit will allow for a better consideration of other optical-based solutions aimed at compensating the plenoptic camera’s lack of spatial depth resolution.

3. Data processing involved in reconstructing 3D dose distributions

The use of a plenoptic camera to retrieve the 3D light pattern emitted from within a plastic scintillator volume calls for reconstruction algorithms intimately related to those used in single-photon emission computed tomography (SPECT). Various reconstruction methods could have been used, namely analytical algorithms (e.g., filtered backprojection) or iterative algorithms, of which the most popular are algebraic reconstruction techniques (ARTs), maximum-likelihood (ML) and expectation maximization (EM). For the plenoptic image-based reconstruction, an iterative algorithm was selected considering that analytical algorithms, albeit quicker, rely on a large number of projections to attain adequate precision. As for the different iterative algorithms, ARTs were discarded due to the fact that they do not, in their basic form, account for statistical error related to projection measurements. A ML-EM algorithm was retained for the task as it allowed for noise and statistical error modeling as well as incorporation of known physical constraints [11, 12]. The algorithm was implemented in Python because of the language’s versatility and its ease of development. However, because of relatively long reconstruction times, other alternatives will need to be explored such as porting the most time-consuming subroutines to a low-level programming language (e.g., C/C++) or to more suitable processing architecture such as graphical processing unit (GPU) using a library such as PyCUDA.

The reconstruction can be divided into four main blocks. (1) All plenoptic images acquired of the scintillating volume during a dynamic beam delivery are imported and filtered. (2) The second block of the algorithm calculates the system projection matrix used in the tomographic reconstruction algorithm. Depending on the scintillator volume, the chosen voxel resolution and the number of pixels sampled from the sensor, the sparse matrix representing the directionally-weighted propagation of rays through the plenoptic system can easily saturate the computer’s memory. To prevent memory saturation and to accelerate computation of the matrix, this step is divided into subfunctions of smaller extent, which can then easily be run in parallel using a default multiprocessing package. (3) A BEV constraint is produced from 2D EPID images acquired in cine mode during radiation delivery. For each plenoptic acquisition, a corresponding 3D matrix is generated by scaling the 2D EPID image along the beam’s axis. This matrix serves to optimize the shape of the reconstructed 3D dose slice-by-slice based on a Gaussian convolution. (4) Finally, the last block of code takes the outputs of the preceding blocks and iteratively reconstructs the 3D dose distribution corresponding to each plenoptic acquisition. In addition to improving computation speed, future development will include using additional physical constraints to better guide the reconstruction process.

4. Conclusion

With the rise of high-conformity dynamic RT, such as volumetric modulated arc therapy (VMAT) and robotic radiosurgery, the temporal dimension is becoming increasingly important. It must be possible to tell both ‘where’ and ‘when’ a discrepancy occurs between the plan and its delivery. A 3D scintillation dosimeter could be ideal for such an end-to-end verification, whereby sources of deviation between planning and delivery could be identified. However, careful optimization of the optical design as well as the data processing is required to have a clinically viable system.

5. References

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