TECHNICAL NOTE

Sensitivity Improvement of Optical Computed Tomography Scanners for Scanning Gel Dosimeters

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Abstract

Despite recent advances in Optical Computed Tomography (OCT) systems, there is skepticism regarding the sensitivity and performance of these systems for gel dosimetry. The optical sensitivity of gel dosimeters changes at different wavelengths and their maximum sensitivities are achieved at the particular wavelengths. Therefore, the optimal wavelength must be used for optical scanning of the gel dosimeters. Since OCT systems are used for scanning different gel dosimeters, a mono-energy light source would not lead to optimal performance. Commercially available multi-wavelength lasers provide a variety of power and wavelength ranges. Adopting multi-wavelength light sources instead of single-wavelength laser sources increases the price of the system by less than 5%, which is justifiable and ignorable due to the considerably increased performance and sensitivity. In the proposed approach, only the wavelength of the scanning beam would be optimized based on the gel dosimeter type. The rules for determining the opacity of the beam path across the sample are the same for all gel dosimeters. Therefore, the change in wavelength of the scanning beam didn't change the scanning procedure and there is no additional complexity in the OCTs with the multi-wavelength light sources.

Keywords: Optical Computed Tomography; Polymer Gel Dosimetry; Multi-Wavelength Lasers; Optical Sensitivity Of the Gel Dosimeters.

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1. Introduction

In addition to chemotherapy and surgery, radiation therapy has a special role in the treatment of cancers. About half of the cancer patients receive radiation treatments [1]. The most important goals of radiation therapy are to deliver the required dose to the target homogeneously and simultaneously sparing surrounding healthy organs. Nowadays, more advanced and complex treatments such as Intensity-Modulated Radiation Therapy (IMRT), Stereotactic Radiosurgery (SRS), Stereotactic Body Radiation Therapy (SBRT), Volumetric Modulated Arc Therapy (VMAT), and Image-Guided Radiation Therapy (IGRT) are used in radiation treatments that can deliver the favored dose distribution with high conformity and precision to the treatment volume. Any error in this dose delivery can result in an insufficient dose to the tumor or a high dose to adjacent critical tissues [2], so more precise dosimetry methods must be used to confirm treatment plans. Conventional dosimeters that measure point or two-Dimensional (2D) dose distributions are not capable of demonstrating the dose distribution in a three-Dimensional (3D) format. In addition, they are not cost-effective in terms of time for 3D arrangements and measurements. Gel dosimetry is the only real 3D dosimetry method that can three-dimensionally measure the dose distribution and confirm treatment plans [3].

Gel dosimetry is a highly sensitive method that can measure low doses around 50 cGY. Polymer gel dosimeters, as one of the most common types of gel dosimeters used in dosimetry studies, have good stability after irradiation and are capable of maintaining the initial pattern of dose distribution over a relatively long time [4]. In Fricke gel dosimeters, ferric ions diffuse in the gel matrix, and dose distribution after irradiation is not spatially stable. The ferric ion diffusion could be reduced by adding chelating agents to the gel formulation [5, 6]. These gel dosimeters are also tissue-equivalent phantoms that can be used to measure the absorbed doses in a three-dimensional arrangement.

In irradiated Fricke gel dosimeters, the ferrous ions (Fe2+) were oxidized due to the radiation-induced changes and turned into ferric ones (Fe3+). The color change occurring in Fricke gel dosimeters with increasing their absorbed doses is the base of the optical readout of radio-chromic gel dosimeters [5]. In polymer gel dosimeters, the gel sample is first irradiated. Monomers that are three-dimensionally dispersed in the polymer gel structure would change to the polymer during a free radical-induced polymerization mechanism. The polymer-to-monomer ratio increases with the absorbed dose increasing and the gel sample gets harder [4, 7]. The 3D dose distributions could be determined by quantifying the radiation-induced polymerization at each point of the gel dosimeter. Therefore, the gel dosimeters record the absorbed doses in a 3D arrangement. To extract the dose magnitudes, the gel dosimeter must be imaged by a scanning system. For evaluating the dose distributions in the gel dosimeters, the calibration curve of the dosimeter in the used scanning system must be determined. First, a certain number of gel dosimeters are prepared. Based on the favored radiation dose range, the dosimeters are exposed to different dose magnitudes. The index corresponding to each scanning method (for example, R2 (S^{-1}) in MRI, optical attenuation (cm^{-1}) in optical scanning method, etc.) is read for irradiated gel dosimeters. The index values are plotted against the absorbed dose magnitudes of the dosimeters [8, 9]. The curve fitted to these points is the calibration curve of the dosimeter and can be used to evaluate the 3D dose distribution in the gel dosimeters. Gel dosimeters are scanned using several imaging methods. The standard method for gel dosimeter scanning is the Magnetic Resonance Imaging (MRI) approach. In magnetic fields, materials have different T1 and T2 relaxation times which are related to the inherent characteristics of the materials [3]. For polymer gel dosimeters, whenever the monomer is converted to polymer and the properties of the polymer gel change, its magnitudes of T1 and T2 relaxation time change. For Fricke gel dosimeters, the ferrous ions (Fe2+) and ferric ions (Fe3+) have four and five unpaired electrons, respectively [5]. Therefore, ferric ions (Fe3+) are more paramagnetic. T1 and T2 relaxation time of the Fricke gel dosimeters, which depends on the presence of paramagnetic agents in the sample, will change by increasing their absorption dose magnitudes [6]. Due to the lack of access to an MRI system for gel dosimetry applications and the high cost of sample scanning by this method, alternative methods have been developed in recent decades. Optical Computed Tomography (OCT) is one of the methods used to scan the gel dosimeter samples. This

method is similar to that of the X-ray CT, except that a visible light source (for example, a laser) is used instead of the X-ray tube [10]. Achieved innovations along with the fast, accurate, and cheap scanning possibility of the OCT method, exhibit the significant potentials and benefits of the method for gel dosimetry applications.

Non-irradiated polymer gels are inherently transparent. However, after irradiation, they increasingly become cloudy with the absorbed dose increasing and their light attenuations increase [4]. In this scanning method, the opacity of the beam path across the sample is determined by the Beer equation (Equation 1):

$$I = I_0 \, exp\,[-\int_{ray-path} \mu(l)dl] \qquad (1)$$

Where I is the measured signal intensity, I0 is the initial signal intensity in the absence of the sample, μ is the light attenuation coefficient and l is the optical path length across the sample [7, 10]. For each scanning projection, the optical absorbance magnitudes must be measured at the wavelength with maximum absorption to match the peak response sensitivity of the gel dosimeter. Therefore, the light source of the system must be mono-energy [11]. The intensity of the light beam passing through the gel is recorded by a light-sensitive detector (photodiode or CCD) [7].

Several geometry designs have been proposed for the OCT systems. In some studies, a new OCT system has been proposed and its performance was evaluated [3, 10, 12-15]. Overall, the OCT systems follow the same rules. Currently, Photodiode- and CCD-based scanning models are commercially available categories of the OCT systems. Photodiode-based OCT systems are gold-standard scanning methods. In Gore *et al.*'s study [16], the performance of a photodiode-based OCT system was investigated for 3D gel dosimetry. In this study, a 633nm red He-Ne laser was used to scan the polymer gel samples. In Oldham *et al.*'s study [15], a Uniphase 532 nm green He-Ne laser was used in a similar system to scan the BANG3TM gel. The results of these studies showed that the photodiode-based OCT systems have a percentage error of 5% and spatial resolution of less than 2 mm for 0-10 Gy doses [17]. In this scanning geometry, all translational and rotational motions must be repeated for each projection, so this method is practically very time-consuming. The main reason for the development of other OCT models, especially fan-beam OCTs, was to overcome the problem of time-consuming scanning of the photodiode-based OCT systems.

In Sakhalkar *et al.*'s study [18], the performance of the CCD-based OCT system was compared with an independent readout of the OCTOPUS™-scanner. For both systems, a 633 nm light source was used to scan the same PRESAGE™ dosimeters. The results of the CCD-based OCT system were in agreement with those of the OCTOPUSTM scanner as well as the calculated dose distributions by the ECLIPSE treatment-planning system. In Jirasek *et al.*'s study [13], another CCD-based scanning system was developed for gel dosimetry applications. In this system, the NIPAM gel was scanned by a fan beam of 543 nm laser light.

Despite recent advances in OCT systems, there is skepticism regarding the sensitivity and performance of these systems for gel dosimetry applications. The optical sensitivity of gel dosimeters changes at different wavelengths and their maximum sensitivities are achieved at the maximum absorption wavelengths. Therefore, the optimal wavelength must be used for optical scanning of the gel dosimeters. For example, the PRESAGETM dosimeter has maximum sensitivity at the wavelength of 633 nm [18], while maximum sensitivity for NIPAM and PAGAT dosimeters was reported in the blue part of the visible light spectrum. These dosimeters have the lowest sensitivity in the red part of the spectrum. The scanning of NIPAM and PAGAT dosimeters at 633 nm can reduce the sensitivity by 58% and 73%, respectively [11]. The available OCT systems could not provide the optimal results for all gel dosimeters.

The sensitivity of gel dosimeters is strongly dependent on the wavelength, and even changing the monomer type could cause a significant change in the optical sensitivity of the gel dosimeters [11]. Therefore, selecting the optimal wavelength for scanning gel dosimeters is one of the most important steps in the design and construction of an OCT system. Since OCT systems are used for scanning different gel dosimeters, a mono-energy light source would not lead to optimal performance. Multi-wavelength laser sources could be used to solve this problem. Commercially available multi-wavelength lasers provide a variety of power and wavelength ranges. Each wavelength can be separately controlled and modulated. Adopting multi-wavelength light sources instead of single-wavelength laser sources increases the price of the system by less

than 5%, which is justifiable and ignorable due to the considerably increased performance and sensitivity of the scanner. The authors believe that the utilization of multi-wavelength lasers can improve the accuracy and sensitivity of the OCT systems for gel dosimetry applications. For scanning each gel dosimeter, the optimal wavelength could be exerted using multi-wavelength lasers. In the proposed approach, only the wavelength of the scanning beam would be optimized based on the gel dosimeter type. The rules for determining the opacity of the beam path across the sample are the same for all gel dosimeters. Therefore, the change in wavelength of the scanning beam doesn't change the scanning procedure and there is no additional complexity in the OCTs with the multi-wavelength light sources.

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