Signal requirements for 3D optically stimulated luminescence dosimetry

Mads L. Jensen¹, Rosana M. Turtos¹, Jacob S. Nyemann¹, Brian Julsgaard¹,², Ludvig P. Muren³,⁴, Peter Balling¹,²
¹Department of Physics & Astronomy, Aarhus University, Denmark
²Interdiciplinary Nanoscience Center (iNANO), Aarhus University, Denmark
³Danish Centre of Particle Therapy, Aarhus University Hospital, Denmark
⁴Department of Medical Physics, Aarhus University/Aarhus University Hospital, Denmark
E-mail: mlj@phys.au.dk

Abstract. For 3D optically stimulated luminescence (OSL) based dosimeters to be clinically applicable, certain standards must be met. Among these are low detectable doses with high accuracy and precision, ideally comparable with those of point-like detectors. By investigating a model of the central part of an OSL readout-system, we present an estimate of \(\sim 4 \cdot 10^7\) photons Gy \(\cdot\) mm\(^3\) as the minimum required signal from OSL active materials embedded in a transparent matrix to allow measuring doses of 0.1 Gy with an accuracy and precision of 2%. Further, 2D spatially resolved measurements of OSL emission from commercially available LiF:Mg,Cu,P pellets are presented and discussed.

1. Introduction

Three dimensional dosimetry is a valuable tool in modern radiotherapy where treatment techniques are becoming increasingly complex. Quality assurance is key in such treatments and 3D dosimetry is an optimal way to provide this. Currently available 3D dosimeters consist of more or less tissue equivalent gels which change optical density according to the radiation dose [1]. The dose deposited in the gels can then be measured by the changes in optical density through optical-computed-tomography scanning. There are challenges of such gel dosimeters, and numerous studies on correcting these are available [1–4]. Unfortunately, one major drawback, namely the permanent nature of the changes in optical density, cannot be corrected. Solutions have been suggested [5,6], but the search for a reusable 3D dosimeter is ongoing.

As proposed by our group, optically-stimulated-luminescence (OSL) active micro/nanoparticles embedded in a transparent matrix could provide a solution to many of the challenges associated with this problem [7, 8]. Metastable trapped states in the active material will be populated when irradiated, and information about the deposited dose is stored in the density of these populated trapped states. To read out this information, the trapped states are emptied by optically stimulating recombination of the trapped electron-hole pairs. The intensity of light emitted from such recombinations will then hold information about the deposited dose. From this point on, this light will be referred to as the OSL signal. Optically manipulating a laser to form a light sheet using a Powell lens and collimating optics will make it possible to read out a single plane of the dosimeter at a time. Moving the dosimeter and hence...
emptying another plane, would then provide the third dimension. Such a dosimeter would be reusable, as it can be reset by bleaching with either light or heat.

In the following, the minimum OSL signal required from the active material for dosimetric applicability will be discussed. This is based on an analysis of the central part of the optical readout setup, i.e. light transmission from the dosimeter to the camera including the intermediate optical elements. Moreover, preliminary spatially resolved measurements of commercially available OSL active LiF:Mg,Cu,P (MCP) pellets will be presented and discussed.

2. Considerations on OSL signal

For a dosimeter to be clinically feasible for patient-specific quality assurance, it must be able to detect low doses with high accuracy and precision. In order to provide a lower bound on the dosimeter signal, this analysis is based on more point-like detector characteristics, requiring detectable doses as low as 0.1 Gy with an accuracy and precision of 2% in a relevant voxel volume. This corresponds to a signal-to-noise-ratio (SNR) of 50, which under the assumption of a shot-noise limited readout-camera yields a minimum detected photoelectron number of $N = \sqrt{\text{SNR}} = 2500$. This number dictates the minimum detected signal needed from a voxel in the dosimeter, which can be converted to a minimal OSL signal needed from the active particles by backtracking through the system illustrated in Figure 1 accounting for signal losses in all elements.

Starting from the dosimeter side, there will be some attenuation of the signal inside the dosimeter described by Lambert-Beer's law of attenuation. The transmitted signal will be given by the coefficient

$$k_{\text{att}} = \exp(-\mu z),$$

where $z$ is the depth in the dosimeter and $\mu$ is the material-specific attenuation coefficient. This is under the assumption that the opening angle, $\theta_{\text{max}}$, corresponding to a solid angle of the lens area, is small. Through a ray-matrix analysis, it can be shown that $\theta_{\text{max}}$ relates to the lens diameter $d$ as $\theta_{\text{max}} = \frac{d}{2n_df_2}$, where $n_d$ is the refractive index of the dosimeter and $f_2$ is the focal length of lens 2. This is valid under small-angle and thin-lens approximations. As the active material is emitting isotropically, the fraction of signal that will be detectable, i.e. emitted in the solid angle related to $\theta_{\text{max}}$, is given by the coefficient

$$k_{\Omega} = \frac{1}{2}(1 - \cos \theta_{\text{max}}) \approx \frac{1}{16} \frac{m^2}{n_d^2 f_2^2},$$

where $m = -f_1/f_2$ is the magnification and $f_{\#1} = f_1/d$ is the f-number for lens 1.

Reflection at the interface between the dosimeter and air will also cause signal loss. Again, considering small angles, it can be justified to estimate this loss using the normal-incidence case of the Fresnel equations, yielding the following transmission coefficient

$$k_T = \frac{4n_d}{(n_d+1)^2}.$$

Both lenses, the filters, and the quantum efficiency of the camera will give rise to further signal losses
and the transmission can be described by coefficients, $k_l$, $k_f$ and $k_{QE}$, respectively, which can be measured or looked up in data sheets.

Regarding spatial resolution, one mm$^3$ is a clinically relevant voxel size. The required number of photons emitted from such a voxel per given Gy of dose can now be estimated as

$$N_{\text{voxel}} = \frac{2.500}{0.1 \text{ Gy}} \prod_{i} \frac{1}{k_i},$$

where $i$ runs over all correction coefficients. A spatial resolution of one mm$^3$ is readily achievable with the transverse resolution ensured by a large CCD and matching optics, and the longitudinal given by light sheet thickness.

Using silicone (Sylgard 184) as a typical elastomer, the refractive index and attenuation coefficient is measured at a wavelength of 370 nm to be 1.45 and 0.07 cm$^{-1}$, respectively. Any scattering from the embedded active particles is neglected in this analysis. Taking the dosimeter to have dimensions of a 10×10×10 cm$^3$ cube and considering relevant CCD sizes to be around 3×3 cm$^2$, the magnification will need to be $m \approx 0.30$. This dosimeter size also yields a maximum depth of $z_{\text{max}} \approx 10$ cm. Further, as the product between solid angle and area is conserved, the f-number of lens 1 will need to be small in order to get the most signal, say about 0.70. Filter, lens transmission, and camera quantum efficiency are set to 0.55, 0.95 and 0.50, respectively. With this,

$$N_{\text{min}} \sim 4 \cdot 10^7 \text{ photons Gy}^{-1} \text{ mm}^{-3}.$$  

This number does not only provide information about the required OSL signal from the active material, but also holds information about the concentration of active material needed in the transparent matrix.

Investigation of the OSL signal from commercially available MCP pellets has been undertaken in the group, and these were found to emit $3.3 \cdot 10^8$ photons Gy$^{-1}$ mm$^{-3}$ [9]. With the above result, this implies a minimum concentration of $\sim 10$ % MCP powder uniformly dispersed in silicone, for the dosimeter to fulfill the precision criteria. With a concentration this high, refractive index matching of the active particles to the transparent matrix would be essential for scattering to be negligible. This estimated concentration exceeds the result from [9] mainly due to a more conservative estimate of the applied magnification used here. More traditional OSL-active materials, such as Al$_2$O$_3$:C, have a higher OSL-efficiency [10], and using these would thus lower the minimally needed concentration of active material.

In summary, this motivates further investigation of the OSL signal from MCP powder embedded in silicone for comparison with the result above and validation of applicability of this material. Such measurements are to be performed in the immediate future.

3. Preliminary 2D readout

As a first step towards a spatially resolved OSL readout, preliminary measurements of commercially available MCP pellets irradiated with 20 Gy have been made using a Falcon Blue 285b-cl EMCCD camera from Raptor Photonics. The pellets were emptied with a laser of wavelength 460 nm transmitted through a circular aperture and imaged onto the pellet with an average intensity of $\sim 750$ mW cm$^{-2}$. Appropriate filters, resulting in a spectral window between 310 and 400 nm, were used to ensure that only OSL signal was detected. Specifically, three OD4 hard coated filters with cut-off wavelength at 400, 425 and 525 nm from Edmund Optics were used along with two colored glass filters, one from Thorlabs (FGUV5S) and one from Edmund Optics (BG-38). The exposure time of the camera was 2 seconds and the optimal gain setting on the camera using 5-by-5 binning was found to be 2450. The first and last frames from this measurement, corrected for a background contribution, are displayed in Figure 2 in the leftmost and rightmost inset, respectively.

In order to investigate whether the obtained spatially resolved signal had indeed OSL origin, the decay of the spatially resolved signal was extracted by fitting all of the frames with a 2D Gaussian
function. Due to a fast decaying signal and a fairly high and fluctuating noise level, a proper Gaussian fit was only possible for the first frame. Amplitudes for the remaining frames were obtained by locking the centroid and width of the Gaussian function to that corresponding to the first frame. Hence, only the amplitude along with the offset and slope of the background plane were fit parameters in all frames except the first. Amplitudes from this analysis can be seen plotted against time in Figure 2.

To compare the obtained decay-signal from the spatially resolved measurement, one of the pellets was investigated in a setup for time-resolved spectral characterization described in [9]. The same laser settings were used to read out this pellet, except with a slightly lower average intensity of $\sim 600 \text{ mW/cm}^2$.

This setup allowed for measurement of the pellet’s OSL decay by integration over all OSL relevant wavelengths, which is assumed to be representative for all the pellets’ decay-behavior when exposed to identical doses. The resulting decay-signal is shown in Figure 2, where the normalized decay-signal from both measurements is seen to be quite similar implying that the spatially resolved data is indeed an OSL signal.

4. Conclusion
We present an estimate of $\sim 4 \cdot 10^7 \text{ photons Gy}^{-1} \text{ mm}^3$ as the minimum number of photons that an OSL active material embedded in a transparent matrix should emit to be clinically applicable for patient-specific quality assurance. Furthermore, we present 2D spatially resolved measurements from a commercially available MCP pellet. By comparing the decay-pattern of these measurements with that from spectral measurements of the pellets, the signal is attributed to spatially resolved OSL signal. Together, these two results are very promising for the 3D OSL dosimetry concept.

5. Acknowledgements
This research was supported by the Novo Nordisk Foundation.

6. References
[1] Doran SJ 2009 Appl. Radiat. Isotopes 67 393-8
[2] Høye EM, Balling P, Yates E S et al 2015 Phys. Med. Biol 60 5557
[3] Høye EM, Skyt P S, Balling P et al 2017 Phys. Med. Bio. 62 73-89
[4] D’Errico F, Lazzeri L, Dondi D et al 2017 Radiat. Meas. 106 612-7
[5] Taño J, Hayashi S, Hirota S et al 2019 J. Phys. Conf. Ser. 1305 012034
[6] Juang T, Adamovics J, Oldham M 2015 J. Phys. Conf. Ser. 573 012039
[7] Sadel M, Høye EM, Skyt PS et al 2017 J. Phys. Conf. Ser. 847 12044
[8] Balling P, Høye EM, Muren LP et al 2018 European Patent Application; EP3264137A1
[9] Nyemann JS, Turtos RM, Julsgaard B et al 2020 Radiat. Meas 138 106390
[10] Akselrod MS 2011 AIP Conf. Proc. 1345 274-302