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Technical note

Temporally separating Cherenkov radiation in a scintillator probe exposed to a pulsed X-ray beam

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ABSTRACT

Cherenkov radiation is generated in optical systems exposed to ionising radiation. In water or plastic devices, if the incident radiation has components with high enough energy (for example, electrons or positrons with energy greater than 175 keV), Cherenkov radiation will be generated. A scintillator dosimeter that collects optical light, guided by optical fibre, will have Cherenkov radiation generated throughout the length of fibre exposed to the radiation field and compromise the signal. We present a novel algorithm to separate Cherenkov radiation signal that requires only a single probe, provided the radiation source is pulsed, such as a linear accelerator in external beam radiation therapy. We use a slow scintillator (BC-444) that, in a constant beam of radiation, reaches peak light output after 1 microsecond, while the Cherenkov signal is detected nearly instantly. This allows our algorithm to separate the scintillator signal from the Cherenkov signal. The relative beam profile and depth dose of a linear accelerator 6 MV X-ray field were reconstructed using the algorithm. The optimisation method improved the fit to the ionisation chamber data and improved the reliability of the measurements. The algorithm was able to remove 74% of the Cherenkov light, at the expense of only 1.5% scintillation light. Further characterisation of the Cherenkov radiation signal has the potential to improve the results and allow this method to be used as a simpler optical fibre dosimeter for quality assurance in external beam therapy.

1. Introduction

In any optical system exposed to ionising radiation, Cherenkov radiation is generated throughout the components. The cause of this Cherenkov radiation is when charged particles travel faster than the local speed of light in a medium. While Cherenkov radiation itself can be used for dosimetry purposes [1], the focus of this work is removing Cherenkov radiation contamination in scintillation dosimetry, where a scintillator is used as the radiation conversion material, and the scintillation light generated is guided by optical fibre to an optical detector. Because the air kerma rate of plastic scintillators is both energy-independent and linear over a large range of energies (as is the in-water dose response over MeV photon energies, as shown by Beddar et al. [2]), the integral of the detector signal over a period of time provides a measure of the relative dose during that time. In such a dosimeter, Cherenkov radiation is generated not only in the scintillator, but also any optical fibre exposed to the radiation field. The Cherenkov spectrum overlaps with the output spectrum of scintillators, and so compromises the total signal measured by the detector. Optical dosimeters using plastic scintillators are very desirable due to their water equivalence, and so methods to remove the Cherenkov contamination are required for valid dosimetric applications.

There are currently several successful methods for removing the Cherenkov radiation component from the signal. The simplest method, developed by Beddar et al. [3,4], uses a probe without the scintillator to measure only the Cherenkov signal. This requires both probes and fibres to be kept together to ensure they are exposed to the same amount of radiation. Using a Cherenkov probe is considered the standard method for Cherenkov removal in the field. However, this method is not valid in regions of high dose gradient, such as in synchrotron microbeams, where the Cherenkov radiation generated in each fibre may be significantly different.

Using an optical fibre with an air core can remove the generation of Cherenkov radiation altogether [5,6]. This makes an arrayed system practical, as demonstrated by Naseri et al. [7]. However there are several limitations of using an air core optical fibre. The attenuation of

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the air core fibre is an order of magnitude greater than plastic core fibres, so the scintillation light signal strength must be very large to compensate for this. Further, the air core fibres are very inflexible, making *in vivo* applications impractical.

Optical filtration can be used to reduce the contribution of Cherenkov radiation in the total signal, by exploiting the fact that the Cherenkov intensity is proportional to $1/\lambda^3$, so filtering longer wavelengths will preferentially filter the Cherenkov light more than the scintillator [8]. However there will still be scintillator light removed from the signal. For example, a study into Cherenkov separation methods identified that the most efficient filter reduced the scintillation signal by 56%, and the Cherenkov radiation by 82% [9]. The optical filtration can also be improved by using a scintillator that emits at higher wavelengths [10].

Spectral analysis can also be used to remove the Cherenkov radiation. By measuring the spectrum of the incident light signal, and comparing the intensities when the scintillator signal peaks, and a region where there is minimal scintillation light, Fontbonne et al. [11] presented a mathematical method for calculating the Cherenkov contribution. Using a CCD camera Frelin et al. [12] was able to use this method to remove the Cherenkov signal from the total signal. This has also been achieved with a gallium nitride inorganic scintillator, removing only 1.2% of the scintillation light [13]. A commercial scintillator dosimeter has been developed using spectral analysis of comparable accuracy to other products [14,15].

Methods to temporally separate the Cherenkov signal from the scintillation signal in a pulsed radiation source has been explored [16]. The authors claim a removal of 99.9% of Cherenkov light, at the expense of 44% of the scintillation light. They used a scintillator with a long decay time (257 ± 6 ns) and integrated the detector response over a 5 ns duration, 700 ns after the 450 ns linear accelerator (LINAC) electron beam pulse began. The LINAC pulse was chosen to be 450 ns to ensure the scintillation signal was not saturated, so that the tail of the signal was proportional to the integral of the signal and hence the absorbed dose. However for longer pulse durations this method is not valid as once the signal saturates, the tail is no longer proportional to the total dose delivered.

We present a novel method for pulsed radiation sources that is valid for arbitrarily long pulses. This method has been developed to be used with a water equivalent scintillator dosimeter, due to solutions presented previously often not being compatible with a water equivalent dosimeter. It relies on the different rates of rise time between the Cherenkov signal and the scintillator signal. By using a scintillator with a slow rise time, the two components of the signal will be separate enough to be discriminated. BC-444 plastic scintillator (from Saint-Gobain Crystals) has a rise time (τ_r) of 19.5 ns, and a decay time (τ_d) of 285 ns. When energy is instantaneously deposited in the scintillator (instantaneously meaning over a duration much less than τ_r) at $t = 0$, the light output as a function of time ($I(t)$) is modelled by [17]:

$$I(t) \propto e^{(-t/\tau_d)} - e^{(-t/\tau_r)} \tag{1}$$

For a continuous beam, where there is energy being deposited in every instance of time between $t = 0$ and $t = T$ (for a clinical linear accelerator (CLINAC) this is typically 3.5 μs), the instantaneous light output can be integrated up to any point in time, and so the net light output at any point in time can be modelled as:

$$I_{net}(t) \propto \begin{cases} \tau_d\,[1-e^{(-t/\tau_d)}] - \tau_r\,[1-e^{(-t/\tau_r)}], & \text{if } 0 < t < T \\ \tau_d e^{(-t/\tau_d)}\,[e^{(T/\tau_d)}-1] - \tau_r e^{(-t/\tau_r)}\,[e^{(T/\tau_r)}-1], & \text{if } t \geqslant T \end{cases} \tag{2}$$

Using the rise and decay times of BC-444, the actual time for the scintillation signal to reach its maximum (in a 3.5 μs radiation pulse) is closer to 1 μs, rather than 19.5 ns. As the Cherenkov light is detected much quicker than 1 μs (the Cherenkov light is generated instantaneously as the charged particle crosses the fibre, and the transmission will take roughly 50 ns, with a dispersion of 8 ns due to the possible transmission paths in the fibre), the two signal can be distinctly measured on a waveform with a high enough temporal resolution.

2. Materials and methods

The dosimeter probe design is identical to that in a previous work [18], but uses BC-444 plastic scintillator instead of BC-400, due to the slower rise time. BC-444, of thickness 500 μm, was optically coupled to a plastic core optical fibre with core diameter 1 mm (Eska CK-40), giving a sensitive volume of 0.393 mm_3. The end of the probe was coated in BC-620 diffusive reflective paint (Saint-Gobain Crystals) to maximise the captured light. The fibre light signal was measured with a RCA-4526 photo-multiplier tube (PMT). The PMT signal was sampled at 1 GHz with a digital oscilloscope, corresponding to a sample time of 1 ns, which is sufficient to detect the rise of the Cherenkov light signal. Measurements were recorded as voltage readings from the oscilloscope.

The method presented was developed to be tested in a Varian LA1-EX CLINAC 6 MV X-ray field. The field size used was 10 × 10 cm^2 with a source to surface distance of 100 cm. The X-ray beam is fired in pulses of 3.5 μs, with 2.7 ms between each pulse firing. The pulsed nature of the beam is vital to the method, which relies on the difference in times taken for the Cherenkov radiation and the scintillation light to be detected.

We present a novel algorithm to determine the Cherenkov contribution to the total light signal, based on two assumptions: that the Cherenkov light output is constant once it peaks, and the scintillation light rises much slower than the Cherenkov light does. The algorithm works in four main steps:

1. Determine the time (and hence voltage reading) when the Cherenkov signal peaks.
2. Determine the time (and hence voltage reading) when the Cherenkov signal cuts out (stops).
3. Use the first voltage (Cherenkov peak value) to subtract from the total voltage signal between the times calculated in steps 1 and 2.
4. Integrate the remaining signal between the Cherenkov peak time and the Cherenkov stop time, relative to the baseline voltage. Integrate the total signal after the Cherenkov stop time, relative to the baseline after the beam is off.

We applied this algorithm to produce a beam profile and depth dose plot of the CLINAC field. We investigated the effects of using different methods to calculate the Cherenkov start time, as well the optimisation of the smoothing method, on the accuracy of the algorithm.

The first limitation of this is that the scintillation signal generated before the Cherenkov peak time will be neglected from the total integration, and so will slightly underestimate the scintillation light signal. The second limitation is that if the assumption that the Cherenkov contribution remains constant after peaking is not valid, the subtracted signal will not be correct.

To determine the Cherenkov peak time and Cherenkov stop time, we analyse the derivative of the voltage waveform. Note that the voltage waveform is inverted, as the PMT is supplied with a negative high voltage. There are three important features in the derivative relevant to our algorithm: the global minimum (when the Cherenkov signal is increasing the fastest), the global maximum (when the Cherenkov signal is dropping off the fastest) and the local maximum within 400 ns after the global minimum. The derivative, and two of these features are illustrated in Fig. 1.

The local maximum soon after the global minimum indicates when the Cherenkov signal has peaked, and so searching within a 400 ns window from the easily found global minimum provides the time and voltage reading of the now constant Cherenkov contribution.

The derivative is simply calculated using a first order backward difference method on the data set x_i:

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