Physica Medica 30 (2014) 454–461

Contents lists available at ScienceDirect

Physica Medica

journal homepage: http://www.physicamedica.com

Original paper

Under-response correction for EBT3 films in the presence of proton spread out Bragg peaks

F. Fiorini [a,*], D. Kirby [b], J. Thompson [a], S. Green [b], D.J. Parker [c], B. Jones [a], M.A. Hill [a]

[a] Gray Institute for Radiation Oncology and Biology, Department of Oncology, University of Oxford, Roosevelt Drive, Oxford OX3 7DQ, UK
[b] Hall-Edwards Radiotherapy Research Group, Department of Medical Physics, University Hospital Birmingham NHS Trust, Birmingham B15 2TH, UK
[c] School of Physics and Astronomy, University of Birmingham, Edgbaston, Birmingham B15 2TT, UK

ARTICLE INFO

Article history:
Received 7 August 2013
Received in revised form
24 November 2013
Accepted 30 December 2013
Available online 22 January 2014

Keywords:
3D dosimetry
EBT3 Gafchromic films
Under-response correction
FLUKA Monte Carlo code

ABSTRACT

We present a study of the under-response of the new Gafchromic EBT3 films and a procedure to accurately perform 2D and 3D proton dosimetry measurements for both pristine and spread out Bragg peaks (SOBP) of any energy. These new films differ from the previous EBT2 generation by a slightly different active layer composition, which we show has not effected appreciably their response. The procedure and the beam quality correction factor curve have been benchmarked using 29 MeV modulated proton beams. In order to show the correction to apply when EBT3 films are used as treatment verification tools in anthropomorphic phantoms, two simulation studies involving clinical energies are presented: a SOBP for eye treatments and a SOBP to treat 20 cm deep and 5 cm thick tumours. We find maximum under-responses of 37%, 30% and 7.7% for the modulated 29 MeV beam, eye and deep tumour treatment, respectively, which were attained close to the end of the peak tails, due to a higher proportion of very low energy protons. The maximum deviations between corrected and uncorrected doses were for the three cases, respectively, 20.7%, 8.3% and 2.1% of the average dose across flat region of the SOBP. These values were obtained close to the distal edge of the SOBPs, where the proportion of low energy protons was not as high as on the tail, but there still was a number of protons high enough to deposit a reasonable amount of dose in the films.

Introduction

The ideal form of quality assurance in radiotherapy dosimetry should include three-dimensional dose verification. In the case of proton radiotherapy ionisation chambers have been most commonly used, but despite their convenience in terms of immediate results they do not easily provide 2D or 3D dose distributions. On the other hand, radiochromic Films (RCF) are well-established and reasonably cheap dosimeters for conventional radiotherapy [1–3], but the amount of work required to extract dose information in the presence of clinical proton radiotherapy makes them highly time consuming even if the resolution they can give for 2D or 3D dose maps is much higher than many commercially available conventional dosimeters. This is not only due to the long handling process (including calibration batch by batch, cutting, scanning and analysis film by film off line), but also to the fact that their response is energy dependent with increasing under response at low particle energies, or high LET [4,5] (the energy dependence in the case of photons was seen to be much lower [6]). In fact, the darkening effect caused by the presence of organic monomers which polymerise under irradiation and which allows us to measure an increased optical absorption with increased absorbed dose is compromised at high LET irradiations. In one of the theories explaining this effect the polymerisation sites are spaced out with some separation: if all sites close to a single ionising particle track are hit, the polymerisation of the film is locally saturated and part of the particle energy loss remains unmeasured [7]. This means that whenever RCFs are used to detect the dose released by low energy ions, the response has to be corrected with a correction factor dependent on the energy of the crossing particles [8].

The present work is focused on the last generation of EBT GafChromic films, EBT3, which were introduced into the market in 2012. EBT3 films, as well as the previous generation EBT2 films, are particularly useful for radiotherapy dosimetry because of: their high saturation dose (up to 3×10^3 Gy [9]) and minimum detectable dose of around 0.01 Gy; their excellent spatial resolution ($\sim$25 μm); their faster optical density growth after exposure; their strong and compact composition which allow them to be used in water phantoms for a few hours without degradation and with the

* Corresponding author. Tel.: +44 7595919690.
E-mail addresses: francesca.fiorini@oncology.ox.ac.uk, francesca.fiorini83@gmail.com (F. Fiorini).

1120-1797/$ – see front matter © 2014 Associazione Italiana di Fisica Medica. Published by Elsevier Ltd. All rights reserved.
http://dx.doi.org/10.1016/j.ejmp.2013.12.006

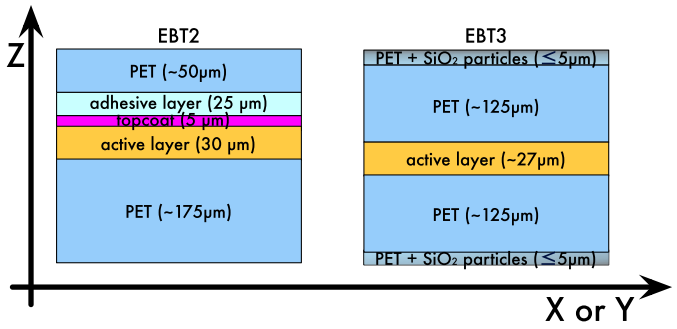

Figure 1. Geometric differences between EBT2 and EBT3 films.

water only penetrating 1–2 mm along the edges; their very low sensitivity to visible light; response independence on the dose rate at high dose rates and high doses [9], and also because they are nearly tissue equivalent. The biggest improvement with respect to their predecessor, EBT2, is the presence of a coating including silica particles (amounting to around 0.1% by weight) which help minimise Newtons rings patterns from forming [10]. Another improvement lies in the fact that EBT3 films are now symmetrical along their thickness (*Z* axis) as shown in Fig. 1, so that the depth of the active layer, and therefore the position at which the dose is measured, is independent on which side is facing the beam.[1] In order to have a 3D map of the delivered dose, a stack of films needs to be used: each single film will be able to record the dose along the irradiated surface (*XY* plane) with a resolution equal to the pixel size of the scanned film image and the sequence of films will show how the dose changes along the depth (*Z* axis) with a resolution equal to the thickness of the films (~290 μm).

With this work we assess the under-response of EBT3 films down to energies lower than 2 MeV where we expect to see the largest under-response starting from the assumption that the material composition of their active layer is not too different from the active material of EBT2 films (Ashland, 2012, private communication, see Tables 1and 2) for which the under-response is already known [11]. In the literature we found another group comparing EBT2 and EBT3 film response under proton irradiation [12]. They showed that there was no apparent difference between the response of the two film types, however the resolution at energies lower than 10 MeV was not high enough to provide accurate correction factor values to apply to the doses extracted from the EBT3 films and obtain the real absorbed ones. In this work we determined that the correction factor curve used in Ref. [11] to correct for the under-response of EBT2 films can effectively be used also for EBT3 films even for energies lower than those tested in Ref. [12].

Recently, Carnicer and collaborators have presented a correction curve for EBT3 films determined using a proton SOBP for eye treatment [13]. This means that their analysis implicitly assumes a specific energy distribution spectrum for the protons irradiating the films, and is therefore not extendible to other conditions. For this reason, contrarily to what the authors state, the method and correction curve they present cannot be used for films irradiated by any proton energy, but only for those irradiated by a SOBP identical (or very similar) to the one used to determine the shown correction curve. In our study, the explained method to correct for the under-response of the EBT3 films, as well as the parameterised quality correction factor curve, can instead be considered for any proton energy. We also illustrate the relevance of the correction in the presence of spread out Bragg peaks (SOBPs) for clinical and non-clinical situations. In particular, for the non-clinical initial low energies, $E_{in} < 30$ MeV, an experiment carried out using the proton beam accelerated by the cyclotron of the University of Birmingham is reported and the procedure to obtain the correct dose deposition explained. For the clinical initial energies ($39 \leq E_{in}$ (MeV) ≤ 60 for typical eye treatments and $195 \leq E_{in}$ (MeV) ≤ 217 for treatments of tumours 20 cm deep and 5 cm thick) Monte Carlo simulations are presented as examples showing the correction to be applied for both cases. The explanation about the conversion from dose-to-EBT3-film to dose-to-water usually required in clinical situations and the conversion from range in film stack to range in water required in 3D dosimetry in water phantom measurements are also given.

Table 1
Material composition of EBT2 films: percentages by atom.

EBT2	H	Li	C	N	O	Na	S	Cl	K	Br
Act. Layer	58.3	0.8	29.6	0.1	10.7	–	–	0.3	0.1	0.1
Adh. Layer	57.1	–	33.3	–	9.5	–	–	–	–	–
TopCoat	56.9	0.9	25.7	–	15.6	–	–	0.9	–	–
PET	36.4	–	45.5	–	18.2	–	–	–	–	–

Table 2
Material composition of EBT3 films: percentages by atom.

EBT3	H	Li	C	N	O	Na	S	Cl	K	Br
Act. Layer	58.2	0.8	29.2	0.1	10.7	0.1	0.1	0.9		
PET	36.4	–	45.5	–	18.2	–	–	–		
PET + SiO_2	PET = 99.986, SiO_2 = 0.014									

Materials and methods

Calibration to dose-to-film

The first step to follow in order to use EBT3 films for accurate dose distribution determination is the calibration, which relates the dose deposited in the active layer of the films to the film optical density (OD). The doses deposited in the film active layers have to be determined using a reference dosimeter, such as an ionisation chamber. In order to keep the calculations of the under-response as easy as possible, we suggest to calibrate the films to dose-to-film (or better dose-to-EBT3 active layer) and then convert these doses to dose-to-water (or to any other material of interest) after the doses from the films have been corrected for the under-response. The advantage of this lies in the fact that the LET correction is kept independent on the material for which the doses are needed, because it only depends on the active layer material (which will always be the same if EBT3 films are used). Using dose-to-water, dose-to-tissue or dose-to-Perspex would required a separate calculation of the appropriate LET correction for each one of these materials.

One of the most accurate methods to extract the OD (or better $OD_{red,corrected}$)[2] from the irradiated films is reported in the *Gafchromic EBT2 Self-developing film for radiotherapy dosimetry* guide,

[1] However it is necessary to remember that, like their predecessors, EBT3 are not symmetrical along *X* and Y axes due to the preferred orientation of the active molecules (LiPCDA) along their long axis so that they scatter light differently in the orthogonal direction. For this reason, marking the films is still important in order to remember the orientation during irradiation and scanning.

[2] $OD_{red,corrected}$ is the optical density extracted from the red channel and corrected according to the values of the optical density extracted from the blue channel as explained in the *Gafchromic EBT2 Self-developing film for radiotherapy dosimetry* guide.