DEVELOPMENT OF 3D DOSE VERIFICATION SYSTEM FOR SCANNED ION BEAM AT HIMAC

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Abstract

A 3D (three-dimensional) dose verification system has been developed as a part of new project involving 3D pencil beam scanning at HIMAC (Heavy Ion Medical Accelerator in Chiba). This system provides the easy and rapid dose measurements. It consists of a water column, a scintillator and a charge-coupled device, set at isocentor. The scinillator is directly attached to the downstream side of the water column. One of the great advantages of this system is to obtain 2D dose map at once, by correcting LET-dependent quenching. Quenching is corrected by measurement slice-by-slice, of a certain water depth. We present the results of 3D dose measurement and comparison with the ionization chamber measurement.

INTRODUCTION

A new project using 3D pencil beam scanning at HIMAC has been underway since 2006 [1], [2]. In this project, there is a plan to construct new treatment rooms, and to extend existent beam line. As part of new project we developed a dose verification system. We describe in this paper the methods of measurment 3D dose distribution and the results. The results are compared with the ion chamber measurement. Then, we found to be a quite useful and helpful in the QA (quality assurance) process. Although the conventional measurement of dose distribution by scanning an ion chamber gives us an accurate dose, it requires considerable time and effort. Furthermore, the ion camber has poor spatial resolution in comparison to sharp dose fall-off by scanned ion beams. To easily obtain the high spatial resolution, we developed a 3D dose verification system by using a scintillator and a CCD (charge-coupled device) camera. The CCD, as measuring instrument, enables us to take the high resolution measured data simply.

Dosimetry with a scintillator and a CCD was developed by Boon et al [3]. They measured the dose distribution by scanning proton beam. And it is used for QA at PSI [4]. However, the scintillator suffers from quenching especially for high LET (Linear Energy Transfer). In our case, scintillator suffers from quenching effect around bragg peak by scanned clinical carbon beam. The brightness is not proportional to physical dose. Therefore, we propose a practical method which makes it possible to easily correct quenching effect. It translates the brightness to dose by slice-by-slice measurement and using the premeasurement. The 2D dose distribution of the certain depth is determined at one measurement. In order to obtain 3D dose distribution, we measure repeatedly at various depths. The measured images after quenching correction were compared with the ion chamber measurement. These differences are almost within 4% inside the target volume.

MATERIAL AND METHODS

Experimental Setup

An experiment was carried out using a secondary beam line (SB1) at HIMAC. In the experiment we employed a mono-energetic 350 MeV/u ^{12}C beam. As shown in Figure 1, the measurement system was set at isocentor. This system consists of a water column, a scintillator, a mirror and a CCD camera. The CCD+screen system was placed in an opaque (black) box to shade light. This water column was designed for the non-waterproof detector.

Figure 1: Photograph of measurement system set at isocentor.

For obtaining high luminescent efficiency and spatial resolution, we chose thin ZnS:Ag(10mg/cm^2) as a scintillator. It is coated on black acrylic board. This scintillator emits light in wavelengths around 450 nm. The mirror which reflects the light on the surface was placed inside the black box. The mirror is set to be the angle of 45 degrees to beam axis. Scintillation light was reflected on mirror, and then measured by the CCD. The CCD camera (Type BU-41L, 1360x1024 pixels, Bitran Corp., Japan) was installed perpendicular to the beam axis, and the distance to the beam axis was 500 mm. Since, the CCD has a peak response in wavelength around 500nm, it corresponds very close to the wavelength of the scintillation.

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Quenching Correction Methods

Brightness is not proportional to dose at high LET, because the scintillator suffers from quenching. Therefore, we established a practical method which make possible to quenching effect based on slice-by-slice measurement. In other words, brightness is translated into dose in this method according to the following steps.

At first, it is necessary to prepare the quenching correction data. The depth dose distribution and the depth brightness distribution are measured by using the ionization chamber and the CCD system, respectively. By using this data, as shown in Fig. 2 (a), we can define the correction factor as a function of depth, which correspond to the ratio of the quenching. After this preparation, we measured the brightness of one irradiation with slice-by-slice manner, as shown in Fig. 2 (b). In other words, we measured each energy and also LET, because all particle pass through the same amount of material thickness. While assuming the same quenching over the whole screen, the observed light intensity is proportional to only with particle flux.

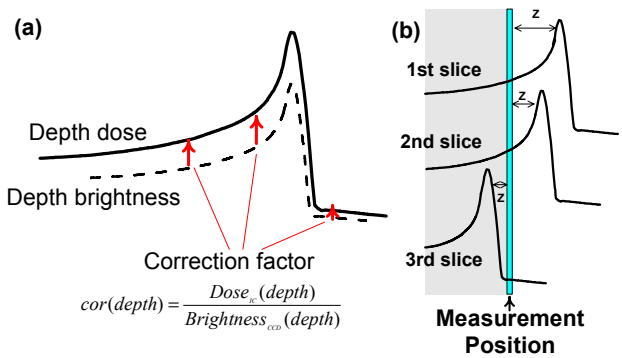
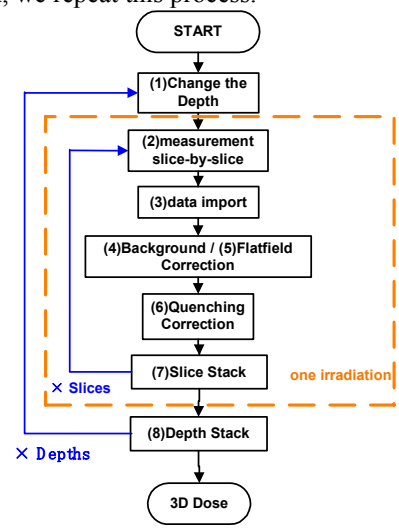


Figure 2: (a) Definition of the correction factor as a function of depth, involving premeasured depth dose distribution and depth brightness distribution. (b) Concept of quenching correction. When we measured by scintillator at a certain depth, the brightness is corrected by premeasured correction factor: cor(depth).

Figure 3 shows the process of conversion into dose from brightness after the measurement. (1) We set the measurement depth of the detector. (2) The brightness images are measured by slice-by-slice manner in one irradiation. In other words, we measure each indent energy at same measurement depth. The fluorescence is integrated during each slice of irradiations. (3) These brightness images are imported to the PC. The measured brightness distributions are corrected according to the following processes. (4) In order to filter the thermal noise from these images, the background image measured without irradiation is subtracted from the measured image. (5) After the background correction, in order to correct the sensitivity of each position of CCD and the position dependent thickness difference of the scintillator, these images are corrected by flat field correction. The images are divided by the flat field image. The flat field image is defined as brightness distribution irradiated by pencil beam at even intervals on plane at the same depth. (6) Next process is the quenching correction. This process makes it possible to convert from brightness to dose. In this correction, the quenching correction factor multiplies to the measured date. This factor is defined as rate of integral dose measured by ion camber in φ120 mm and integral brightness measured by CCD in the same size for pencil beam at each depth (figure 4). (7) Summing up of these corrected images represent the 2D dose distribution (lateral dose distribution). (8) To obtain the 3D dose distribution, we repeat this process.



$$Dose(x,y,z) = \sum_i cor(z - z_i) \times \frac{dat(x,y) - BG(x,y)}{FL(x,y) - BG(x,y)}$$

Figure 3: Flowchart of conversion from brightness to dose, and equation for translation from brightness to dose involving the correction.

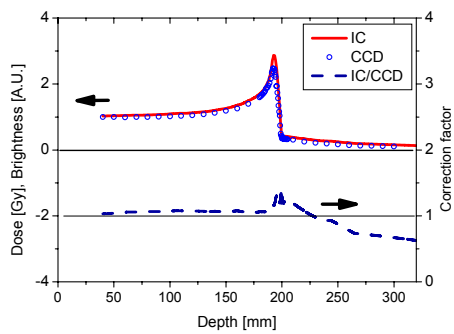


Figure 4: Quenching correction factor :cor(depth) at each depth. The rate of dose and brightness.

RESULT

Dose Linearity

In two types of range sifter, dose linearity is confirmed under the same incident energy. Dose integral inside ϕ

120 mm irradiated pencil beam is estimated. The test results show that this system response linearity, and make possible to comparing at same depth.

Comparing With Ion Chamber Matrix

In this study, the PTV of an osteo-sarcoma treated at HIMAC modified for scanning irradiation is selected as a sample clinical target. In which irradiation parameters were optimized to obtain the uniform physical dose of 0.5 Gy by using the research version of the planning code. Parts of result measured by CCD are shown figure 6. And comparing with ion chamber matrix's one (2D Array XDR, PTW Freiburg, Germany) is shown figure 7. The ion chambers are 5 mm x 5 mm x 5 mm in size, and the center-to-center spacing is 10 mm. In total there are located 729 chambers in a matrix of 27 x 27. The result by CCD is adopted to ion chamber matrix. As a result there is good agreement and variation in rate around 4%. This variation is probably controlled by the decrease in signal-noise ratio of the ion chamber, because it fluctuate in low dose measurement. In addition it seems that uneven thickness of the scintillator is relatead to the debasement of result.

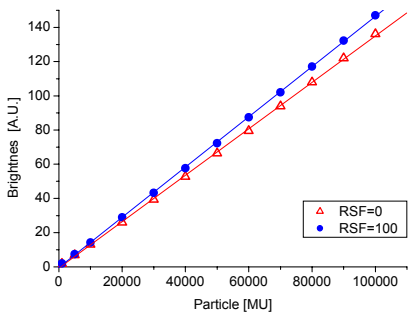


Figure 5: Dose response as function of monitor units for rang shifter setting of 0 and 100 mm.

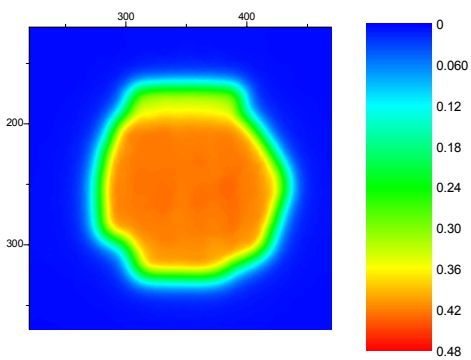


Figure 6: Measured dose distribution at depth 117mm by scintillator and CCD camera.

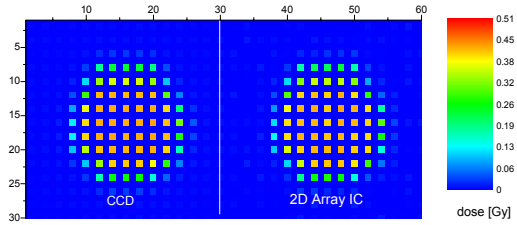
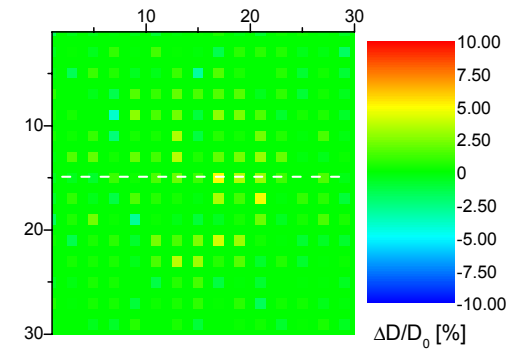


Figure 7: Measured dose at depth 117mm by the CCD system and the 2D ionization chamber matrix.



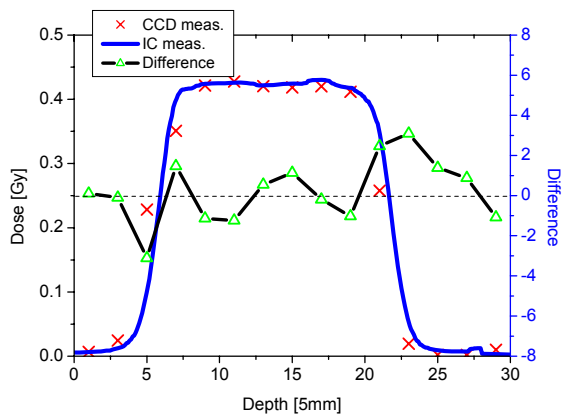


Figure 7: comparison 2D arrayed ion chamber with CCD.

SUMMARY

We developed the easy and quick dose verification system by using scintillator and CCD. The result is good agreement to ion chamber matrix. Difference between the ion camber matrix is within 4%.

REFERENCES

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