

The influence of an immobilization mask on the absorbed dose in the build-up region of photon beams

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Abstract

The purpose of this work is to study how a mask, used for immobilization, alters the depth dose in the build up region of photon beams as compared to the situation in the absence of the mask. Depth dose curves in the build up region of photon beams (^{60}Co , 6 MV) and with field sizes $5 \times 5 \text{ cm}^2$ and $15 \times 15 \text{ cm}^2$, were measured by 3-4 batches of LiF TLDs (more than 50 LiF TLDs, 0.13 mm thin) alternatively by a plane parallel ionisation chamber (NACP 02). The experiments were carried out in the presence and absence of the mask. To ensure the linearity of the dosimeters, the dose was limited to 1 Gy. The data show that the mask can significantly increase the dose at the surface from 11% to 76% for $5 \times 5 \text{ cm}^2$ and from 25% to 83% for $15 \times 15 \text{ cm}^2$ in the ^{60}Co . In the 6 MV beam it alters the surface dose from 7% to 48% for $5 \times 5 \text{ cm}^2$ and from 18% to 56% in $15 \times 15 \text{ cm}^2$ field. The mask will actually change the dose in the whole build up region but the dose increase will be less with increasing depth up to nearly the maximum dose where no more influence of the mask is shown. The

data show that the relative influence of the mask in term of increased absorbed dose in the build up region is higher in small fields compared to large fields but in term of the absolute absorbed dose, in large fields is of course higher than in small fields. The effect of the mask on the surface dose is higher in ^{60}Co as compared to 6 MV. The results show that the mask will substantially increase the absorbed dose in the build up region. This is particularly important at high doses where skin reactions can be a problem. This dose increase should be verified experimentally for each radiation beam and type of mask in order to obtain for considering to decrease early and late radiation induced effects in normal tissues.

Introduction

The principle of curative radiotherapy is to provide maximum dose to the target volume without inducing severe early and late complications in normal tissues and thus maximizing the quality of life for the patients⁽¹⁾. Various approaches are combined to fulfill this aim. For instance,

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megavoltage machines are chosen to deliver the maximum dose at deeply located tissues without injuring the skin⁽²⁾, multiple beam portals and non coplanar techniques are combined to maximize the tumour dose while minimizing the dose to normal tissues and methods of treatment planning optimization including intensity modulation using scanned beams, multileaf collimators, etc are applied⁽³⁻¹⁰⁾. In addition, the patient is immobilized appropriately in order to achieve reproducibility in the every day treatments⁽¹¹⁾. The mask, a particular kind of immobilization device, is used to immobilize the head and neck region⁽¹²⁾. It is commonly a thermoplastic 2 mm thick sheet, which becomes flexible in hot water while it reobtains rigidity after about 5 min exposure in air. It can be shaped covering the head and the patients face from the middle of the forehead to the neck thus resulting in accurate patient repositioning. However, the mask is also irradiated during the treatment thus altering substantially the beam properties. This is due to the interactions of the primary and even secondary photons with the mask itself and the consequence production of secondary electrons which increase the dose to the skin and alter the whole build up region of the depth dose curve^(2, 13-19). It has been shown that these interaction processes may result in an increasing skin dose by up to 113% in a 6 MV bremsstrahlung beam respectively 165% in a ^{60}Co gamma ray beam⁽¹⁷⁾. The direct effect of the increasing dose in the presence of the mask is that the skin sparing effect will be lost. For a normal therapeutic treatment with tumour dose of 60 Gy, the skin is expected to receive doses of about

10-15 Gy. However, in the presence of the mask the skin dose may be increased up to 20-30 Gy which may result in erythematous skin while cumulative doses of about 50-60 Gy may result in moistly epithelolysis. It is however worthwhile to note that the relation between dose and induced skin effects depends strongly on the sensitivity of the individual patients to the irradiation. Measurements of the dose in the build up region are difficult since the presence of conventional dosimeters will perturb the radiation beam during lack of electronic equilibrium. This effect depends directly on the size and the material of the dosimeter⁽²⁰⁻²⁵⁾. Thus, the dosimeters should be preferred very thin and ideally based on material with properties as close as possible to tissue or water. The thin TLD or the plane parallel ionisation chamber with a thin entrance window is suitable for dose determination in regions like the build up of the radiation beam where lack of electronic equilibrium⁽²⁶⁾.

Materials and Methods

Build up curve

The build up curve is defined as the part of increasing dose with depth in an irradiated medium until a balance is achieved between the most penetrating electrons and those attenuated. At this particular point the maximum absorbed dose to the irradiated medium is obtained.

Irradiation units

Photon beams were obtained by a ^{60}Co gamma ray unit (Alcyon II, GE) and a 6 MV Linear accelerator (Philips SL75-5). For the ^{60}Co unit the maximum dose is provided at the depth of about 5 mm in water

for a field size of 10 x 10 cm² while for the 6 MV bremsstrahlung beam and the same field size it is located at the depth of 15 mm. This implies that during normal treatments the skin sparing effect during high dose delivery is substantial.

TLDs Dosemeters

LiF-Teflon, thermoluminescence dosimeters (TLDs) with density 2.64 g. cm⁻³ were used for the measurements⁽²²⁾. The LiF TLDs have thickness 0.13 mm, corresponding to water equivalent thickness of 0.34 mm, and are 12 mm in diameter. During irradiation traps are induced in the TLD material itself. The traps are then released when the TLD is heated giving off the energy in terms of light. The light can be measured accurately by a photo multiplier and this signal can be correlated to radiation dose⁽²⁷⁾

TLDs sensitivity calibration^(22,27,28)

The TLDs were zeroed at 300°C for 0.5 h and then rest in room temperature for 15 min. They were then stored in 80°C for 24 h, again rest in room temperature for 15 min and finally irradiated in a ⁶⁰Co gamma beam. After irradiation they were rest in room temperature for 1 h, then in 80°C for 15 minutes and finally rest in room temperature for 15 min. The signal was measured by a TLD 7300 Teledyne reader. The individual sensitivity of the dosimeters in each batch was measured 3 times. The average sensitivity was then obtained for each dosimeter. Then mean readout of the whole batch was then defined as:

$$\bar{X} = \frac{\sum_{i=1}^N X_i}{N} \quad \dots(1)$$

where $\bar{X}$ is the mean value of the readout of the dosimeters, N is the number of dosimeters and $\sum X_i$ is the total value of readout of all dosimeters in the same batch. The sensitivity of each dosimeter, C_i , was defined as:

$$C_i = \bar{X} / X_i \quad \dots (2)$$

where X_i is the readout value of i-dosimeter. Dosimeters with individual sensitivity variation of more than $\pm 5\%$ were then disregarded.

Ionisation chamber

A plane parallel ionisation chamber (NACP 02) with front wall of 0.2 mm mylar foil and 0.5 mm thick contacting high voltage electrode of graphite and with the effective point of measurement at about 0.8 mm water equivalent taken to be at the inner of the inside surface of the chamber window was also used for this kind of measurements⁽²⁴⁾. The chamber is placed in a solid (PMMA) phantom with dimension 29.5 x 29.5 x 9.5 cm³. The ionisation chamber is connected through a constant voltage supply to -400 V and the signal is read through a digital electrometer in terms of charge integrated over a constant time of 60 seconds.

Measurements

The measurements were performed in the two following conditions: with field sizes of 5 x 5 cm² and 15 x 15 cm² with the mask on the top of phantom alternatively with the same field sizes but without the mask. The mask itself is a layer with 2 mm constant thickness with perforated holes 3

mm in diameter (POSIFIX-4, Posicast PR-3 precut). The gantry was always set to 0 degree and the collimator at 90 degree. In order to measure the depth dose in the presence and absence of the immobilization mask the different depths were simulated using slabs of polystyrene which are rather water equivalent with density of 1.06 g. cm^{-3} . A polystyrene phantom, 5 cm in thickness, was used to obtain back scattering. To ensure linearity of the TLDs the dose was limited to about 1 Gy at the dose maximum. For the measurements with the TLDs the relative depth dose, PDD, normalized to the dose at the dose maximum is obtained by as:

$$\text{PDD (\%)} = (D_a / D_m) \times 100 \dots (3)$$

where D_a is the mean readout corrected signal for individual sensitivity of the dosimeters irradiated at any depth and D_m the corresponding mean readout corrected signal at the depth of the maximum dose. In order to improve the accuracy of the results the experiments were repeated at least three times and the results are expressed as mean values with one standard deviation as an indication of the uncertainties involved. In the case of the plane parallel ionisation chamber the results are the mean of several repeated charge integrations all over the same constant 100 monitor unit for 6 MV and irradiation time of 60 seconds for ^{60}Co .

Results

The actual thickness of the polystyrene slabs, used in ^{60}Co measurement, measured with the accuracy of 0.1 mm are 1.0, 2.0, 3.1, 4.1, 5.1 mm and those used in the 6 MV beam are 3.0,

6.0, 9.0, 12.0, 15.0 mm. These thicknesses were converted to water equivalent depth by density correction and taking into account the effective point of measurement⁽²⁴⁾. For the TLDs, the effective point of measurement is located at the middle of each dosimeter corresponding to a water equivalent depth of 0.17 mm. In the plane parallel ionisation chamber (NACP 02) the effective point of measurement is located at 0.8 mm equivalent water located under the inner surface of the entrance window of the chamber. Since the measurements were performed in a constant source to dosimeter distance the results were further corrected for the difference in SSD by applying the inverse square law, as well as for the differences in the depth dose due to the change in SSD⁽³⁴⁾. Particularly for the plane parallel ionisation chamber no specific correction was applied for the temperature and pressure since both were constant during the experiments. The influence of the 2 mm thick mask to the doses at the surface and in the whole build up region, is shown in Fig. 1-2 for the TLD measurements and Fig. 3-4 for the plane parallel ionisation chamber measurements. Based on the TLDs measurements it is shown that the presence of the mask increases the surface dose in the ^{60}Co beam from 11% to 76% for the $5 \times 5 \text{ cm}^2$ field while for the $15 \times 15 \text{ cm}^2$ field the dose increases from 25% to 83% respectively. Beneath the surface, in the build-up region, the mask will also result in an increasing dose but the rate of increase is less step towards the maximum dose. The corresponding increases in the 6 MV bremsstrahlung beam are from 7% to 48% for the $5 \times 5 \text{ cm}^2$ field and from 18% to 56%

in the $15 \times 15 \text{ cm}^2$ field. The increase of the surface dose is due to the interaction of the primary and secondary beam components with the mask itself resulting in production of secondary electrons which then increase the surface dose⁽¹³⁻¹⁹⁾. At the same time the mask itself may slightly absorb contaminating electron from the treatment head and air⁽²⁹⁾ but this effect is minor compare to the influence of the mask. Obviously, the presence of the mask almost eliminates the skin sparing effect of megavoltage therapeutic beams. From the Fig. 3, 4 and 5, it is that in spite of the appropriate density corrections for the depth in the phantom and the effective point of measurement, the surface dose measured with the plane parallel ionisation chamber

is consistently higher than that measured by TLDs. For the $5 \times 5 \text{ cm}^2$ field size in the ^{60}Co beam the surface dose as measured by plane parallel ionisation chamber is in the absence of mask about 32% and increase to about 82% in the presence of the mask. Similarly, in the 6 MV bremsstrahlung beam and for the same field size the surface dose is correspondingly 25% and 56%. This effect is mainly due to the thick front wall and the perturbation effects of the plane parallel ionisation chamber as described by Nilsson and Montelius⁽²⁶⁾. It is thus shown that the plane parallel ionisation chamber can be used for relative measurements and surface doses if appropriate correction for the perturbations are applied^(21,26).

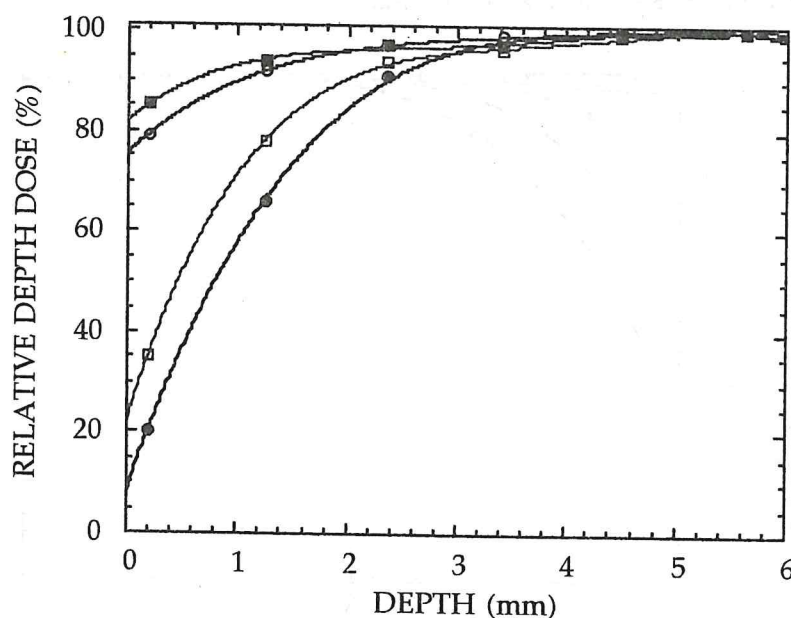


Fig.1. Relative depth dose curves of ^{60}Co measured by LiF TLDs for two different field sizes without respectively with the immobilization mask. Full circle: $5 \times 5 \text{ cm}^2$, without the mask. Open circle: $5 \times 5 \text{ cm}^2$, with the mask. Open rectangle: $15 \times 15 \text{ cm}^2$, without the mask. Full rectangle: $15 \times 15 \text{ cm}^2$, with the mask.

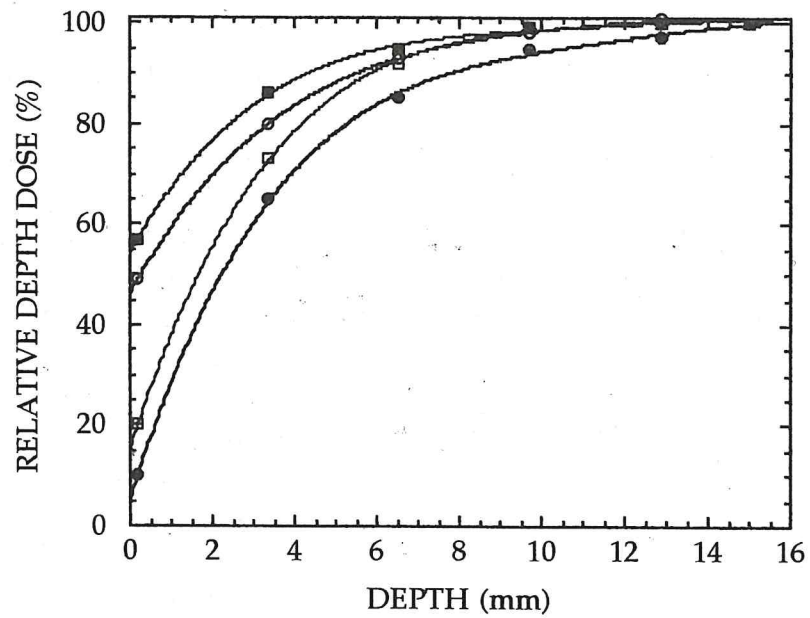


Fig.2. Relative depth dose curves of 6 MV measured by LiF TLDs for two different field sizes, without respectively with mask. Open circle: 5 x 5 mm², with the mask. Full circle: 5 x 5 mm², without the mask. Open rectangle: 15 x 15 mm² without the mask. Full rectangle: 15 x 15 mm² with the mask.

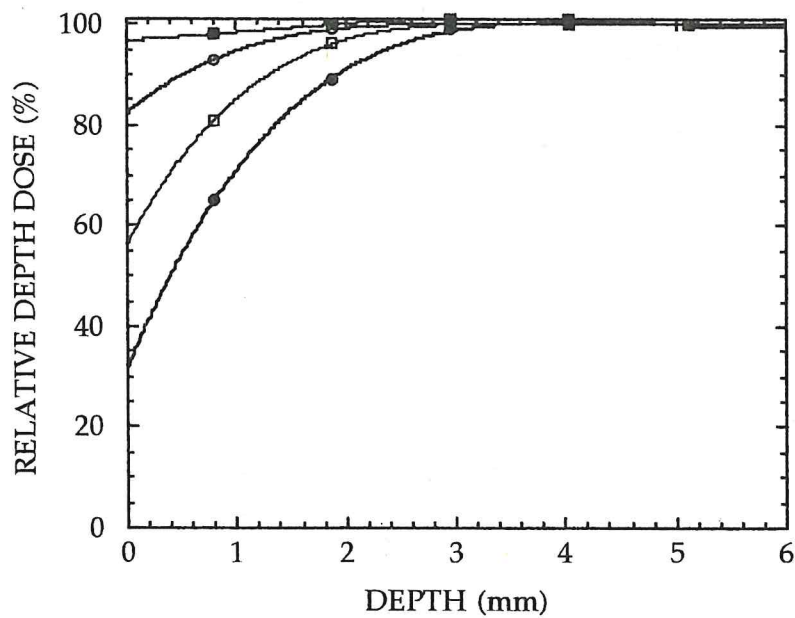


Fig.3. Relative depth dose comparison of ⁶⁰Co measured by the plane parallel ionisation chamber for two different field sizes without respectively with the mask. Open circle: 5 x 5 cm², with the mask. Full circle: 5 x 5 cm², without the mask. Full rectangle: 15 x 15 cm² with the mask. Open rectangle: 15 x 15 cm² without the mask.

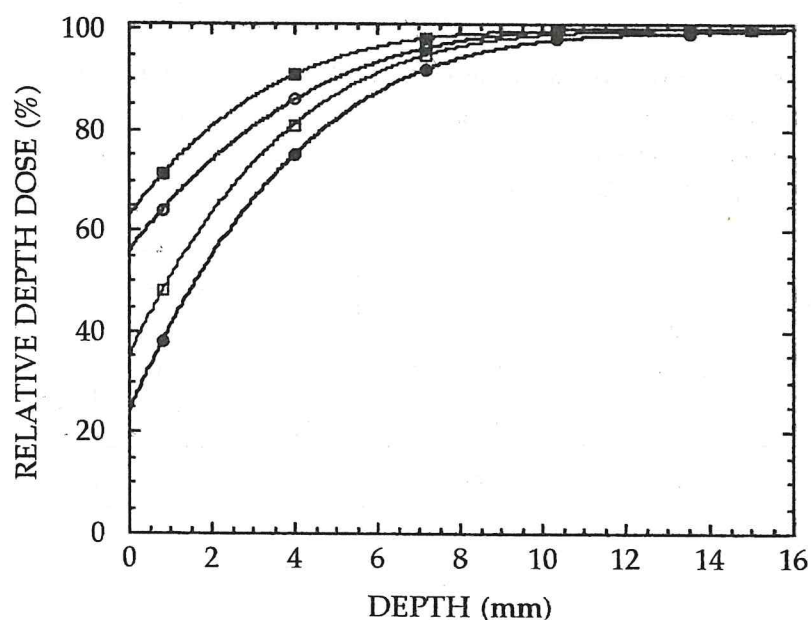


Fig.4. Relative depth dose comparison of 6 MV measured with the plane parallel ionisation chamber for two different field sizes without respectively with the mask. Open circle: 5 x 5 mm², with the mask. Full circle: 5 x 5 mm², without the mask. Open rectangle: 15 x 15 mm² without the mask. Full rectangle: 15 x 15 mm² with the mask.

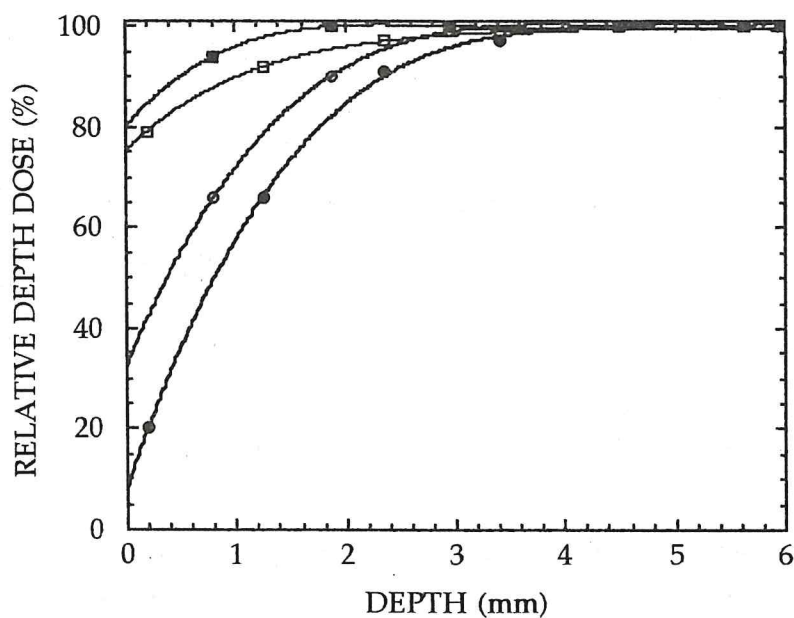


Fig.5. Comparison of depth dose curves between measured by LiF TLDs and the plane parallel ionisation chamber. Field size 5 x 5 cm², ⁶⁰Co beam. Full circle: without the mask, LiF TLDs. Open circle: without the mask, the plane parallel ionisation chamber. Open rectangle: with the mask, LiF TLDs. Full rectangle: with the mask, the plane parallel ionisation chamber.

Table 1. The percentage absorbed dose increase due to the influence of the mask for small fields and large fields of ^{60}Co beam measured by LiF TLDs and the plane parallel ionisation chamber. %wm: % relative depth dose, with the mask. %nm, % relative depth dose, without the mask.

depth (mm)	5x5 cm ² %wm-%nm (LiF)	15x15 cm ² %wm-%nm (LiF)	5x5 cm ² %wm-%nm (plane parallel)	15x15 cm ² %wm-%nm (plane parallel)
0	65	58	50	40
0.8	39	28	27	17
2	10	5	7	4
3	2	2	3	0
4	0	0	0	0
5	0	0	0	0

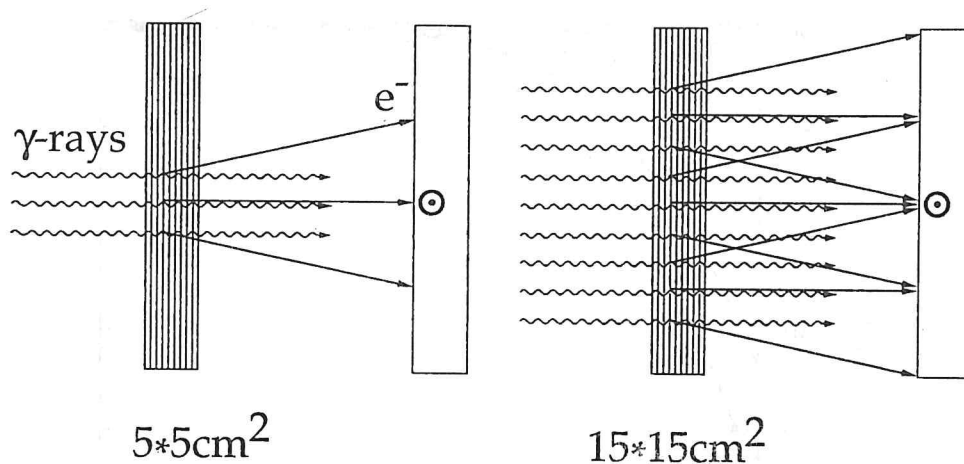


Fig.6. Schematic illustration of the generation of secondary electrons in the mask. The electrons that reach the dosimeter in the phantom are numerous for a broad beam than in a narrow beam and thus the dose is higher. The same mechanism is applicable in the presence of the immobilization mask. The mask will produce more electrons for low photon energies since the build up region is much shallower and the complete slowing down spectrum of secondary electron is achieved much faster.

Discussion

In radiotherapy, immobilization allows reproducible treatments thus increasing the possibility of delivering a precise and higher total dose to the tumour and decreasing the dose to the surrounding normal tissues^(11,12). However, the mask itself decreases the skin sparing effect since it increases the doses in the build up region thus delivering the highest dose to the surface and the subsequent tissues. This dose increase is mainly due to compton electrons set in motion by the primary photons but even due to the presence of beam contaminating electrons and photons. Since the electrons have a certain angular spread around the forward direction, the dose in a small field is generally lower than that in a large field as illustrated in Fig. 6. This process also explains the field size dependence in the absence of the mask since the air volume in front of the patient act as a source of electrons^(19,29). Ideally, when the patients need to be immobilized with a mask, it should be cut in the locations of the beam portals. On the other hand, if an increased skin dose is desired, particularly for superficial tumours, the mask can, at least partially, be used like a bolus to increase the dose. For 6 MV, the dose increase in the build up region due to the mask is less than in ^{60}Co , which is explained by the more forward directed electrons which in addition have longer pathlength due to their higher energy. In this experiment only one constant layer of the mask, 2 mm in thickness and with perforated holes of standard size of 3 mm in

diameter was used. Normally, when the mask is warmed up and individually fitted on a patient, the thickness may vary substantially over the whole surface. But the more thin mask, the less its immobilization capacity. Generally, the doses in the build up region will be altered depending on type of material and thickness of the mask, beam type, beam energy, and the angle of incidence of the beam^(13-19,34,35). The present data show that the mask will increase the dose equally to 1.65 mm of water in $5 \times 5 \text{ cm}^2$ field and 1.5 mm of water in $15 \times 15 \text{ cm}^2$ field in a ^{60}Co beam. On the other hand, in the 6 MV beam it will effect to the dose equally to 2 mm of water both in the $5 \times 5 \text{ cm}^2$ and $15 \times 15 \text{ cm}^2$ fields. In the present work it is proved that the precise determination of the percentage depth doses in the build up region is more accurate using extremely thin LiF TLDs⁽²²⁾. A plane parallel ionisation chamber (NACP 02) was also used since the thickness of the wall for this dosimeter was considered thin enough and the perturbation effects on the radiation beam relatively low. However the difference in the measured surface dose between the two types of dosimeters is 21% in the ^{60}Co beam. Thus it is shown that this plane parallel ionisation chamber substantially overestimates the surface dose. The data are very important. They should be obtained to consider in dose planning for optimization in order to keep the radiation induced reactions to normal tissues at a low level and to improve the quality of life of the patients.

Conclusion

The mask will alter the build up curve, the slope of the curve will be changed, the doses in the build up region tend to increase to maximum dose. The affect on the dose distribution in low energy beams are higher than in high energy beams. The percentage of the increase dose at the surface is the highest and under skin will be also increased but the percentage of absorbed dose increase reducing as a function of depth up to maximum dose the influence of the mask is not shown.(Table 1) When using the mask in low energy beams, the mask should be cut off to maintain skin sparing, but it can be left in the field when the tumour extends to the surface. In higher energy beams the effect of the mask in build up region is lower, the curve shown less steep than of ^{60}Co . In larger fields, the doses in the build up region will be higher than in smaller fields, the rate of increase absorbed dose by the mask in small fields is higher than in large field.

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References

- 1) Brahme A. Optimization of radiation of radiation therapy for cancer patients. FORUM, Trends in Experimental and Clinical Medicine 1994 ; 4 : 569-584.
- 2) Beauvais H, Bridier A, Dutreix A. Characteristic of contamination electron in high energy photon beams. Radiother Oncol 1993 ; 29 : 308-316.
- 3) Brahme A. Dosimetric precision requirement in radiation therapy. Acta Radiol Oncol 1984 ; 23: 379-391
- 4) Brahme A. Multileaf collimation of neutron, photon and electron beams Proc 3rd Ann ESTRO meeting, Jerusalem 1984 : 218.
- 5) Brahme A. Optimization of radiation therapy and the development of multileaf collimation. Int J Radiat Oncol Biol Phys 1993 ; 25 : 373-375.
- 6) Brahme A. Optimization of stationary and moving beam radiotherapy techniques. Radiotherapy Oncol 1988 ; 12 : 129-140.
- 7) Brahme A. Treatment optimization using physical and radiobiological objective functions. In : Medical Radiology Radiation Therapy Physics. Editor : A. R. Smith. Springer Verlag, Berlin 1995 : 209-246.
- 8) Spiridon VS, Chui CS. Generation of arbitrary intensity profiles by combining the scanning beam with dynamic multileaf collimation. Med Phys 1996 ; 23 : 1-8.
- 9) Webb S. Optimization by simulated annealing of three - dimensional, conformal treatment planning for radiation fields defined by a multileaf collimator : II. Inclusion of two-dimensional modulation of the x-ray intensity. Phys Med Biol 1992 ; 37 : 1689-1704.
- 10) Webb S. Optimised Three-Dimensional Treatment Planning for Volumes with Concave Outlines, Using a multileaf Collimator. Advance Radiation Therapy Tumour Response Monitoring and Treatment Planning. Springer-Ver, Berlin Heidelberg 1992 : 495-501.
- 11) Barish RJ, Lerch LA. Patient immobilization with a low temperature splint/brace material. Radiology 1978 : 127 : 548.
- 12) Bentel CG. Immobilization of patients undergoing Radiation therapy for head and neck malignancies. Alpha Cradle® patient repositioning system field guide. Smithers Medical Products, Inc: 182.
- 13) Ciesielski B, Reinstein L, Weilopolski L, Meek A. Dose enhancement in build-up region by lead, aluminum, and lucite absorbers for 15 MV photon beam. Med Phys 1989 ; 16 : 609-613.
- 14) Caring SER, Windeyer BY, Smither DW. British practice in Radiotherapy. Buterworth 1955 : 3-65.
- 15) Khan FM, Moore VC, Levitt SH. Effect of various atomic number absorbers on skin dose for 10-MeV x-rays. Radiology 1973 ; 109 : 209-212.
- 16) Nilsson B. Electron contamination from different materials in high energy photon beams. PMB 1985; 30: 139-151.
- 17) Thilmann C, Adamietz IA, et al. Increase of surface dose using wound dressings during percutaneous radiotherapy with photon and electrons. Radiother Oncol 1996 ; 40 : 181-184.

- 18) Nilsson B, Brahme A. Absorbed Dose from Secondary Electrons in High Energy Photon Beams. *Phys Med Biol* 1979 ; 5:901-912.
- 19) Nilsson B, Brahme A. Contamination of high-energy photon beams by scattered photons. *Trahlentherapie* 1981 ; 157 : 181- 186.
- 20) Kron T, Metcalfe P, Wong T. Thermoluminescence dosimetry of therapeutic x-rays with LiF ribbons and rods. *Phys Med Biol* 1993 ; 38 : 833-845.
- 21) Wittkomper FW, et al. Perturbation correction factors for some ionisation chambers commonly applied in electron beams. *Phys Med Biol* 1991 ; 36 : 1639-1652.
- 22) Nilsson B, Sorcini B. Surface dose measurements in clinical photon beams. *Acta Oncol* 1989 ; 28 : 537-542.
- 23) Rocha MPO. The effect of build up cap materials on the response of an ionisation chamber to ⁶⁰Co gamma rays. *Phys Med Biol* 1993 ; 38 : 793-804.
- 24) Technical Reports series No. 277 . Absorbed dose determination in photon and electron beams. An international code of practice. International Atomic Energy Agency, Vienna; 1978.
- 25) ICRU report 37. International Commission of Radiation units and measurements. Stopping Powers for Electrons and Positrons 1984.
- 26) Nilsson B, Montelius A. Fluence perturbation in Photon beams under non equilibrium conditions. *Med Phys* 1986 ; 13 : 191-195.
- 27) McKinlay AF. Thermoluminescence dosimetry. *Medical Physics Handbooks* 5 1981 ; 32, 103, 127, 132.
- 28) Joelsson I, Rud'en BI, et al. Determination of dose distribution in the pelvis by measurement and by computer in gynaecology radiation therapy. *Acta Radiologica* 1972 ; 11 : 292.
- 29) Nilsson B, Brahme A. Electron contamination from photon beam collimators. *Radiother Oncol* 1986; 5 : 235 - 244.
- 30) Christ G, Major J, et al. Dose the photon quality factor depend on the type of ionization chamber ? *Radiol Oncol* 1995 ; 36 : 153-155.
- 31) Mitie C, Leunes, GV, et al. Is it necessary to repeat quality control procedures for head and neck patients? *Radiother Oncol* 1991 ; 21 : 201-210.
- 32) Khatb EIE. Conversion of ionization measurements to radiation absorbed dose in non-water density material. *Phys Med Biol* 1992 ; 37: 2083-2094.
- 33) Gold S, Chmidt H, Sherwin WK. Reaction to ionizing radiation. *J Am Acad Demtol* 1960; 3 : 551-579.
- 34) Sjogren R, Karlsson M. Electron contamination in clinic high energy photon beams. *Med Phys* 1996 ; 23 : 1873-1881.
- 35) Ostwad PM, Kron T. Surface dosmeasurement for highly oblique electron beams. *Med Phys* 1996 ; 23 : 1413-1419.