DOSIMETRIC CHARACTERISTICS OF CIRCULAR 6-MeV X-RAY BEAMS FOR STEREOTACTIC RADIOTHERAPY WITH A LINEAR ACCELERATOR*

A. Wysocka

The Andrzej Sołtan Institute for Nuclear Studies
Department of Accelerator Physics
05-400 Otwock-Swierk, Poland
email: wysocka@ipj.gov.pl

J. Rostkowska, M. Kania, W. Bulski, and J. Fijuth

Maria Skłodowska-Curie Memorial Cancer Centre and Institute of Oncology
Roentgena 5, 02-781 Warszawa, Poland

(Received December 7, 1999)

Dosimetric characteristics of 6 MeV circular X-ray beams of diameters ranging from 7.5 to 35.0 mm are reported. The 6-MeV X-ray beam from Clinac 2300CD was formed using additional cylindrical BrainLAB's collimators. The mechanical stability of the entire system was verified. Specific quantities measured include tissue maximum ratios (TMR), beam profiles (off-axis ratios OAR) and relative output factors. Measurements of these parameters were performed in a water phantom using small cylindrical ionization chambers and a diamond detector. Comparison of TMR values measured with the ionization chamber and the diamond detector showed no significant differences. It was shown that the latter yields more accurate results for beam profiles than ionization chambers. The mechanical and dosimetric characteristics of this radiotherapy unit are found to be suitable for stereotactic radiosurgery and radiotherapy.

PACS numbers: 87.53.Dq, 87.53.Ly

1. Introduction

Stereotactic radiosurgery (SRS) is a method of treatment of small intracranial lesions such as: arteriovenous malformations, benign or malignant primary tumours and isolated metastases, using external beams of radiation.

* Presented at the NATO Advanced Research Workshop, Krzyże, Poland 2–4 September 1999.

In this technique a high dose of radiation is delivered to the stereotactically localized lesion, while minimally irradiating adjacent normal brain tissue.

Radiosurgery can be performed either with gamma beams from multi-mini cobalt sources and heavy charged particle beams from cyclotrons, or with narrow photon beams from isocentric linear electron accelerators [1–6]. Stereotactic radiotherapy (SRT) combines the target and dose localization characteristics of SRS with the biological advantages of dose fractionation [7]. Clinical results are excellent [8]. The first clinical attempts of the linac-based SRT in Poland are planned to be carried out in 1999 in Institute of Oncology in Warsaw.

The concept of stereotaxis is to localize accurately in space any desired region of the brain. The stereotactic frame "locks" the patient's head into a fixed position with respect to a coordinate system related to the imaging and treatment machines [9]. Computerized tomography (CT), magnetic resonance imaging (MRI) [10] and digital subtraction angiography (DSA) [11] are used as imaging techniques.

Once the target volume is defined a dose planning can be performed. A treatment planning system for radiosurgery takes into account a large number of irradiation parameters such as the collimator diameters, number of arcs, their angular positions and angular lengths and the weight of the arcs, in order to obtain distributions appropriately adapted to each individual treatment plan [12]. Very precise dosimetric data form the input data for treatment planning code.

Once a central point of the tumour is located at the isocentre of the accelerator a whole treatment is performed by combining rotation of an accelerator with a sequential number of patient couch positions.

The quality of the radiosurgical treatments depends on two fundamental parameters:

- mechanical tolerances due to gantry and couch rotations, and
- the dose distribution and the actual dose delivered by a therapy unit with narrow photon beams.

These two mechanical and dosimetric characteristics of the therapy unit have been investigated by the authors in this work.

2. Materials and methods

2.1. Accelerator and collimation

The 6 MeV X-ray beam from Clinac 2300CD (Varian,USA) was formed using additional cylindrical BrainLAB's collimators. Nominal field diameters at the isocentre range between 0.75 cm and 3.50 cm in steps of

0.25 cm. The rectangular collimator of the linac was set to a field of 5×5 cm^2 at isocentre. Radiosurgical technique involves rotation of the accelerator gantry and the treatment coach. Therefore, the three principal mechanical axes of the gantry, the turntable and the collimator should intersect at a common point (the isocentre) and remain stable during all rotation.

Verification of alignment for Clinac 2300CD accelerator has been performed with Winston–Lutz test [13]. A small, 3mm, tungsten ball was positioned at the radiation isocentre and radiographic exposure was made in each of eight standardized positions of linac gantry and turntable. The ball was centred within acceptable limits (0.5 mm) on all exposures, so we concluded that alignment was correct.

2.2. Detectors

Accurate dosimetry of small-field photon beams used in stereotactic radiosurgery (SRS) and radiotherapy (SRT) is difficult because of the presence of lateral electronic disequilibrium and steep dose gradients. The detectors used for measurements of absorbed dose distribution must be small in respect to the size of radiation field and must have a sufficient spatial resolution. Therefore small volume ion chamber, diode, diamond detector and film are proposed for that purpose [14–18].

For small fields, film is often used as a dosimeter of choice [9,15,16] due to its high degree of spatial resolution, limited generally by the size of the aperture of the optical densitometer. However, film dosimetry is often difficult due to variability in optical density associated with exposure conditions, film orientation, processing and reading.

The application of diamond detector in SRS relative dosimetry has been presented by the authors of [9,17,18]. A commercially available PTW diamond detector was used in this study and evaluated against ion chambers. The effective measuring volume of the diamond detector is 3 mm in diameter and 0.3 mm in thickness.

The inner diameters of a 0.125 cm^3 thimble chamber and a 0.015 cm^3 thimble chamber are 5.5 mm and 1.8 mm, respectively. The latter was used for the measurements presented here. The detailed description and intercomparison of the penumbra width measurements performed with two small volume ion chambers and the diamond detector will be presented by the authors elsewhere.

A PTW-MP3 water-scanning system was used with a 0.015 cm^3 PTW ion chamber and a PTW diamond detector. The orientation of the ion chamber axis and the sensitive thickness of the diamond detector were parallel to the beam axis. TMR and OAR measurements were performed at Source Surface (of phantom) Distance SSD equal 92.5 cm and depth in the phantom

d equal 7.5cm. Since most cranial tumours are within a range of 7.5 cm, the OAR at depth 7.5 cm was entered in the treatment planning system for dose calculation. ROF values were measured at SSD = 98.5 cm and $d = d_{\max} = 1.5$ cm.

3. Results

The three following characteristics of the small fields required by the treatment planning system BrainSCAN were measured for each stereotactic collimator, for diameters ranging from 0.75 to 3.50 cm at the isocentre:

— tissue maximum ratio (TMR),

— off axis ratios (OAR) and

— relative output factors (ROF).

3.1. TMR

Tissue Maximum Ratio (TMR) is defined as:

$$\mathrm{TMR}(c,d) = \frac{D(c,0,d)}{D(c,0,d_{\max})}\,, \tag{3.1}$$

where d is depth in the phantom, $d_{\max}$ is reference depth, c is collimator diameter. The $d_{\max}$ was treated at a constant depth of 1.5 cm within the limits of experimental accuracy. The TMR values measured with the diamond detector and the small volume (0.015 cm^3) ion chamber are within 0.6 % of each other. Figure 1 shows TMR as a function of depth for collimator diameter equal 0.75 cm and 3.50 cm, measured with ion chamber.

3.2. OAR

Off Axis Ratio (OAR) defined as:

$$\mathrm{OAR}(c,r,d) = \frac{D(cr,d)}{D(c,0,d)} \tag{3.2}$$

is the ratio of the dose measured at a radial distance r relative to the dose at the central axis for a collimator diameter c. The diamond detector should be used to provide an accurate profile for small fields due to its unique radiation properties [19]. The detailed comparison of the OAR measurements (profiles) with small volume ion chamber and diamond detector for the circular collimators will be presented elsewhere. Measurement results for beam profiles performed with this detector are presented in figure 2.

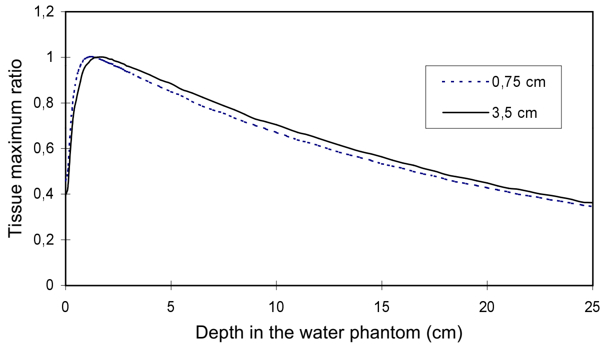

Fig. 1. Tissue maximum ratio (TMR) of 6 MeV X-ray beam, measured for 0.75 cm and 3.50 cm collimator diameters using 0.015 cm^3 ion chamber.

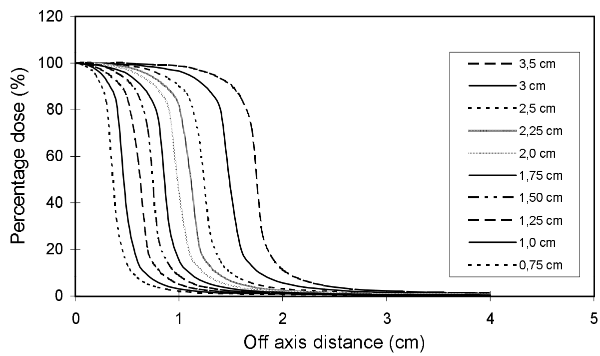

Fig. 2. Beam profiles (OAR) of 6 MeV X-ray beam formed with collimators of 0.75 cm to 3.00 cm in diameter, measured with diamond detector.

3.3. ROF

Relative Output factor ROF is defined by:

$$\mathrm{ROF}(c) = \frac{D(c, 0, d_{\max})}{D(10 \times 10\mathrm{cm}^2, 0, d_{\mathrm{cal}})} \tag{3.3}$$

and is the ratio of the dose at a depth $d_{\max}$ on the central axis for a collimator diameter c relative to the dose measured at the same point in a standard 10×10 cm^2 calibration field. Output factor measurements were made along the central axis of the beam at the isocentre, using ion chamber. From performed investigation (detailed values will presented elsewhere) we could see that ROF for the circular collimator is the function of primary jaws field size setting. Figure 3 shows the ROF values for the beam formed with the representative cilcular collimators and with fixed primary jaws field size setting: 5×5 cm^2.

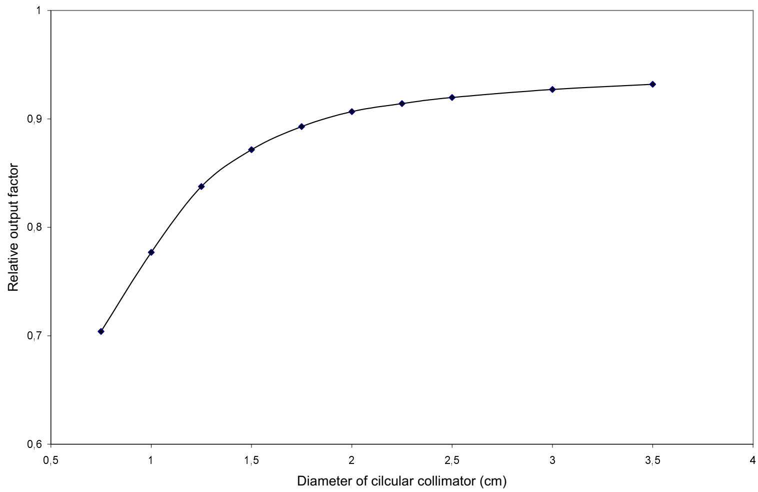

Fig. 3. Relative output factors (ROF) of 6 MeV X-ray beam formed with collimators of 0.75 cm to 3.0 cm in diameter and with primary jaws field size setting $5 \times 5\text{cm}^2$, measured using ion chamber.

4. Conclusions

Dosimetric parameters TMR, OAR, ROF are presented based on the measurements with diamond detectors and ion chambers. Comparison of TMR values measured with small ionization chamber (0.015 cm^3) and diamond detector showed no significant differences. The diamond detector was found to be more appropriate choice for beam profile measurements than the ionization chambers and yields more accurate results. Collimator output factors can be measured with ionization chamber with sensitive volume 0.015 cm^3. Relative output factor (ROF) for the circular collimator is the function of the primary jaws field size setting. We suggest that measurements of TMR, OAR and ROF for all circular stereotactic collimators should be made with a fixed primary jaws field size setting: $5 \times 5\ \text{cm}^2$.

The mechanical and dosimetric characteristics of radiotherapy unit (Clinac 2 300 CD with BrainLAB's collimators) are found to be suitable for stereotactic radiosurgery and radiotherapy.

REFERENCES

[1] O.O. Betti, V.E. Derechinsky. *Acta Neurochirurgica* **12**, 385 (1984).

[2] F. Colombo, A. Benedetti, F. Pozza, R.C. Avanzo, C. Marchetti, G. Chierego, A. Zanardo, *Neurosurgery* **16**, 154 (1985).

[3] G.H. Hartmann, W. Schlegel, V. Sturm, B. Kober, O. Pastyr, W.J. Lorenz, *Int. J. Rad. Oncol. Biol. Phys.* **11**, 1185 (1985).

[4] W. Lutz, K.R. Winston, N. Maleki, *Int. J. Radiat. Oncol. Biol. Phys.* **14**, 373 (1988).

[5] P.H. McGinley, E.K. Butker, J.C. Landry, *Phys. Med. Biol.* **35**, 649 (1990).

[6] E.B. Podgorsak, A. Oliver, M. Pla, P. Lefebvre, J. Hazel, *Int. J. Radiat. Oncol. Biol. Phys.* **14**, 115 (1988).

[7] R.W. Laing, A.P. Warrington, J. Graham, J. Britton, F. Hines, M. Brada, *Radiother. Oncol.* **27**, 22 (1993).

[8] J. Fijuth, L. Kepka, *Nowotwory* **48**, (1999), in press (in Polish).

[9] A. Wysocka, *Nukleonika*, **41**, 11 (1996).

[10] M.H. Philips, M. Kessler, F.Y.S. Chuang, K. Frankel, J.T. Lyman, J.L. Fabrikant, R.P. Levy, *Int. J. Raiat. Oncol. Biol. Phys.* **20**, 881 (1991).

[11] F.J. Bova, W.A. Friedman, *Int. J. Raiat. Oncol. Biol. Phys.* **20**, 891 (1991).

[12] A. Wysocka, *Rep. Practical Oncology* **3**, 59 (1998).

[13] K.R. Winston, W. Lutz, *Neurosurgery* **22**, 454 (1988).

[14] G. Chierego, P. Francescon, F. Colombo, F. Pozza, *Radiother. Oncol.* **29**, 214 (1993).

[15] Ch.F. Serago et al., *Phys. Med. Biol.* **37**, 1943 (1992).

[16] B.E. Bjarngard, J.S. Tsai, R.K. Rice, *Med. Phys.* **17**, 794, (1990).

[17] P.W. Hoban, M. Heydarian, W.A. Beckham, A.H. Beddoe, *Phys. Med. Biol.* **39**, 1219 (1994).

[18] I.J. Das et al., *Radiother. Oncol.* **38**, 61 (1996).

[19] S.N. Rustgi, *Med. Phys.* **22**, 567 (1995).