

## Kertas Asli/Original Articles

# Measurement of Dosimetric Parameters and Dose Verification in Stereotactic Radiosurgery (SRS) (Pengukuran Parameter Dosimetri dan Pengesahan Dos dalam Radiosurgeri Stereotaktik)

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### ABSTRACT

*The first part of this study was about measurement of dosimetric parameters for small photon beams to be used as input data for treatment planning computer system (TPS) and to verify the dose calculated by TPS in Stereotactic Radiosurgery (SRS) procedure. The beam data required were percentage depth dose (PDD), off-axis ratio (OAR) and scattering factor. Small beams of 5 mm to 45 mm diameter from a circular cone collimator in SRS were used for beam data measurements. Measurements were made using pinpoint ionisation chamber (0.016cc). In the second part of this study, we reported the importance of carrying out quality assurance (QA) procedures before SRS treatment which were found to influence the accuracy of dose delivery. These QA procedures consisted of measurements on the accuracy in target localization and treatment room laser alignment. The calculated TPS dose for treatment was verified using pinpoint ionisation chamber and thermoluminescent detector (TLD) 100H. The deviation mean between measured and calculated dose was -3.28%. The measured dose obtained from pinpoint ionisation chamber is in good agreement with the calculated dose from TPS with deviation mean of 2.17%. In conclusion, pinpoint ionisation chamber gives a better accuracy in dose calculation compared to TLD 100H. The results are acceptable as recommended by International Commission on Radiation Units and Measurements (ICRU) Report No. 50 (1994) that dose delivered to the target volume must be within  $\pm 5\%$  error.*

*Keywords: Stereotactic radiosurgery (SRS); treatment planning system (TPS); pinpoint ionisation chamber; thermoluminescent detector (TLD); dose verification*

### ABSTRAK

*Bahagian pertama kajian ini adalah mengenai pengukuran parameter dosimetri bagi bim foton halus untuk digunakan sebagai data input bagi sistem perancangan rawatan berkomputer (TPS) dan untuk mengesahkan dos yang dikira oleh TPS dalam prosedur radiosurgeri stereotaktik (SRS). Data bim yang diperlukan adalah peratus dos kedalaman (PDD), nisbah paksi terluar (OAR) dan faktor penyerakan. Bim halus berdiameter 5 mm hingga 45 mm daripada kolimator kon membulat dalam SRS digunakan bagi pengukuran data bim. Pengukuran dilakukan menggunakan kebuk pengionan bersaiz kecil (0.016cc). Dalam bahagian kedua kajian ini, kami laporkan kepentingan melaksanakan prosedur jaminan kualiti (QA) sebelum rawatan SRS yang didapati mempengaruhi ketepatan pemberian dos. Prosedur QA ini terdiri daripada pengukuran ketepatan dalam penyetempatan sasaran dan penjajaran laser bilik rawatan. Dos TPS yang dikira untuk rawatan disahkan menggunakan kebuk pengionan kecil dan pengesan termoluminescent (TLD) 100H. Purata sisihan antara dos pengukuran dan pengiraan adalah -3.28%. Dos pengukuran yang diperolehi daripada kebuk pengionan kecil bersetuju dengan dos yang dikira oleh TPS dengan purata sisihan 2.17%. Sebagai kesimpulan, kebuk pengionan bersaiz kecil memberikan ketepatan yang lebih baik dalam pengiraan dos berbanding TLD 100H. Keputusannya boleh diterima seperti yang disyorkan oleh Suruhanjaya Antarabangsa Mengenai Unit Sinaran dan Pengukuran (ICRU) Laporan No. 50 (1994), bahawa dos yang diberikan kepada isipadu sasaran mesti berada dalam ketakpastian  $\pm 5\%$ .*

*Kata kunci: Radiosurgeri stereotaktik (SRS); sistem perancangan rawatan (TPS); kebuk pengionan kecil; pengesan termoluminescent (TLD); pengesahan dos*

### INTRODUCTION

For installing and commissioning a new SRS treatment planning computer, comprehensive set of beam data of the linear accelerator (linac) machine must be measured and downloaded into the treatment planning software. Three basic beam parameters to be measured for dose

calculation are percentage depth dose (PDD), field size scatter factors and single beam profiles. There were two crucial components in stereotactic radiosurgery (SRS) in order to achieve the aims of treatment planning. The first component is to deliver precise and uniform dose to the target lesion with multiple noncoplanar beams. The second factor is the accuracy in positioning of patient; this can be

achieved by introducing the stereotactic apparatus. With combination of these two factors, SRS can provide uniform dose and sharp dose falloff for small lesion and at the same time produce minimum damage to the surrounding critical organs such as the brain tissue (Khan et al. 2011).

The geometric position of the isocenter during rotations is usually assumed to be inside a virtual spherical volume. Minimizing the isocenter movement could improve the accuracy of SRS treatment and this issue has been considered with special intention. The AAPM Task Group Report 142 (2009) recommends up to  $\pm 1$  mm deviation between the radiation and mechanical isocenter is acceptable for SRS treatments. This recommendation made because the result of study has been reported that 2 mm positioning error in spine SRS could lead to more than 5% loss of tumour coverage and more than 35% increase in dose delivery to the healthy tissues. With an accuracy of 1 mm of SRS treatment, the error in dose delivery was reduced to less than 2%. It must note that errors in SRS treatment delivery are nonrecoverable, since the treatment is delivered in one session. The isocenter verification process is compulsory to be performed before each SRS treatment (Rowshanfarzad et al. 2011).

Although dose calculation can be carried out using dosimetric parameter of beam data measurements, the method of beam data is collected presents challenges to the physicist. In particular, narrow radiosurgery beams with sharp dose gradient require small volume of detectors and chambers with high spatial resolution. For central axis measurement such as depth dose, tissue phantom ratio, field size scatter factor, the detector dimensions should be significantly smaller than the field sizes. Special care is required when selecting and handling the required dosimetry equipment. For small field sizes, it is particularly important to correctly align the water phantom and the detector movement direction in relation to the beam axis and the beam center. Even if the detector size is suitable for the small fields to be measured, accurate sensitivity correction which is related to the energy dependency of the detector signal must be applied in accordance with the specifications provided by manufacturer of the detector.

Particular attention may be paid to tests for treatment planning systems (TPSS) that deal with specialized techniques such as stereotactic radiosurgery. Absolute dose verification entails a rigorous comparison between measured dose using detectors and dose produced by TPS based on the input data (measured beam data). In order to determine the best dosimetry system for measurements at the small focal point of beams, several different detectors were investigated such as an ionisation chamber and LiF thermoluminescent dosimeter chips. All of these detectors were calibrated in a phantom against an ionisation chamber whose calibration is traceable to secondary standard dosimetry laboratory (SSDL), Agency Nuclear Malaysia. Ionisation chamber measurement of these small beams and steep dose gradients often suffer from lack of lateral equilibrium and chamber volume effect. Due to the general

uncertainty and difficulty in these measurements, measured data should be verified using two or more dosimetric methods (AAPM Report No. 47 1994). It is important for practicing medical physicist to be able to quantify errors involved in the use of particular TPS in clinical use at his or her department. In this study we chose to verify the absolute dose from measurement with different detectors compared to calculated dose from SRS treatment planning system before it can be clinically used.

## MATERIALS AND METHODS

### BEAM DATA MEASUREMENTS

Percentage depth dose (PDD) establishes the depth variation along the central axis of the beam. Measurement was being done for 6 MV photon energy generated by linear accelerator Primus (Siemens Medical, USA) using pinpoint ionisation chamber (PTW, Freiburg, Germany). Circular cone with 45 mm diameter in size was slotted into the cylindrical housing that being mounted to the head of linear accelerator. Pinpoint ionisation chamber was inserted to the chamber holder and placed perpendicular to radiation beam; 105 cm distance from the source. The PDD of 45 mm diameter cone measured from 0 cm depth to the desired depth which is 300 mm depth inside the water. Mephysto mc<sup>2</sup> software (PTW, Freiburg, Germany) was used to analyze the readings and plotted the PDD. The circular cone of 45 mm diameter then replaced with other circular cones that available consequently and the procedure was repeated for each diameter.

Off-axis Ratio (OAR) measurement setup for 5mm to 45 mm diameter cone was slightly similar as PDD measurement but for OAR, pinpoint ionisation chamber moved perpendicularly to the beam central axis at measurement depth 7.5 cm (Figure 2). For field size scatter factor measurement, the gantry and collimator position was being set to zero degree while source to surface distance (SSD) was set at 100 cm. Then, the chamber was placed 5 cm beneath water phantom surface. The irradiation was repeated three times to get the average value. The circular cone of 45 mm diameter then replaced with other circular cones that available consequently and the procedure was repeated for each diameter. The scatter factor for each circular cones used were calculated by normalizing the average of charge measured to average of charge of obtained at reference field size  $10 \times 10$  cm<sup>2</sup>.

### QUALITY ASSURANCE (QA) PROCEDURES BEFORE SRS TREATMENT

Brown-Roberts-Wells (BRW) frame was placed on the radiosurgery head phantom. The Brainlab computed tomography (CT) - localizer attached to BRW frame and placed on the couch of Light Speed Qx/i (GE,USA) CT scanner. The phantom and the frame were extended beyond the CT couch. 2.4 mm slices thickness used for

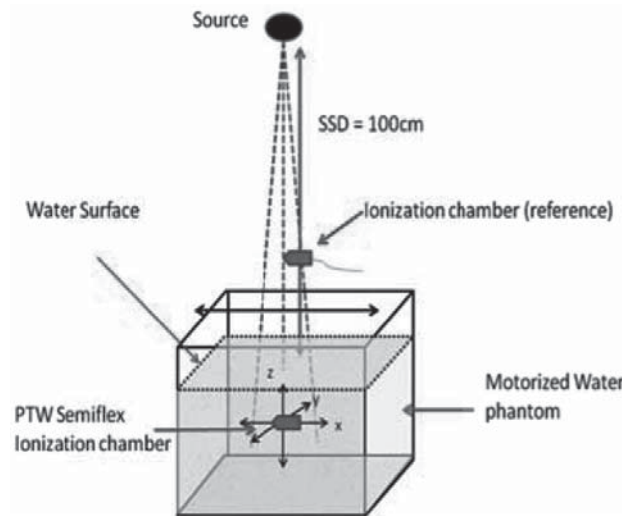


FIGURE 1. The experimental set up of PDD and OAR measurements

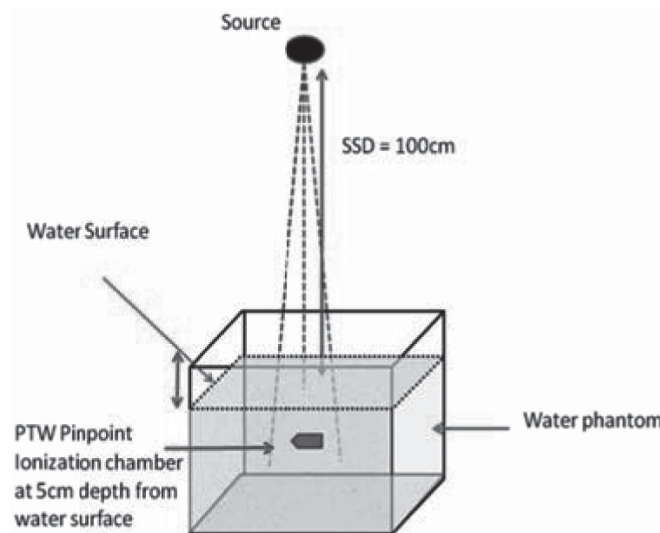


FIGURE 2. The experimental set up of scatter factor measurement

scanning and the CT images were transferred to i-Plan (Brainlab, Germany) treatment planning workstation through digital imaging and communication in medicine (DICOM) transfer. By using iPlan RT Image 4.1, we contoured the skull as external contour and each of the four objects. The autocenter function was used to position the isocenter to each of the structures. Then, by moving the isocenter superiorly until it reached the top of each structure the coordinates of the top centers of these objects were determined and compared with the vendor-provided values. The accuracy of localization can be evaluated using the total localization errors, using the formula  $r = ((\Delta AP)^2 + (\Delta Lat.)^2 + (\Delta Vert.)^2)^{1/2}$ ,  $\Delta AP$ ,  $\Delta Lat.$ ,  $\Delta Vert$  represent the components of errors in AP, lateral and

vertical directions respectively.  $\Delta r$  represents the total localization error.

Apart from conventional fixed radiotherapy, it is very important to confirm the accuracy of the isocenter, as SRS treatment uses the rotation of gantry and couch. To do this, mechanical isocenter standard (MIS; Radionics Inc., USA), rectilinear phantom pointer (RLPP; Radionics Inc., USA), and laser target localizer frame (LTLF; Radionics Inc., USA) were used. The MIS aligns the wall and ceiling lasers; therefore, the accuracy of MIS was checked by films verification before adjustment of these lasers. The accuracy of laser alignment was verified using RLPP. A verification film was irradiated by 6 MV photon beam through a lead ball attached to RLPP for  $0^\circ$ ,  $90^\circ$  and  $270^\circ$  gantry angle and the couch fixed at  $0^\circ$

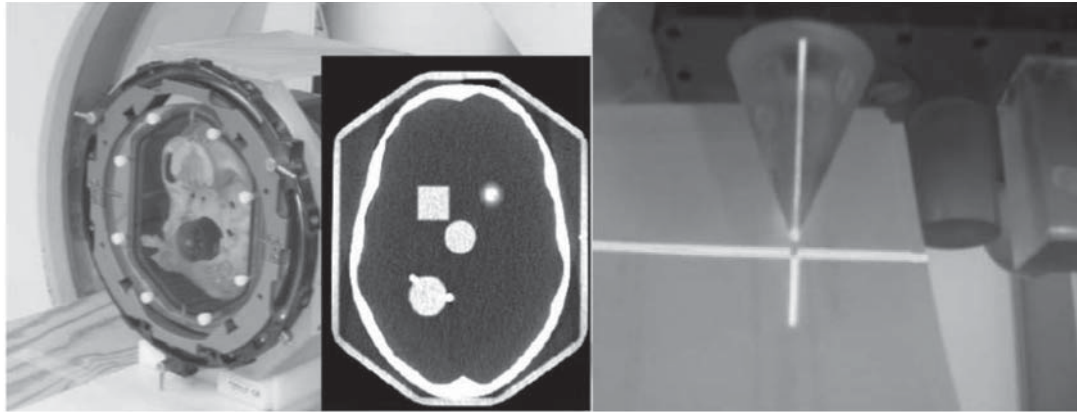


FIGURE 3. Radiosurgery head phantom (Radionic Inc.,USA) setup

angle. The X-Omat therapy verification film (Kodak, USA) was used for verification. This technique was introduced by Lutz, Winston and Maleki at Harvard Medical School in 1998 (Rowshanfarzad et al. 2011). This test is relatively simple conducted before every SRS procedure.

#### DOSE VERIFICATION

The intensity modulated radiotherapy (IMRT) Thorax phantom (CNMC, USA) was scanned with the same parameter as radiosurgery head phantom. The CT scanning was done with ionisation chamber inserted inside the phantom. CT images of thorax phantom were contoured and the active volume of ionisation chamber was assigned as target volume in Phantom Mapping Software (Brainlab, Germany). From that, the previous treatment plan based on the radiosurgery head phantom was loaded into the verification plan (thorax phantom). The isocenter position could be assigned at the center of the active volume of ionisation chamber. After that, the IMRT Thorax phantom was positioned on the treatment couch accordingly based on room's laser alignment.

For the thermoluminescent detector (TLD) irradiation, the IMRT Thorax phantom could not be used because the phantom only has insert for ionisation chamber and not for TLD chips. It is impossible to put TLD chips inside the phantom. As a substitute, solid water phantom with TLD insert was used. The assumption is made that dose calculated base on IMRT Thorax phantom give same value as inside the solid water phantom due to both phantoms were basically made from same materials. The irradiation of TLD chips were done at same depth based on treatment plan on the IMRT Thorax phantom. 4 TLD chips were used for every irradiation and the TLD chips were positioned totally inside the irradiation field. After 24 hours of irradiation, the TLD chips were readout using TLD reader with a reading profile of: preheat temperature of 135°C for 10 second, acquire temperature rate 10°C/s,

acquire maximum temperature of 240°C for 23.33 second and annealing temperature of 240°C for 10 second. The results directly converted from charge to absorbed dose by the reader. Previously, the TLD chips and reader was calibrated. The TLD reader was assigned a reference correction factor (RCF) factor for batch of TLD 100H chips. The RCF was 0.4258 nC/μGy. Each chip was also assigned ECC values. The reproducibility of TLDs evaluated in three time exposures to be within 5%.

Before irradiation procedure for both ionisation chamber and TLD chips carried out, the QA procedures before SRS treatment were done to make sure accurateness of isocenter coordinates of the target. Comparisons between the calculated dose by i-Plan RT Dose 4.1 (Brainlab, Germany) treatment planning and measured dose from the ionisation chamber and TLD were made. The difference (R) between the calculated dose ( $D_c$ ) and mean measured dose ( $D'_m$ ) was calculated using the relation  $R = [(D'_m - D_c) / D_c] \times 100\%$ .

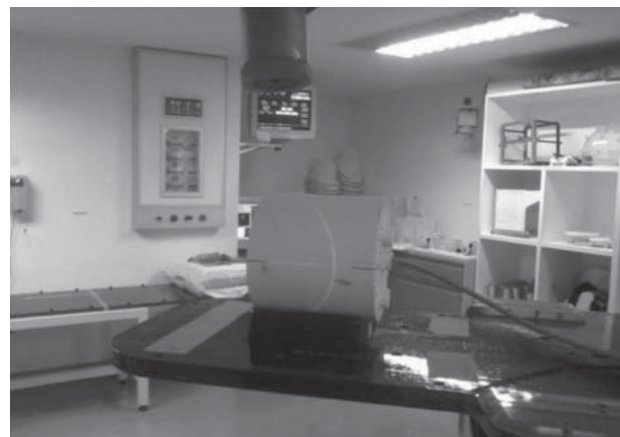


FIGURE 4. IMRT Thorax phantom setup

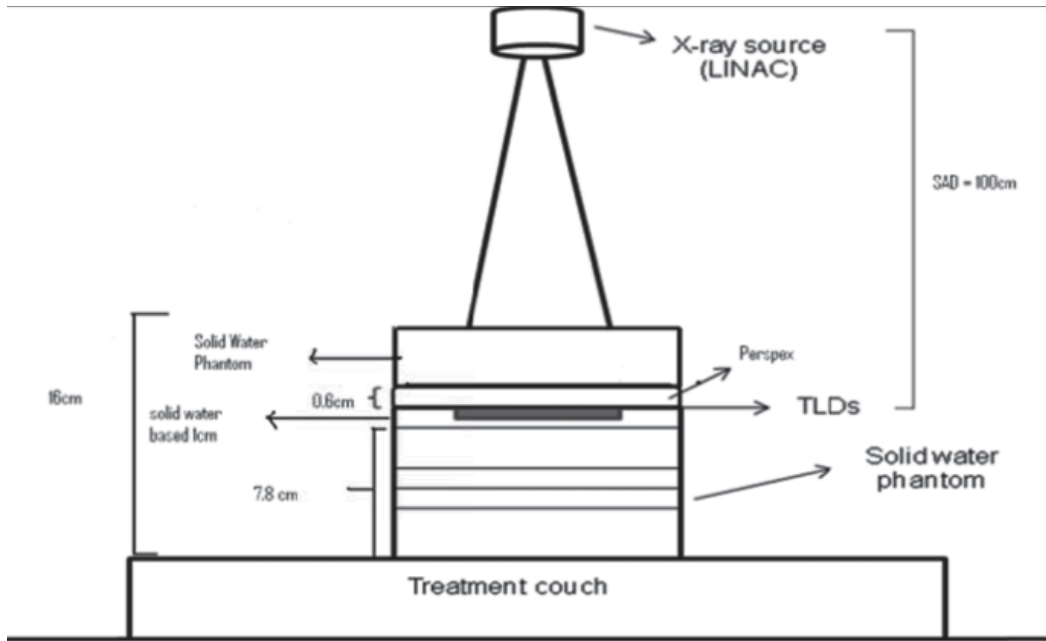


FIGURE 5. Set-up geometry of TLDs position

## RESULTS AND DISCUSSION

### BEAM DATA MEASUREMENTS

The PDDs measured in the water for 5 mm to 45 mm diameter field sizes are presented in Figure 6. In general, we observed a shift to greater depth of  $d_{max}$  with increasing field sizes, with slightly same depth of  $d_{max}$  for field size of 12.5 mm to 22.5 mm and generally increased starting at field size of 25 mm to 45 mm.

Serago et al. (1992) & Verhaegen et al. (1998) also reported that similar shift of  $d_{max}$  to greater depth with increasing field size for a 6 MV accelerator. Such observation is opposite to traditional radiation therapy fields where  $d_{max}$

decreases as field size increases. This is explained on the basis of phantom scattering than collimator scattering. From this figure also it is observed that PDD increased with increasing cone diameter. PDD values were increased due to higher production of scatter radiation at larger field size compared to smaller field sizes of beams. PDD measurement of SRS should be measured with the detectors smaller than field sizes used in order to reduce lateral scatter in-equilibrium to ensure the measurements are accurate. Pinpoint ionisation chamber of 0.016 cm<sup>3</sup> active volume and outer diameter is 4.3 mm was used for measuring PDD. According to Khan et al. (2011), for measurement of central axis depth dose or PDD, an essential criterion is that the sensitive volume of the detector should lie within

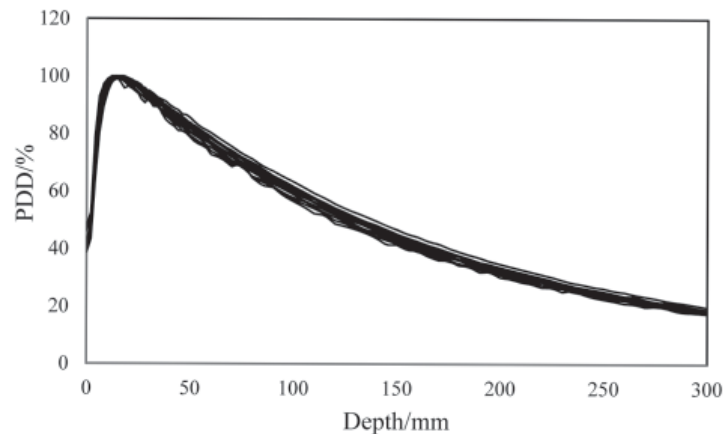


FIGURE 6. PDD versus depth for 13 circular cones of 5, 7.5, 10, 12.5, 15, 17.5, 20, 22.5, 25, 27.5, 30, 32.5 and 45 mm

uniform electron fluence generated by photons within  $\pm 5\%$ . Depth dose show maximum dose at a particular depth that depends on collimator size. However, the algorithm used for dose calculation requires all data to be normalized to the same depth. In this section, this depth is referred to as  $d_{norm}$ . Usually, the normalization depth is chosen to be average depth of peak dose. For 6 MV photon beam,  $d_{norm}$  is 15 mm. In this study, the setup of a detector accurately and precisely at the effective point of measurement which is respect to the center of active volume is the crucial part. Even a slight misalignment of the detector location and direction of motion with central ray could result in the detector being well out of the center of field. This is different compared to the larger field sizes measurements.

Figure 7 shows the Off-axis Ratio (OAR) for various circular cones beam measured at depth 75 mm inside water. It is obvious from the curves in this figure that off-axis ratio of small fields show strong field size dependence with rapidly decreasing as the field diameter decreases. The steepest dose gradient can be seen for 5 mm diameter field size. The summary results of OAR for 13 circular cone diameters measured using ionisation chamber shown in Table 1 which analyzed by Mephysto mc<sup>2</sup> analyzing beam software.

Generally, the dose falls rapidly at the edge of the beams. Therefore, the tissue that situated near the target area could be avoided from getting unnecessary dose. The results of penumbra length less than 4 mm for all 13 circular cone diameters measured using pinpoint

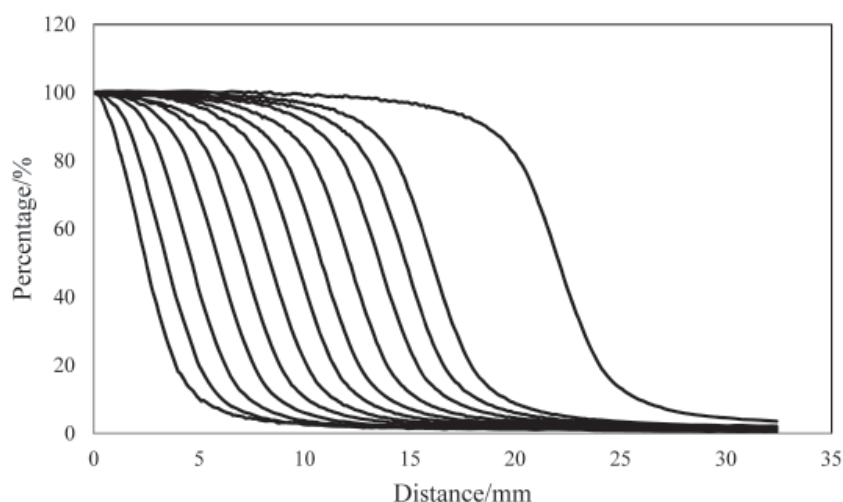


FIGURE 7. OAR for 13 SRS circular cones (plots from left to right correspond to 5, 7.5, 10, 12.5, 15, 17.5, 20, 22.5, 25, 27.5, 30, 32.5 and 45 mm)

ionisation chamber. Three important characteristics which are penumbra, flatness and symmetry could be evaluated from OAR measurements. The flatness is the constancy of intensity across the beam. In the small field size dosimetry, the flatness is not crucial because in small field size the shape of beam profile will be a peak shape because of very steep dose gradient mostly for 5 mm to 15 mm diameter of cones. For symmetry of the beam is important to ensure the intensity of the beam is uniformly distributed. The symmetry is required to be  $\pm 1\%$  to  $\pm 2\%$  from one side of the beam to the other. But for SRS, all the beams are circular in shape and assumed as small then it is not crucial as penumbra measurements. The results showed that all 13 circular cone diameters have symmetry less than 1% measured using ionisation chamber. Besides that, the asymmetry of the beams is reflecting to misalignment of laser since measurement setup is based on the laser alignment. Depth and field sizes are both factors that affect to beam symmetry and flatness. The deeper measurement was made inside medium then the beam more uniform due to increase of phantom scatter.

While in geometric penumbra must be small enough for small field size dosimetry which is range about 2 mm to 3 mm only (Shepard 2009). According Khan et al. (2011), beam collimation in SRS functioning to reduce geometric penumbra, a tertiary collimation system is used to bring the collimator diaphragm closer to the surface. A larger penumbra will cause more scattering electron produce and resulting on the irradiating healthy tissue surrounding near to the target volume. In the open field size dosimetry, the flatness is required for clinically used to be  $\pm 3\%$  recommended by Division of Engineering and Radiation Safety (2012).

The detector size is again important because of the steep dose gradients at the field size edge. The dosimeter, in such case, must have high spatial resolution to measure field penumbra accurately, which is critically important in SRS. Ionisation chamber is the most precise and the least energy-dependence system but usually has a size limitation. The size of ionisation chamber used for this purpose should not only be smaller than the beam diameter

TABLE 1. Results of central axis (CAX) deviation, field size, penumbra, flatness and symmetry for different cone diameters

Cone Diameter/mm	CAX Deviation/mm	Field Size/mm	Penumbra Left mm	Penumbra Right mm	Flatness/%	Symmetry/%
5	-0.07	5.01	2.56	2.52	22.46	0.25
7.5	0.03	6.96	2.79	2.80	20.36	0.16
10	0.06	9.31	2.97	3.01	17.55	0.08
12.5	0.01	11.92	3.13	3.16	14.96	0.16
15	0.06	14.31	3.29	3.20	12.61	0.03
17.5	0.12	16.74	3.39	3.39	11.40	0.23
20	-0.02	19.31	3.41	3.41	9.68	0.21
22.5	0.07	21.61	3.53	3.46	8.89	0.14
25	-0.01	24.30	3.51	3.51	7.84	0.20
27.5	-0.01	27.14	3.49	3.60	6.88	0.20
30	-0.02	29.60	3.56	3.58	6.47	0.17
32.5	0.02	32.19	3.57	3.57	5.88	0.37
45	0.01	44.18	3.85	3.77	4.45	0.27

but should have sufficient spatial resolution to describe the steep dose gradient in the penumbral region accurately. The effect of lack of lateral electronic equilibrium and steep dose gradient were noticed even when measurements were performed with a small volume (less than 0.02 cc) ionisation chamber. There are disadvantages of using the ionisation chamber.

Figure 8 shows the variation of scatter factor with field diameter of SRS cones measured using pinpoint cylindrical ionisation chamber. Charge collected at measuring depth for each cone size was normalized to charge for reference field size 10 cm × 10 cm. The scatter factor for 5 mm diameter cone was considered very small compared to other cone diameter.

Generally, the scatter factor increased with increasing field sizes. The steepest increasing observed at 5 mm to 7.5 mm diameter field size which the value increased rapidly by 74.9% (Figure 8). Ideally, small beam dosimetry or radiosurgery beams exhibit a sharp decrease in output with decreasing field size (Shepard 2009). Then, the values saturated from 22.5 mm to 32.5 mm diameter field size and

slightly increased for 45 mm diameter field size. The lowest scatter factor value was for 5 mm diameter field size which was 47.7% less than normalized value. Opposite that, the highest value was for 45 mm diameter field size which was 7.2% less than the normalized value (charge measured for 10 cm × 10 cm field size). Radionic's collimator housing geometry is longer compared to other collimator housing such as Brainlab and makes the collimation of field more nearer to the isocenter. The distance from isocenter to the lower end of the mounted collimator (with gantry in 0 degree angle) is 230 mm. It allows more primary radiation and scattering radiation coming from circular cone collimator reach the detector and reduce the scattering radiation that produced by interaction with air because the distance from surface of circular cone collimator to the isocenter has reduced.

In the small field dosimetry, there was less contribution of scattering radiation from the medium (water) and it is called as phantom scatter. Khan (1994) was explained the effect of phantom scatter when the field size increased. The explanation is general and agrees what was happen in the small field size dosimetry. The effect of field size on the scatter factor due to phantom scatter alone is significant as long as distance between the point of measurement and the edge of field is shorter than the range of the laterally scatter electron produced. When the certain distance is reaches, there is no further increased in the scatter factor caused by phantom scatter. When the field size is reduced below that required for lateral scatter equilibrium, the scatter factor decrease rapidly. The decreasing scatter factor for the small field size may be cause of large number of direct particles and at the same time the indirect particle was reduced due to collimator housing and circular cone in SRS procedure. Besides that, 50 mm × 50 mm was jaws size also limit the production of scatter radiation by 1 % compared to 60 mm × 60 mm jaws size.

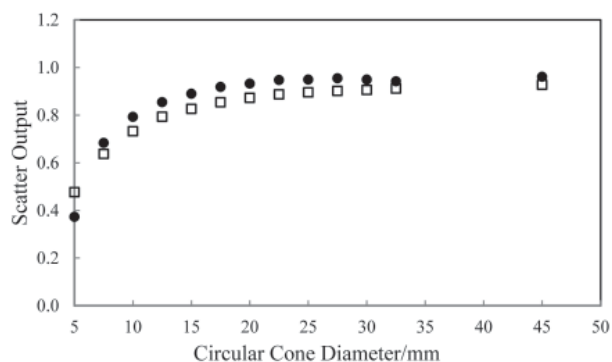


FIGURE 8. Scatter factor for 13 circular cones; (□): 5 cm × 5 cm and (•): 6 cm × 6 cm

QUALITY ASSURANCE (QA) PROCEDURES  
BEFORE SRS TREATMENT

Table 2 showed the total localization error which represent by  $\Delta r$ . The total error which is measured on TPS using Brainlab CT localizer is  $1.1 \pm 0.4$  mm.

TABLE 2. Results of localization errors of CT using Brainlab CT Localizer

No. of measurement	mm			$\Delta r$
	$\Delta AP$	$\Delta Lat$	$\Delta Vert$	
P1	0.3	0.6	0.8	1.04
	0.4	0.6	1.3	1.49
	0.2	0.4	1.3	1.37
	0.2	0.2	0.4	0.49
Mean	0.3	0.5	1.0	1.1
Standard Deviation	0.1	0.2	0.4	0.4

Among the components of localization errors of CT, the errors of vertical direction were much greater than AP and lateral directional errors, mainly due to image resolution. As the accuracy of CT localization is directly related to three dimensional (3D) image resolution. Choi et al. (2001) also reported that, in geometric phantom test, a 3 mm slice was used and the resulting error was  $1.2 \pm 0.5$  mm. This value is comparable to results of the other studies  $0.91 \pm 0.3$  mm for 2 mm slice and  $1.58 \pm 0.5$  mm for 4 mm slice. Our results are slightly agreed with study done by Choi et al. (2001), the resulting error was  $0.91 \pm 0.3$  mm for 2 mm slice thickness used compare to our study which is  $1.1 \pm 0.4$  mm for 2.4 mm slice thickness. From this study, it is found that an increase in slice thickness may affect in the inferior-superior (vertical) direction. The CT localization error increases with increasing slice thickness used.

As shown in Table 3, mean displacement error was  $0.88 \pm 0.8$  mm, mainly due to gantry sag and localization error for the target localizer. The values are still in line with the accuracies suggested by AAPM Report No. 54 (1995) which is 1.0 mm. In general, the target isocentric deviation in the vertical direction is much greater than one in the AP or lateral direction because of gantry sag to compensate for the gantry weight (Choi et al. 2001). In our case, vertical

TABLE 3. Results on the shift distance of Target Positioner

Overall isocenter deviation (linac + Collimator mount + Target Positioner)			
Parameter measured	Distance on Film /mm	Magnification Factor (MF)	Distance on Isocenter /mm
Head- feet at $0^\circ$ ( $HF_0$ )	2	0.952	1.90
Head- feet at $90^\circ$ ( $HF_{90}$ )	1.2		1.14
Left - right	0.1		0.10
Anterior-posterior	0.4		0.38
Mean			$0.88 \pm 0.8$

direction for  $0^\circ$  in gantry angle gave the highest deviation (1.90 mm).

Couch-mounted systems strongly rely on a set of three wall-mounted lasers to provide a frame of reference in the treatment room. The lesion is positioned at machine's isocenter by visually aligning these lasers with stereotactic coordinates that are scribed onto the target positioner. Meeks et al. (1998) reported that, using these lasers as the sole mechanism for positioning radiosurgery patients can result in greater than 1 mm systematic uncertainty. The target coordinates from treatment planning are located at the isocenter of the linear accelerator using LTLF. As the center of exposed beam is equal to the isocenter of linear accelerator, we can estimate the deviation of target coordinates using a fiducial marker of the target localizer. As the target coordinates from treatment planning are defined relative to the BRW frame, if we assume that the frame is firmly attached to the patient's skull, we could detect human errors, the accuracy of LTLF, laser alignment and circular cone's alignment.

DOSE VERIFICATION

In Table 4 shows results of difference between calculated dose by TPS and dose measured using pinpoint ionisation chamber in range 0.52% to 3.48%.

Table 5 showed the TLD also have a good agreement of calculated and measured dose. The mean calculated dose was 275.5 cGy, while the mean measured dose was

TABLE 4. Results of calculated dose and measured dose using pinpoint ionisation chamber

No. of Arc	Beam Direction			Calculated Dose at Isocenter/cGy	Measured Dose at Isocenter/cGy	Dev. (R)/%
	Gantry Start	Gantry Stop	Couch			
1	240	300	0	341.0	342.79	0.52
2	320	10	90	331.0	342.53	3.48
3	320	10	45	336.0	344.59	2.56
			Mean	336.0	343.30	2.17



TABLE 5. Results of calculated dose and measured dose using TLD 100H

No. of Arc	Beam Direction			Calculated Dose at Isocenter/cGy	Measured Dose at Isocenter/cGy	Dev. (R)/%
	Start	Gantry Stop	Couch			
1	0	0	0	85.0	83.26 ± 0.22	-2.05
2	0	0	0	466.0	445.0 ± 1.11	-4.51
			Mean	275.5	264.11 ± 0.48	-3.28

264.11 cGy. Mean measured dose values underestimated compared with calculated dose but the deviation or discrepancy were less than 4% and considered to be within experimental error.

The main errors affecting the result of measured dose using ionisation chamber are likely due to position of the ionisation chamber within straddling fluence segments of the beam (Basran & Yeboah 2008). The ion chamber must be placed at lower-dose gradient region to improve the agreement of measured and calculated dose. However it is very hard to distinguish between high-dose and low-dose region from the stereotactic plans because of the concentration of the radiation's intensity focused to small beam area. In small beam dosimetry, by using a small active volume of ionisation chamber lead to underestimate on the dose result especially if the entire chamber is not enclosed within the central uniform dose region due to displacement error. The reason for an underestimation of dose is lack of lateral electron equilibrium. Besides that, even though an ionisation chamber used are generally thought as small probably it is large enough so that a non-uniform dose is averaged over the ionisation chamber's active volume. As a result, the dose measured to be smaller than it is.

There are many factors that affected the accuracy of using the TLD in measuring the dose. For high sensitivity TLD such as TLD 100H, the annealing procedure is one of the factors affecting the Thermo Luminescent (TL) signal values. The response of lithium fluoride (LiF) TLD is sensitive to thermal history. The TLD materials were annealed before being used to empty the electron traps associated with defect structures in the crystal lattice. The recommended thermal treatment for TLD 100H is annealing at 240° C for 10 minutes. When the LiF is reused, it is important to repeat the same thermal history to maintain the sensitivity of the TLD. But because of the annealing duration is too short, the temperature inside the annealing oven may be fluctuated and that will affect the performance and response of TLD 100H. Another factor contributing to the error of the result is the handling and storage procedures using TLD 100H. Mishandling of the TLD can affect their TL sensitivity, stability, precision and minimum detectable of dose. For the purpose of discussing possible uncertainties in stability and sensitivity associated with the handling of TLD, it is assumed that the choice of a particular form TLD is based primarily on dosimetric considerations. Within limits, the TL sensitivity of TLD

is directly proportional to the mass of active phosphor present. In each TLD chips, the mass of active phosphor present is fixed during manufactured and must not change when used in the measurements. So that, the TLD must be taken care not to scratch or abrade their surfaces.

## CONCLUSION

The accuracy of the dose verification mainly due to the accuracy of measured dosimetric parameters (PDD, OAR and Scatter Factor) which then used as input beam data for TPS's algorithm to calculate the dose. In the same time, accuracy of SRS dose delivery also correlated to the following: CT image localization, isocentric deviation of linear accelerator. The mean overall accuracy for target isocenter localization measured in this study was  $0.88 \pm 0.8$  mm, the values are still in line with the accuracies suggested by AAPM Report No. 54 (1995) which is less than 1.0 mm. In this study, the results of measured dose for both detectors (TLD 100H and pinpoint ionisation chamber) give a good agreement with calculated dose with deviation of less than 4%. The results are acceptable such as recommended by ICRU Report 50 (1994) that dose delivered to the target volume must be within  $\pm 5\%$  error. Instead of measuring the point dose, the isodose distribution could be verified to ensure that the surrounding tissues and organs at risk receive dose within the expectation. The further study must be carried out to verify the dose distribution with dosimetry film and TLD.

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## REFERENCES

- AAPM. 1994. *AAPM Report No.47: AAPM code of Practice for Radiotherapy Accelerators*. USA: American Association of Physicist in Medicine.
- AAPM. 1995. *AAPM report No.54: Stereotactic Radiosurgery*. USA: American Association of Physicist in Medicine.

- AAPM. 2009. *Task Group 142 Report: Quality Assurance of Medical Accelerators*. USA: American Association of Physicist in Medicine.
- Al-Najjar, W.H., Prasad, S.G., Parthasaradhi, K., Bloomer, W.D. & Nanda, R.K. 1998. Dosimetric aspects of small circular fields of 10 MV photon beam. *Medical Dosimetry* 23(1): 39-42.
- Attix, F.H. 1986. *Introduction to Radiological Physics and Dosimetry*. Wisconsin: Wiley VCH Verlag GmbH & Co.
- Basran, P. & Yeboah, C. 2008. Dosimetric verification of micro-MLC based intensity modulated radiation therapy. *Journal of Applied Clinical Medical Physics* 9(3): 2832.
- Choi, D.R., Kinl, D.Y., Ahn, Y.C., Huh, S.J., Yeo, I.J., Nam, D.H., Lee, J.I., Park, K. & Kim, J.H. 2001. Quantitative analysis of errors in fractionated stereotactic radiotherapy. *Medical Dosimetry* 26(4): 315-318.
- Cosgrove, V.P., Jahn, U., Pfaender, M., Bauer, S., Budach, V. & Wurm, R.E. 1999. Commissioning of a micro multi-leaf collimator and planning system for stereotactic radiosurgery. *Medical Dosimetry* 50(3): 325-335.
- Division of Engineering and Radiation Safety. 2012. *Manual of Quality Assurance Programme in Radiotherapy Services*. Putrajaya: Ministry of Health, Malaysia.
- Geotsch, S.J., Murphy, B.D., Schmidt, R., Micka, J., De Werd, L., Chen, Y. & Shockley, S. 1999. Physics of rotating Gamma system for stereotactic radiosurgery. *International Journal Radiation Oncology Biological Physics* 43(3): 689-696.
- Hamilton, R.J., Kuchnir, F.T., Sweeney, P., Rubin, S.J., Dujovny, M., Pelizzari, C.A. & Chen, G.T.Y. 1995. Comparison of static conformal field with multiple noncoplanar arc technique for stereotactic radiosurgery or stereotactic radiotherapy. *International Journal Radiation Oncology Biological Physics* 33(5): 1221-1228.
- ICRU. 1994. *ICRU Report 50; Prescribing, Recording and Reporting Photon Beam Therapy*. USA: International Commission on Radiation Units and Measurements.
- Kahn, F.M. 1994. *The Physics of Radiation Therapy*. Second Edition. USA: William and Wilkins.
- Khan, F.M., Gibbons, J., Mihailidis, D. & Alkhatib, H. 2011. *Khan's Lectures: Handbook of the Physics of Radiation Therapy*. USA: Lippincot Williams and Wilkins.
- Lee, F.T., Larsen, T. & Jaywant, S.M. 2001. Immobilization in stereotactic radiotherapy: The head and neck localizer frame. *Medical Dosimetry* 26(3): 267-273.
- McDermott, L. & Perkins, A. 2004. Comparison of measured and calculated radiotherapy doses in the chest region of an inhomogeneous humanoid phantom. *Australasian Physical & Engineering Sciences in Medicine* 27(1): 16-21.
- Meeks, S.L. Bova, F.J., Friedman, W.A., Buatti, J.M., Moore, R.D. & Mandenhall, W.M. 1998. IRLED-based patient localization for LINAC radiosurgery. *International Journal Radiation Oncology Biological Physics* 41(2): 433-439.
- Oncology Care Systems Group. 1999. *PRIMUS/PRIMARTIMEVATRON; Acceptance Test Procedure*. USA: Siemens Medical Systems Inc.
- Rogers, D.W.O. 1996. *Fundamentals of Dosimetry Based on Absorbed-dose Standards*. National Research Council of Canada, AAPM's Summer School. USA: American Association of Physicist in Medicine.
- Rowshanfarzad, P., Sabet, M., O'Connor, D.J. & Greer, P.B. 2011. Isocenter verification for linac-based stereotactic radiation therapy: Review of principles and techniques. *Journal of Applied Clinical Medical Physics* 12(4): 3645.
- Serago, C.F., Houdek, P.V., Hartmann, G.H., Saini, D.S., Serago, M.S. & Kaydee, A. 1992. Tissue maximum ratios (and other parameters) of small circular 4, 6, 10, 15 and 24 MV x-ray beams for radiosurgery. *Physics in Medicine & Biology* 37(10): 1943-1956.
- Sharma, S.D., Kumar, S., Dagaonkar, S.S., Bisht, G., Dayanand, S., Devi, R., Deshpande, S.S., Chaudry, S., Bhatt, B.C. & Karman, S. 2007. Dosimetric comparison of linear accelerator-based stereotactic radiosurgery system. *Journal of Medical Physics* 32(1): 18-23.
- Shepard, D. 2009. Quality assurance in stereotactic radiosurgery and fractionated stereotactic radiotherapy. AAPM 51st Annual Meeting. California, 26-30 July.
- Sontag, M.R. & Steinberg, T.H. 1999. Performance and beam characteristics of the Siemens Primus linear accelerator. *Medical Physics* 26(5): 734-736.
- Stefan, F., Sipila, P.M., Kulmala, J.A.J. & Minn, H.R.I. 1998. Dose characteristics of in house-built collimators for stereotactic radiotherapy with a linear accelerator. *Physics in Medicine & Biology* 43: 1545-1556.
- Treuer, H., Hoevens, M., Luyken, K., Geirich, A., Koecher, M., Muller, R.P. & Sturm, V. 2000. On isocenter adjustment and quality control in linear accelerator based radiosurgery with circular collimators and room lasers. *Physics in Medicine & Biology* 45: 2331-2342.
- Verhaegen, F., Das, I.J. & Palmans, H. 1998. Monte Carlo dosimetry study of a 6 MV stereotactic radiosurgery unit. *Physics in Medicine & Biology* 43: 2755-2768.
- Wang, L., Li, J., Paskalev, K., Hoban, P., Luo, W., Chen, L., McNeeley, S., Price, R. & Ma, C. 2006. Commissioning and quality assurance of a commercial stereotactic treatment planning system for extracranial IMRT. *Journal of Applied Clinical Medical Physics* 7(1): 21-34.
- Weidlich, G.A., Gebert, J.A. & Fuery, J.J. 1998. Clinical commissioning of Laitenen stereoadapter for fractionated stereotactic radiotherapy. *Medical Dosimetry* 23(4): 302-306.
- Zamzuri, I., Nik Ruzman, N.I., Jafri, A.M., Mar, M., Ahmad, Z. & Biswal, M.B. 2006. Early Malaysian experience on the use of head and neck localizers in precision radiotherapy of intra and extra cranial sites for first 28 cases. *Medical Journal Malaysia* 61(5): 621-625.
- Zamzuri, I., Badriyah, I., Ralmlan, G. I., Pal, H.K., Muzaimi, M., Jafri, A.M., Mar, W., Shafie, A.M., Nik Ruzman, N.I., Biswal, M.B. & Ahmad, Z. 2011. LINAC based radiosurgery and radiotherapy for neurosurgical disease: What have we learn so far. *Medical Journal Malaysia* 66(4): 341-344.

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