Iran. J. Radiat. Res., 2005; 3 (2): 73-78

Quality assurance program for prototype stereotactic system developed for Neptun 10 PC linac

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Background: A prototype stereotactic radiosurgery set was designed and constructed for a Neptun 10 PC linac that is currently being used at Imam Reza hospital in Mashhad. ***Materials and Methods:*** A complete quality assurance program was designed and performed for the constructed system including isocentric accuracy test, localization accuracy test, dose delivery accuracy test and leakage radiation test. Target simulator, control alignment device and plexiglass phantom which were parts of the developed hardware were used to fulfill quality assurance program. ***Results:*** The average isocentric shift resulted from the gantry rotation and couch turning were respectively obtained to be 1.4 and 2 mm. The average localization error in the three coordinates was found to be 2.2 mm. The total treatment uncertainty due to all of the probable errors in the system was equal to 4.32 mm. The dose delivery accuracy test was carried out, the result indicated a 3-7% difference between the given and measured dose. ***Conclusion:*** The quality assurance tests showed consistent performance of the constructed system within the accepted limits; however, some inconsistency might exist in certain cases. The safety of SRS method is increased when the overall uncertainty is minimized and the treatment of the lesions adjacent to critical organs is avoided. **Iran. J. Radiat. Res., 2005; 3 (2): 73-78**

Keywords: *Quality assurance, stereotactic radiosurgery system, linac.*

INTRODUCTION

Stereotactic radiosurgery was initially introduced by a Swedish neurosurgeon, Leksell in 1951[1]. In this method, a high radiation dose is delivered to a small defined volume. Stereotactic radiosurgery (single fraction dose) and radiotherapy (fractionated) are two step processes:

1. Accurately defining the shape, size and location of a lesion in a reference stereotactic frame using CT, MRI or angiography modalities.
2. Performing planned treatment.

With certain modifications, available linear accelerators can be used for stereotactic treatment[2].

The required instruments for stereotactic radiosurgery/radiotherapy include collimators (to limit the radiation field), patient docking device (to affix patient's head to treatment couch), head rings (to attach patient's head to the localizer and centering box), and finally quality assurance devices all of which were designed and constructed in our department.

The spatial accuracy of treatment generally depends on several parameters, which are as follow[3]:

a. Rigidity and immobilization of the stereotactic frame.
b. Image slices thickness and the dimensions of its pixels.
c. Spatial or isocentric accuracy of the linear accelerator.
d. Displacement of head between the imaging and the treatment setup.

Small radiation field and high radiation dose used in stereotactic treatments have drawn attention to the issue of the quality assurance more than any other treatment modalities[4]. We have performed quality assurance test as acceptance program for constructed hardware.

MATERIALS AND METHODS

Neptun 10 PC linear accelerator with 9 MV X-ray was used in this study. The hardware needed for stereotactic treatment was designed according to the physical and

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geometric properties of the linac. Arc therapy option (software and hardware) was installed on the linac enabling the delivery of radiation within the range of 1-8 cGy/deg.

Target simulator

This device as a part of the quality assurance set is composed of a base plate with four couplings, a plexiglass plate within which a 3 mm diameter steel sphere is embedded, a rotating and a stationary column (figure 1).

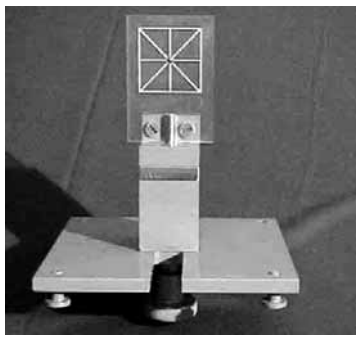

Figure 1. Target simulator used for isocentric accuracy test.

Control alignment device

This device, also as a part of the quality assurance set, is composed of one vertical bar and two rectangular rings to hold the verification film. The proximal end of the bar is attached to collimator assembly and its distal end to the rings (figure 2).

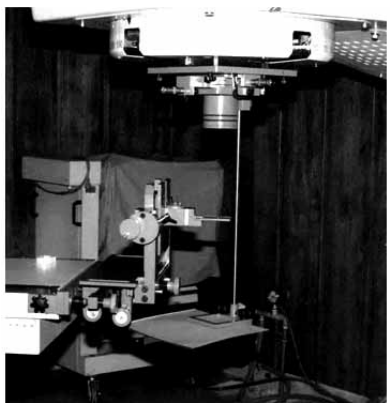

Figure 2. Isocentric alignment control along with target simulator.

Plexiglass head phantom

The main cylindrical body of the phantom is made of plexiglass, 200 mm in height and 140 mm in diameter, including 15, 10 and 5 mm thick slabs. The alignment of the slabs is accomplished by two vertical threaded rods of plexiglass which are fixed at the top with two screws of the same material (figure 3).

Figure 3. Plexiglass head phantom.

The head phantom contains two main slabs namely the localization and dosimetry slab. The first one is composed of two 15 mm thick layer which accommodates three circular, rectangular and triangular objects.

A total of one hundred and fifty holes (1 mm in depth and 2 mm in diameter) were machined into the dosimetry layer to accommodate calibrated TLD microchips. These holes were arranged in concentric radii and 3 mm apart to minimize perturbation of dose distribution and to maintain a suitable resolution[5].

The localization and dosimetry slabs can be placed in every desired height in the head phantom. The upper most slab of the phantom is semispherical in shape to resemble human's head.

The phantom can be attached to the patient docking device through a plexiglass base plate.

Treatment planning software

An ERGO (V 3.1) system provided by 3D Line International s.r.l from Italy was used for treatment planning.

Isocentric accuracy test

The most important factor affecting the accuracy of stereotactic treatment is the isocentric accuracy of linear accelerator[6]. Isocentric accuracy denotes the space where the collimator, gantry and couch rotation axis meet. Dimensions of this space are to be measured in this test.

Practically, the target is placed stereotactically on the mechanical isocenter of the linear accelerator, the point where both the ceiling and wall lasers in the treatment room intersect. Although the emerging beams of the linear accelerator might not necessarily intersect this point, but they intersect another point namely the radiation isocenter. The accuracy of the treatment depends on the proximity of these two points[3]. According to AAPM report no.54 the

gantry, couch and collimator rotation axis has to meet within a sphere of 2 mm in diameter.

The isocentric accuracy test was performed by examining the alignment of telemeter with the isocenter of the the linac, the room laser and the mechanical isocenter.

The largest collimator was mounted on the head of the gantry and the target simulator was connected to the patient docking device by the means of four couplings. The collimator was then aligned horizontally by adjustment of its rotational screws.

The proximal end of the vertical bar of the alignment control device was connected to the collimator holder and its distal end to the rectangular rings to hold radiotherapy verification film (figure 2).

Finally, radiographs of the steel ball were taken at twelve different gantry positions, created angles, by 30 degree rotation intervals and four angles of couch turning in steps of 45 degrees. The radiographs were taken using 100 MU. The distance between the center of the radiation field and the steel ball image was taken as the isocenter shift in each angle (figure 4).

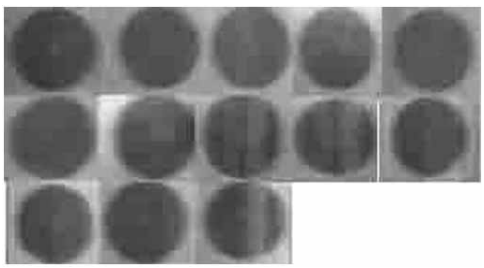

Figure 4. Radiographs taken in isocentric alignment control.

Localization test

To perform the localization test, the localization slab of the plexiglass phantom was placed at a known height which provides the exact coordinates of the three objects. The phantom was attached to the head ring via its base plate which was followed by mounting the localizer on the head ring. Finally the 5 mm thick CT slices were taken using the plexiglass phantom.

The field of view (FOV) has to be selected in a way that it accommodates all the rods of the localizer in the topogram, which in turn allows fiducial markers to be seen in the CT slices. The CT images were then transferred to the treatment planning software (figure 5). The localization of the three objects was performed by digitizing the fiducial markers observed on the CT images.

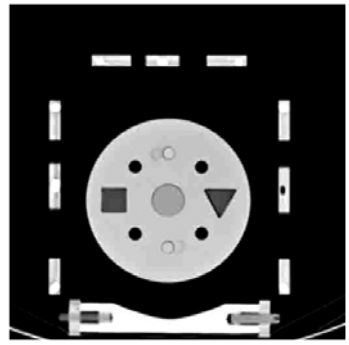

Figure 5. CT scan taken from the localization layer of the phantom.

Dose delivery accuracy test

As it has already been stated, the plexiglass phantom contains a dosimetry slab accommodating TLD microcubes. Initially, the center of the localization slab was localized and defined as the center of the target.

Treatment planning was performed delivering 15 Gy to the target. Finally the radiosurgery treatment was simulated on the plexiglass phantom (figure 6). Irradiation was performed using the 25 mm collimator and five arcs of 180 degrees. The TLD chips were read after the treatment simulation was completed.

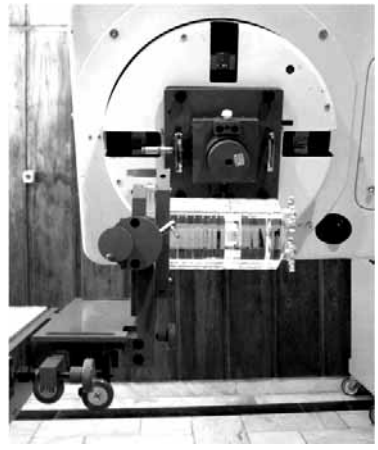

Figure 6. Simulated stereotactic treatment of the head phantom in the dose delivery accuracy test.

Headring displacement test

Headring is a device used to hold the localizer and the centering box on it and it will be connected to the patient's head, and also to the patient docking device by means of

four couplings to keep the head firmly inside the head frame. Thermoplastic mask and dental mold should be custom made for each patient. Tattoos are marked on patient's face and the corresponding parts of thermoplastic mask are also marked, hence any headring displacement between the imaging and treatment could be detected and measured.

Leakage radiation test

This test was performed during the simulation of the radiosurgery treatment on the plexiglass phantom. The TLD chips were placed on the phantom corresponding the approximate location of the eyes and thyroid. The dose received by these two critical organs was measured and compared against the dose (15 Gy) delivered to the target.

RESULTS

Table 1 shows the isocenter shifts under two conditions; couch stationary-gantry rotating and gantry stationary-couch rotating. The average isocenter shift due to the gantry rotation was found to be 1.4 mm with the maximum and minimum values of 2.8 and 0 mm respectively (SD = 1.072 mm). The average isocenter shift due to the couch rotating was equal to 1.975 mm with the maximum and minimum values of 2.5 and 1.4 mm respectively (SD = 0.607 mm).

Finally, the average isocenter shift due to the gantry and couch was 1.544 ± 0.99 mm where 0.99 mm represents one standard deviation. Assuming a normal distribution for the individual isocenter shift, the uncertainty should be less than 3.17 mm (= 1.544 + 1.65 × 0.99) in 95% of the time.

The localization was repeated three times for each individual object. Table 2 shows the results of the localization accuracy test. The values are the average error with one standard deviation.

The total localization error in three orthogonal coordinates is derived from the following equation:

$$\text{Total error} = \sqrt{(\Delta x)^2 + (\Delta y)^2 + (\Delta z)^2}$$

Using the above equation the total localization error was found to be 2.2 ± 1.23 mm. Therefore, by assuming normal distribution, the error associated with the localization should be less than 4.23 mm (2.2 + 1.65 × 1.23) in 95% of the time.

Table 1. Isocenter shifts resulted from gantry rotation and couch turning.

Gantry rotation	Couch angle	Isocenter shift (mm)
0	0	0
30	0	0
60	0	1.6
90	0	0.75
120	0	2.8
150	0	2.2
180	0	2.75
330	0	0
300	0	0.75
270	0	1.5
240	0	2
180	0	2.5
Average	1.4 ± 1.072 mm	
0	-45	1.5
0	-90	2.5
0	+45	1.5
0	+90	2.5
Average	2 ± 0.577 mm	

Table 2. Localization errors for x, y and z coordinates.

x-coordinate	0.3 ± 0.294 mm
y-coordinate	1.525 ± 1.3 mm
z-coordinate	1.525 ± 1.007 mm

For the three simulated stereotactic treatments delivering 15 Gy to the center of the target, there was a difference of 0.4, 0.7 and 1 Gy between the prescribed and measured dose. The results suggested that average error in the dose delivery was 4.6% with 2% standard deviation. Therefore, assuming normal distribution, the uncertainty in the spatial dose delivery will be less than 3.95 mm (2.7 + 1.65 × 0.75) in 95% of the time.

During the three actual treatments the average displacement of the headring and the stereotactic frame was equal to 1.33 ± 0.288 mm consequently.

The dose received by the eyes and thyroid from the three simulated stereotactic treatments was equal to 13.5 and 6.5 cGy

respectively. Therefore, the two organs received 0.9 and 0.4 percent of the prescribed dose.

DISCUSSION

Using 1 and 3 mm thick CT slices would result in a total treatment uncertainty of 2.4 and 3 mm respectively[3]. Therefore, even under the best condition, applying 5 mm thick CT slices a spatial uncertainty better than 4 mm should not be expected.

Being aware of the total spatial uncertainty in the treatment, the diameter of the selected collimator could easily be determined[6]. The localization uncertainty depends on the imaging modality matrix size, the thickness of the image slices and to some extent the type of the localizer being used[3].

Matrix size of the CT slices was 512 × 512 and the corresponding pixel size was 0.7 mm which is ideal for localization. Having in mind the minimum used CT slice thickness of 5 mm, a localization error of 4.23 mm (with accuracy of 95%) could be acceptable.

The isocentric uncertainty depends on the mechanical properties of the linear accelerator. The measured value of the isocentric uncertainty (3.17 mm with accuracy of 95%) is not exactly equal to the value recommended by AAPM report no.54. The headring displacement of 1.8 mm for the constructed non-invasive system is acceptable as compared with the traditional ones[7].

The total spatial uncertainty (TSU) is calculated according to the sum of the squares law as following:

$$TSU = \sqrt{(3.17)^2 + (4.23)^2 + (1.8)^2} = 5.24 \text{ mm}$$

In spite of the above fact, the result of the dose delivery test does not verify the calculated value. The measured value of the spatial uncertainty (3.95 mm) is close to the localization uncertainty of 4.23 mm. In other words, it seems that the linac isocentric shifts have no effect on the total uncertainty. This situation may only occur when the isocentric shifts happen randomly and it does not necessarily add to the localization uncertainty[3].

Regarding the type of the performed dose delivery test, in which the phantom is rigidly attached to the headring, the condition is similar to the invasive fixation method. Therefore, the headring displacement should be taken into consideration for calculating the total spatial uncertainty in the non-invasive method. According to the sum of the squares law the uncertainty is as following:

$$\text{Treatment uncertainty} = \sqrt{(3.95)^2 + (1.8)^2} = 4.34 \text{ mm}$$

The above value is not in agreement with the recommendation of AAPM report no. 54. The dose delivery accuracy test also shows an uncertainty of 7.9% for the absorbed dose (with accuracy of 95%), which exceeds the 5% limit recommended by AAPM. The main reason for the difference may be due to the fact that the measurements were performed in plexiglass with higher electron density than water. Therefore, correction should be made for the difference in the electron density[5].

Another reason for the uncertainty is that TLD chips normally show a supralinear response to the doses above 1 Gy. Even though a dose-response curve was obtained for the TLD chips in the 0 - 15 Gy region and all of the measured values were corrected for supralinearity, the incomplete fitness of the obtained correction function to the actual absorbed dose may be a substantial factor for this difference[8, 9].

The leakage radiation test showed that the eyes and thyroid respectively received 0.9 and 0.4 percent of the prescribed dose which is in agreement with the values obtained from a similar measurement made for the traditional systems[3]. The results verified the necessary radiation protection provided by the constructed instruments.

CONCLUSION

The direct outcome of the quality assurance program is to make the decision whether to begin stereotactic radiosurgery treatment. The isocentric uncertainty behaves randomly and does not necessarily add to the total uncertainty. Since there is some difference between the acceptable limits and the measured value, the following recommendations are suggested:

1. Minimizing the errors by making the possible corrections.

2. Using thinner CT slices.
3. Making the thermoplast mask and the dental mold in order to precisely fit the patient contour carefully.
4. Not treating the lesions that are in close proximity to the critical organs such as the optic nerve and brain stem .
5. Fractionating the applied dose in order to increase the tolerance dose level for the critical organs when possible.
6. Determining the local acceptable level after a trial period.
7. Limiting the effect of isocentric uncertainty by translating the intersection of the room lasers to the radiation isocenter for the error resulted from the gantry. It is also possible to limit the effect of the isocentric uncertainty by moving the linac table in the opposite direction for the errors resulted from the couch angle.

ACKNOWLEDGMENT

The authors would like to extend their special thanks to Engineer Nader Salehi and Dr. Marco Luzzara at 3DLine society of Italy for their generous and invaluable help in making this project possible. All the support provided by the office of vice president for research affairs and Imam Reza radiotherapy department staff are greatly appreciated.

REFERENCES

1. Leksell L (1951) The stereotactic method and radiosurgery of the brain. *Acta Chir Scand,* ***182****: 1222-1223.*
2. Betti O and Derechinsky V (1983) Irradiation stereotaxique multifaisceux. *Neurochriurgie,* ***29****: 295-298.*
3. AAPM (American Association of Physicists in Medicine) (1995) Report No. 54. Stereotactic radiosurgery.
4. Yeung D, Palta J, Fountanesi J, Kun L (1993) Systematic analysis of error in target localization and treatment delivery in stereotactic radiosuregery (SR). *Int J Radiat Oncol Biol Phys,* ***28****: 493-8.*
5. Bazioglu M A, Theodorou K, Kappas C, Kalef-Ezra J, (2002) A multi purpose head phantom for stereotactic radiotherapy. *Physica Medica,* ***18****: 121-127.*
6. Lutz W, Winston K, Maleki N (1988) A system for stereotactic radiosurgery with a linear accelerator. *Int J Radiat Oncol Biol Phys,* ***14****: 373-381.*
7. Kooy HM, Dunbar SF, Tarbell NJ, Mannarino E, Ferraro N, Shusterman S, et al. (1994) Adaptation and verification of the relocatable Gill-Thomas-Cosman frame in stereotactic radiotherapy. *Int J Radiat Oncol Biol Phys,* ***30****: 685-691.*
8. Ertl A, Hartl RFE, Zenetmayer M, Kitz K, Griffit W(1996) TLD array for precise dose measurement in stereotactic radiation techniques. *Phys Med Biol,* ***41****: 2679-2686.*
9. Treuer H, Hoevels M, Luyken K, Gierich A, Kocher M, Muler RP, Sturm V(2000) On isocenter adjustment and quality control in linear accelerator based radiosurgery with circular collimator and room lasers. *Phys Med Biol,* ***45****: 2331-45.*