Original Article PROGRESS in MEDICAL PHYSICS Vol. 26, No. 4, December, 2015
http://dx.doi.org/10.14316/pmp.2015.26.4.193

Commissioning Experience of Tri-Cobalt-60 MRI-guided Radiation Therapy System

Jong Min Park*†‡§, So-Yeon Park*†‡‖, Hong-Gyun Wu*†‡¶, Jung-in Kim*†‡§

*Department of Radiation Oncology, Seoul National University Hospital, Seoul,
†Biomedical Research Institute, Seoul National University Hospital, Seoul,
‡Institute of Radiation Medicine, Seoul National University Medical Research Center, Seoul,
§Center for Convergence Research on Robotics, Advanced Institutes of Convergence Technology, Suwon,
‖Interdisciplinary Program in Radiation Applied Life Science, Seoul National University College of Medicine, Seoul, ¶Department of Radiation Oncology, Seoul National University College of Medicine, Seoul, Korea

The aim of this study is to present commissioning results of the ViewRay system. We verified safety functions of the ViewRay system. For imaging system, we acquired signal to noise ratio (SNR) and image uniformity. In addition, we checked spatial integrity of the image. Couch movement accuracy and coincidence of isocenters (radiation therapy system, imaging system and virtual isocneter) was verified. Accuracy of MLC positioing was checked. We performed reference dosimetry according to American Association of Physicists in Medicine (AAPM) Task Group 51 (TG-51) in water phantom for head 1 and 3. The deviations between measurements and calculation of percent depth dose (PDD) and output factor were evaluated. Finally, we performed gamma evaluations with a total of 8 IMRT plans as an end-to-end (E2E) test of the system. Every safety system of ViewRay operated properly. The values of SNR and Uniformity met the tolerance level. Every point within 10 cm and 17.5 cm radii about the isocenter showed deviations less than 1 mm and 2 mm, respectively. The average couch movement errors in transverse (x), longitudinal (y) and vertical (z) directions were 0.2 mm, 0.1 mm and 0.2 mm, respectively. The deviations between radiation isocenter and virtual isocenter in x, y and z directions were 0 mm, 0 mm and 0.3 mm, respectively. Those between virtual isocenter and imaging isocenter were 0.6 mm, 0.5 mm and 0.2 mm, respectively. The average MLC positioning errors were less than 0.6 mm. The deviations of output, PDDs between mesured vs. BJR supplement 25, PDDs between measured and calculated and output factors of each head were less than 0.5%, 1%, 1% and 2%, respectively. For E2E test, average gamma passing rate with 3%/3 mm criterion was 99.9%±0.1%.

Key Words: MRI-guided radiation therapy system, Commissioning, Co-60, Intensity modulated radiation therapy

Introduction

There has been a strong demand for magnetic resonance image guided radiation therapy (MR-IGRT) system in the field of radiotherapy due to MR image's superior power to distinguish soft tissues over computed tomography (CT) images.[1,2] Recently, commercial MR-IGRT system (ViewRay®, ViewRay Inc., Cleveland, OH, USA) has been introduced to the radiotherapy field and implemented in the clinic.[3] The ViewRay® system consists of on-board MR imaging system with 0.35 T static magnetic field and radiation therapy system with tri Co-60 sources.[3] Since Co-60 is immune from the environments of MR system, differently from linear accelerator (linac), ViewRay® system adopted Co-60 radioisotope as a radiation source. Although penumbra of Co-60 is larger than that generated with a linac, ViewRay® system minimized the pe-

This research was supported by a grant of the Korea Health Technology R&D Project through the Korea Health Industry Development Institute (KHIDI), funded by the Ministry of Health & Welfare, Republic of Korea (grant number: HI14C3459).
Received 23 November 2015, Revised 1 December 2015, Accepted 3 December 2015
Correspondence: Jung-in Kim (madangin@gmail.com)
Tel: 82-2-2072-3573, Fax: 82-2-765-3317

numbra by adopting double-focused multi-leaf collimators (MLCs).[3] To compensate low dose-rate of Co-60, ViewRay® system uses a total of three Co-60 sources spaced 120° apart in a ring-type bore. The combined dose-rate of three Co-60 sources is 550 cGy/min which is comparable with that of commercial linac. Several studies already reported clinical feasibility of ViewRay® system with the results of patient treatment using ViewRay® system.[4-6]

Since the ViewRay® system is a unique system integrated with MR system and Co-60 based radiation therapy system, not much experiences in the society were accumulated yet. Moreover, since the ViewRay® system is a new type of radiotherapy machine of which design is much different from the conventional radiotherapy machines, careful commissioning procedure should be performed before treating patients.[3] Recently, we installed and commissioned the ViewRay® system in our institution, therefore, we report the results of commissioning of the ViewRay® system. We commissioned the ViewRay® system by checking the 1) safety performance of the machine, 2) mechanical accuracy, 3) imaging system performance and 4) radiation therapy system performance.

Materials and Methods

1. Machine description

The brief description of the ViewRay® system is shown in Fig. 1. The source to axis distance (SAD) of the ViewRay® system is 105 cm and the diameter of bore is 70 cm. The ViewRay® system equipped with a total of 3 identical MLC systems for each Co-60 source. Since the maximum capacity of each Co-60 source for ViewRay® system is 15,000 Ci, the initial activities of sources were approximately 15,000 Ci which can provide a dose-rate approximately 200 cGy at the isocenter at the depth of dose maximum (d_{max}) with the maximum field size. Each source locates 120° apart in the ring-type gantry. The leaf width of MLC is 1.05 cm at the isocenter and the maximum field size is 27.3 cm×27.3 cm. Each double-focused MLC has a total of 60 leaves, *i.e.* two opposing banks of 30 leaves. Since there is no X- or Y-jaws, collimation is done by only MLCs. With each MLCs, the ViweRay® system can perform step and shoot intensity modulated radiation therapy (IMRT). For planning and daily patient positioning, volumetric MR image can be acquired with 0.35 T static magnetic field which is generated with superconducting magnet. For gating, single sagittal MR image can be acquired with 4 frame/sec or 3 sagittal images can be acquired with 2 frame/sec during treatment. Collimator and couch rotation are not possible for ViewRay® system.

2. Safety check

The functions of safety interlocks of the system were verified. Proper operations of door interlock, emergency stop button, beam-on warning light, treatment room radiation monitoring system, audio and video system between treatment room and control room and safety function of the machine when occurring power-off were checked. We surveyed radiation leakage of the sources in the treatment room. In addition, we measured MLC transmission with EBT3 Film (Ashland

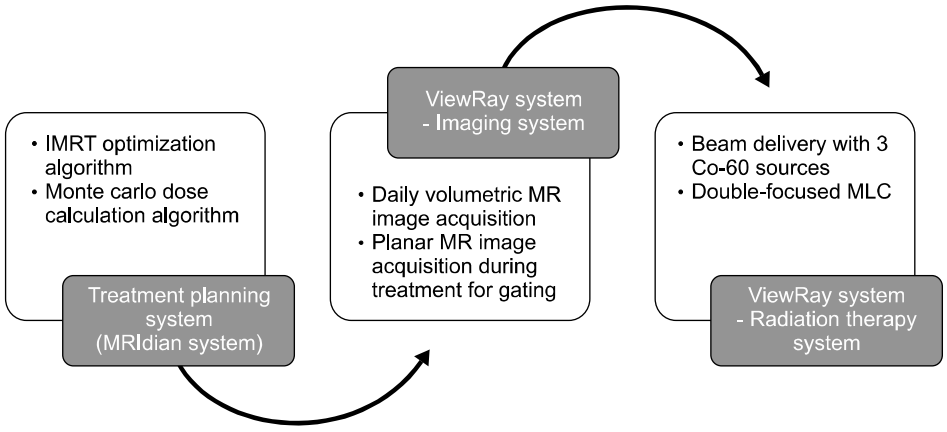

Fig. 1. A brief description of the ViewRay system.

Specialty Ingredients, NJ, USA).

3. Imaging system check

For the MR imaging system, we verified spatial integrities of the MR images of axial, coronal and sagittal planes with a phantom provided by ViewRay Inc. Image contrast and image resolution were verified using American College of Radiology (ACR) magnetic resonance image (MRI) phantom (Newmatic Medical, Chicago, IL, USA). Image uniformity and signal to noise ratio (SNR) were verified using spherical 24 cm diameter phantom (Siemens, Erlangen, Germany). Image uniformity was calculated as follows.[7]

$$\text{Uniformity } (\%)=100\times\left(1-\frac{(ROI\ Signal_{max}-ROI\ Signal_{min})}{(ROI\ Signal_{max}+ROI\ Signal_{min})}\right) \quad (1)$$

where, $ROI\ Signal_{max}$=the maximum signal in the region of interest (ROI) and $ROI\ Signal_{min}$=the minimum signal in the ROI.

The SNR was calculated as follows.[7]

$$\text{SNR}=\frac{0.66\times ROI\ Signal\ mean\ value}{ROI\ Noise\ standard\ deviation} \quad (2)$$

4. Mechanical accuracy of the machine

For couch performance check, verification of couch level and orthogonality of the couch movement with respect to the imaging plane with ViewRay Daily QA phantom (ViewRay Inc. Cleveland, OH, USA). Accuracy of couch positon indicator was verified with ruler. When performing couch repositioning, the coincidence between initial positon and position after couch repositioning was checked based on MR images with ViewRay Daily QA phantom. Since ViewRay system has a total of 3 sources, the isocenters of each source should be coincident. We verified the coincidence of isocenters of radiation therapy system using ViewRay Daily QA phantom with EBT3 film. The irradiated film was analyzed using RIT113 software (Radiological imaging technology, CO, USA). In addition, since ViewRay system has a total of 3 types of isocenters which are imaging system isocenter, radiation beam isocenter and virtual isocenter defined by room laser outside the bore, we verified the coincidence of those 3 isocenters. The verification of coincidence between imaging system isocenter and virtual isocenter has been done with ViewRay Daily QA phantom while that between radiation beam isocenter and virtual isocenter was done with MR-compatible IC profiler (Sun Nuclear Corporation, Melbourne, FL, USA). Mechanical accuracy of MLCs and alignment of MLCs were verified with solid water phantom and EBT3 film. To check MLC beam alignments, we delivered 1 cm×13.65 cm (Y1=0 and Y2=13.65 asymmetric beam) at the gantry angle of 90° and 1 cm×13.65 cm (Y1=−13.65 and Y2=0 asymmetric beam) at the gantry angle of 270° to the EBT3 film which was sandwiched between 1 cm thickness solid water phantoms. The solid water phantoms with EBT3 film were setup to be vertical to the beam direction. If MLCs were aligned perfectly, a continuous line-pattern irradiation could be acquired. If not, a discontinuity in the irradiated pattern at the isocenter would be acquired. The MLC positioning accuracy test was done following a protocol provided by manufacturer with a wire jig phantom provided by manufacturer and EBT3 film.[7] Field size accuracy was checked by comparing calculated field size in the treatment planning system (TPS) to the measured field size with EBT3 film. We verified gantry angle indicator accuracy using MR-compatible ArcCHECK (Sun Nuclear Corporation, Melbourne, FL, USA). Timer accuracy was verified with stop watch and shutter timer error was checked using an equation by Attix as follows.[8]

$$\delta=\frac{(R1*240-R2*30)}{R2-R1} \quad (3)$$

where, $R1$=reading of beam irradiation for 30 sec and $R2$=reading of beam irradiation for 240 sec.

5. Radiation therapy system check

Reference dosimetry was performed with MR-compatible Exradin A12 ionization chamber (Standard Imaging Inc., Madison, WI, USA) which had been calibrated before measurements at the secondary standards dosimetry laboratory (SSDL). An in-house water phantom specially designed for MR environment was used for reference dosimetry. For reference dosimetry, American Association of Physicists in Medicine (AAPM) task group 51 (TG-51) protocol was used and the measured percent depth doses (PDDs) were compared to BJR supplement 25.[9] Since head 2 cannot go to the gantry

angle of 0°, reference dosimetry using water phantom was performed for head 1 and 3 only. After that, the relative readings of all the sources of each head were compared with solid water phantom at gantry angle of 90° which is an angle that every source can be located. Output factors were measured with various field sizes using Exradin A28 chamber (Standard Imaging Inc., Madison, WI, USA) and those measurements were compared to the calculated values using the TPS. The consistency of field sizes between gantry angle of 0° and 90° were verified with EBT3 film. The attenuation by couch at various gantry angles were verified with A28 chamber inserted in the ViewRay Daily QA phantom. Similarly, the attenuation by coils were verified and compared to the calculated value in the TPS. The output consistency at various gantry angles were verified with A28 chamber and ViewRay Daily QA phantom. Finally, we verified dosimetric consistency when performing gating based on real-time MR images with motion phantom.

6. End to end (E2E) test

We generated a total of 8 IMRT plans in the TPS (4 head and neck cancer, 2 liver cancer and 2 prostate cancer patients) and performed pre-treatment patient-specific quality assurance (QA) with ArcCHECK. For each patient, we measured dose distributions with 3 head mode (full use of all the 3 heads) and 2 head mode (head 1 disabled, head 2 disabled and head 3 disabled). With measured and calculated dose distributions, we performed global gamma evaluation with 3%/3 mm gamma criterion (threshold value=10%).[10] In addition, we compared gamma passing rates with and without beam interruption during IMRT delivery.

Results

1. Safety check

The door interlock function, every emergency stop button function, beam-on warning light, room radiation monitoring system, audio and video system operated properly. During power-off, couch was retracted automatically and Co-60 sources moved automatically in the safe positions. The treatment door can be opened during power-off. The radiation leakages at each source container were less than 2 mR/hr. The maximum and average radiation leakage of MLC were 0.844% and 0.101%, respectively, which were less than tolerance values of 1% and 0.375%.

2. Imaging system check

The values of SNR and uniformity with every coil of the ViewRay® system were shown in Table 1. Since each value of SNR and uniformity met the tolerance level provided by manufacturer, imaging system performance of ViewRay® system in our institution was acceptable.[7] For imaging spatial integrity test, every tested point within 10 cm radius about the isocenter showed deviations less than 1 mm in the axial, sagittal and coronal planes. Every point within 17.5 cm radius showed deviations less than 2 mm in every plane which were axial, sagittal and coronal planes. Since the tolerance level provided by manufacturer was 100% and 90% passing rates within radii of 10 cm and 17.5 cm, respectively, the results were acceptable.

In the cases of image contrast and spatial resolution test, the number of feature line which was visible was 24 and the identifiable line thickness was 0.9 mm (tolerance level=18 and 0.9 mm, respectively).

3. Mechanical accuracy of the machine

Deviations in the x, y and z directions, *i.e.* transverse, longitudinal and vertical directions, when registering MR image of ViewRay daily QA phantom acquired at the closest location to the machine from the MR image acquired at the furthermost end to the machine were 0.03 mm, 0.06 mm and 0.03 mm, respectively (tolerance level=2 mm). The average deviations be-

Table 1. The values of signal to noise ratio (SNR) and image uniformity.

	Axial	Sagittal	Coronal
Body coil			
SNR	15.38	15.36	15.41
Uniformity (%)	71.58	74.17	72.83
Torso coil			
SNR	52.14	48.65	43.43
Uniformity (%)	74.46	76.01	82.54
Head and neck coil			
SNR	46.07	46.56	41.16
Uniformity (%)	77.36	75.98	81.35

tween the values of couch indicator and ruler in the x, y and z directions were 0.2 mm, 0.1 mm and 0.2 mm, respectively. For couch repositioning accuracy test, the deviations in the x, y and z directions were 0.25 mm, 0 mm and 0.25 mm, respectively (tolerance level=1 mm).

For the coincidence between radiation system isocenter and virtual isocenter, deviations in the x, y and z directions were 0 mm, 0 mm and 1.3 mm for head 2, respectively. For the other heads, deviations were less than 1 mm. Therefore, we re-aligned MLC positions of head 2 in the vertical directions by 1 mm. After re-alignment, the deviations were 0 mm, 0 mm and 0.3 mm, respectively. The deviations between virtual isocenter and imaging isocenter in the x, y and z directions were 0.6 mm, 0.5 mm and 0.2 mm, respectively. After re-aligning of MLCs, the deviations between imaging system and radiation isocenter were 0 mm, 0.2 mm and 0 mm, respectively. In the case of MLC alignment test, the offsets of the center line of the irradiated area between left side and right side for head 1, 2 and 3 were 0.3 mm, 0.1 mm and 0.12 mm, respectively (tolerance level=2 mm). The results of MLC positioning accuracy and reproducibility is shown in Table 2. The maximum and average deviations between calculated and measured field sizes was 0.8 mm and 0.1 mm, respectively. The maximum and average deviations of penumbrae was 0.19 mm and 0.7 mm, respectively. In the case of gantry angle indicator accuracy, the maximum and average deviations were 0.2° and 0.06°, respectively. Timer error was less than 0.06 sec and shutter timer error was less than 0.0056 sec.

Table 2. Multi-leaf collimator (MLC) positioning accuracy.

MLC	Gantry angle (°)	Deviation (cm)
Head 1	0	0.02±0.01
Head 3	0	−0.04±0.02
Head 1	90	−0.01±0.02
Head 2	90	0.03±0.01
Head 3	90	−0.05±0.02
Head 2	270	−0.03±0.01
Head 3	270	−0.02±0.02

Table 3. Differences (%) in percent depth doses (PDDs) between measured values and the values provided by BJR supplement 25.

Depth (cm)	4.2 cm× 4.2 cm	6.3 cm× 6.3 cm	10.5 cm× 10.5 cm	12.6 cm× 12.6 cm
0.5	0.0	0.0	0.0	0.0
1	0.2	0.7	0.6	0.5
5	−0.2	0.2	0.3	−0.1
10	−0.6	0.2	0.0	−0.2

4. Radiation therapy system check

The absolute output deviations measured with Exradin A12 chamber according to AAPM TG-51 protocol were −0.27% for head 1, −0.23% for head 2 and −0.45% for head 3.[11] The differences between PDDs calculated in the TPS and PDDs provided by BJR supplement 25 are shown in Table 3.[9] Those differences were less than 1%. The differences between PDDs measured in water phantom and PDDs calculated in the TPS for head 1 and 3 are shown in Table 4. Those differences were also less than 1%. Since head 2 cannot placed at the gantry angle of 0°, PDDs of head 2 couldn't be measured with water phantom. We compared PDDs of head 1, 2 and 3 in the solid water phantom, which were measured at the gantry angle of 90° and the deviations of PDDs were less than 1%. The differences in output factors between calculated and measured were shown in Table 5, which were less than 2% (tolerance level=2%). The maximum deviations in the field size between gantry angles of 0° and 90° for head 1, 2 and 3 were 0.7 mm, 0.43 mm and 0.34 mm, respectively. The maximum difference between calculated and measured dose due to couch attenuation was observed at the gantry angle of 180° and the value

Table 4. Differences (%) in percent depth doses (PDDs) between measured values and calculated values in the treatment planning system (TPS) for head 1 and 3.

Depth (cm)	4.2 cm× 4.2 cm	10.5 cm× 10.5 cm	21 cm× 21 cm	27.3 cm× 27.3 cm
		Head 1		
0.5	0.00	0.00	0.00	0.00
1	0.50	−0.37	−0.11	−0.09
5	0.25	−0.11	−0.19	0.55
10	0.48	0.19	0.45	0.99
		Head 3		
0.5	0.00	0.00	0.00	0.00
1	0.42	−0.53	−0.30	−0.19
5	0.31	−0.09	−0.30	0.52
10	0.37	0.05	0.37	0.90

Table 5. Deviations in output factors (%) between measured values and calculated values in the treatment planning system (TPS).

Field size	Head 1 (%)	Head 2 (%)	Head 3 (%)
2.1 cm×2.1 cm	−0.211	−2.230	−3.379
4.2 cm×4.2 cm	−1.269	−1.452	−1.341
6.3 cm×6.3 cm	−0.710	−0.770	−0.775
8.4 cm×8.4 cm	−0.002	0.012	0.007
10.5 cm×10.5 cm	0.000	0.000	0.000
12.6 cm×12.6 cm	−0.062	−0.201	−0.026
14.7 cm×14.7 cm	0.301	0.021	0.334
16.8 cm×16.8 cm	0.680	0.170	0.674
21 cm×21 cm	0.618	−0.366	0.487
27.3 cm×27.3 cm	1.373	−0.331	1.361

was 1.7%. The maximum difference between calculated and measured dose due to head and neck coil and torso coil were 0.34% and 0.42%, respectively. Beam attenuation by coil was approximately 1%. The output errors due to gantry angle were less than 0.7%. The difference between delivered dose during gating based on real-time MR images using motion phantom and calculated dose was 0.54%.

5. E2E test

The gamma passing rates of IMRT plan with and without beam interruption during beam delivery were 99.8% and 99.7%, respectively, showing no change in gamma passing rates due to beam interruption. The average gamma passing rates with 3 head mode (no head was disabled) was 99.9%± 0.1%. Those values of 2 head mode, which were head 1, 2 and 3 was disabled, were 99.9%±0.3%, 99.6%±0.7% and 99.7%±0.4%, respectively. The minimum value of gamma passing rates was 98.7% when the head 3 was disabled and delivering head and neck IMRT plan.

Discussion

For ViewRay® system, user doesn't have to input beam data into the TPS for commissioning the TPS since dose calculation algorithm for ViewRay® system is based on authentic Monte Carlo calculation algorithm and the machine geometry is already modeled by manufacturer. If some deviations between the calculated values from TPS and actual machine performance were observed, actual machine calibration is performed not the TPS calibration change. Therefore, during commissioning period, we checked machine performance compared to the results acquired from TPS. As shown in the results, the deviations between performance of actual machine and values shown in the TPS were less than tolerance level which was provided by manufacturer or based on international protocol such as AAPM TG-40 or TG-142.[7,12,13] Therefore, the ViewRay® system installed in our institution was appropriate to be used to treat patients. Moreover, the results of E2E test, which were gamma passing rates with ArcCHECK, were very high, therefore, it seems that the ViewRay® system's IMRT technique is appropriate to be used in the clinic.[10]

Since design of the ViewRay® system is much different from those of conventional linacs, careful consideration is needed for commissioning the system. For example, there is no cross-hair as well as light field for ViewRay® system. The design of ViewRay® system is ring-type, therefore, some dosimetric devices couldn't be used due to the bore size which is 70 cm diameter. In addition, most of conventional dosimetric devices cannot be used due to static MR environment of the ViewRay® system. Up to date, since there is a limitation in the use of dosimetric devices which are compatible for ViewRay® system, some deivces should be manufactured at the site to commission the system.

One of the uniqueness of the ViewRay® system is that there are multiple isocenters. Since ViewRay® system uses a total of 3 heads with 3 Co-60 sources, those isocenters should be coincident. In addition, TPS cannot recognize each source, *i.e.* TPS considers every source and every MLCs are identical, therefore, the activity of the source and each MLC performance should be similar as possible. Besides of 3 isocenters of radiation therapy system, imaging system isocenter and virtual isocenter exist in the ViewRay® system. Since the design of ViewRay® system is ring-type, patient setup should be done outside of the bore. Therefore, initial patient setup should be done with virtual isocenter which is defined by room laser. Especially for phantom or dosimetric device setup such as ArcCHECK or IC profiler, virtual isocenter accuracy is important since those dosimetric device cannot be imaged with MR system. The imaging isocenter also should be coincident with both virtual isocenter and radiation therapy system isocenter. If imaging isocenter is different from the isocenter

of radiation therapy system, whole treatment could be wrong.

Conclusion

The biggest benefit of the ViewRay® system is that it is the first MR-IGRT system in the world as well as we can see patient's anatomy including tumors and OARs during treatment. We commissioned the ViewRay® system in our institution recently. As shown in the results, it seems to be appropriate to use ViewRay® system for an accurate and precise image-guided radiotherapy. Since the design of the ViewRay® system is quite unique and it is operated under MR environment, available dosimetric devices and dosimeters are limited up to date. Therefore, some devices should be manufactured at the site for commissioning as well as periodic quality assurance (QA) of the ViewRay® system. Therefore, careful consideration is needed for the use of the ViewRay® system since not enough experience was accumulated in the society. We hope that our commissioning experience would be helpful for the society.

References

1. **Stam MK, van Vulpen M, Barendrecht MM, et al**: Kidney motion during free breathing and breath hold for MR-guided radiotherapy. Phys Med Biol 58(7):2235-2245 (2013)
2. **Seierstad T, Hole KH, Saelen E, et al**: MR-guided simultaneous integrated boost in preoperative radiotherapy of locally advanced rectal cancer following neoadjuvant chemotherapy. Radiother Oncol 93(2):279-284 (2009)
3. **Mutic S, Dempsey JF**: The ViewRay system: magnetic resonance-guided and controlled radiotherapy. Semin Radiat Oncol 24(3):196-199 (2014)
4. **Wooten HO, Green O, Yang M, et al**: Quality of Intensity Modulated Radiation Therapy Treatment Plans Using a Co-60 Magnetic Resonance Image Guidance Radiation Therapy System. Int J Radiat Oncol Biol Phys 92(4):771-778 (2015)
5. **Wooten HO, Rodriguez V, Green O, et al**: Benchmark IMRT evaluation of a Co-60 MRI-guided radiation therapy system. Radiother Oncol 114(3):402-405 (2015)
6. **Kishan AU, Cao M, Wang PC, et al**: Feasibility of magnetic resonance imaging-guided liver stereotactic body radiation therapy: A comparison between modulated tri-cobalt-60 teletherapy and linear accelerator-based intensity modulated radiation therapy. Pract Radiat Oncol 5(5):330-337 (2015)
7. **Rajiv L, Gopi K, Sathish K**: Acceptance Test Procedure for MRIdian System. ViewRay Inc., Cleveland, OH (2015)
8. **Attix FH**: Introduction to radiological physics and radiation dosimetry. 1st ed, Wiley, New York, NY (1986)
9. **Day M, EGA A**: Central axis depth dose data for use in radiotherapy: A survey of depth doses and related data measured in water or equivalent media. Brit J Radiol Suppl 17:1-147 (1983)
10. **Ezzell GA, Burmeister JW, Dogan N, et al**: IMRT commissioning: multiple institution planning and dosimetry comparisons, a report from AAPM Task Group 119. Med Phys 36(11): 5359-5373 (2009)
11. **Almond PR, Biggs PJ, Coursey BM, et al**: AAPM's TG-51 protocol for clinical reference dosimetry of high-energy photon and electron beams. Med Phys 26(9):1847-1870 (1999)
12. **Klein EE, Hanley J, Bayouth J, et al**: Task Group 142 report: quality assurance of medical accelerators. Med Phys 36(9):4197-4212 (2009)
13. **Kutcher GJ, Coia L, Gillin M, et al**: Comprehensive QA for radiation oncology: report of AAPM Radiation Therapy Committee Task Group 40. Med Phys 21(4):581-618 (1994)

자기공명영상유도 Co-60 기반 방사선치료기기의 커미셔닝 경험

*서울대학교병원 방사선종양학과, †서울대학교병원 의생명연구원, ‡서울대학교 의학연구원 방사선의학연구소,
§차세대융합기술연구원 로봇융합연구센터, ‖서울대학교 의과대학 방사선응용생명과학 협동과정,
¶서울대학교 의과대학 방사선종양학교실

박종민*†‡§ · 박소연*†‡‖ · 우홍균*†‡¶ · 김정인*†‡§

본 연구는 뷰레이 시스템의 커미셔닝 결과에 대한 보고이다. 먼저, 시스템 안전장치의 적절한 작동을 확인했다. 영상시스템에 대한 평가를 위해 신호 대 잡음비와 영상의 균질도, 공간적 무결성을 확인했다. 카우치 동작의 정확성 및 축교점의 일치성을 평가했다. 미국의학물리학회 특별업무단51규약 프로토콜에 따라 절대선량을 측정했다. BJR supplement 25에서 제공하는 심부선량백분율과 측정한 값의 차이, 치료계획에서 계산한 값과 측정한 심부선량백분율의 차이를 확인했다. 더불어, 출력인수에 대하여, 측정값과 계산값의 차이를 구했다. 최종 검증 단계로, 8개의 세기변조방사선치료계획을 사용하여 감마평가를 수행하였다. 커미셔닝을 수행한 결과, 모든 안전장치는 적절히 구동함을 확인했다. 신호 대 잡음비 값과 영상 균질도 값은 허용범위 이내임을 확인했다. 공간적 무결성 확인 결과, 반지름 10 cm 및 17.5 cm 안의 모든 지점에 대하여 각각 1 mm 및 2 mm 이내의 오차를 확인했다. 카우치는 x, y, z 방향으로 각각 0.2 mm, 0.1 mm, 0.2 mm의 오차를 보였다. 방사선 축교점과 가상 축교점 사이에는 x, y, z 방향으로 0 mm, 0 mm, 0.3 mm의 오차를 보였다. 영상시스템의 축교점과 가상 축교점 사이에는 0.6 mm, 0.5 mm, 0.2 mm의 오차를 보였다. 다엽콜리메이터의 평균적 구동 오차는 0.6 mm였다. 측정한 출력의 오차는 0.5% 이내, 심부선량백분율 오차는 1% 이내, 출력인수 오차는 2% 이내였다. 세기조절방사선치료 감마평가 결과값이 99.9%±0.1%였다.

중심단어: 자기공명영상유도 방사선치료 시스템, 커미셔닝, 코발트-60, 세기조절방사선치료