# Radiation Therapy in Cochlear Implant Recipients

*Michael S. Gossman, †Claudiu G. Treaba, and †Jonathon R. Kirk

*Radiation Oncology Department, Tri-State Regional Cancer Center, Ashland, Kentucky; and †Cochlear,
Denver Research and Technology Labs, Denver, Colorado, U.S.A.*

**Hypothesis:** Processes of scattering and attenuation were investigated to determine the consequence on dose distributions by having a cochlear implant in the field of therapeutic radiation.
**Background:** Radiation oncology medical accelerator beams of 6- and 18-MV x-ray energy were used. Five cochlear implants were investigated.
**Methods:** Each implant model was individually studied using computer dose modeling and through exercises in radiation measurement during live delivery.
**Results:** No side scatter was detected, and negligible backscattering was observed for the primary device housing and electrodes. Attenuation consequences were found to be dependent on the model of cochlear implant studied and specifically dependent on the material composition of each device.
**Conclusion:** The maximum attenuated dose change for the study was found to be −8.8% for 6 MV and −6.6% for 18 MV. This study presents the first comparison of therapeutic radiation delivery versus computerized treatment simulation involving cochlear implants. **Key Words:** Cochlear—Implant—Radiation—Therapy.

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Since the first commercial multichannel cochlear implant was marketed by Cochlear in 1985 (1), use of the device has grown exponentially. As of mid-2010, more than 188,000 people worldwide have received a cochlear implant system (2). Recent developments in the manufacturing of cochlear implants by the market leader, Cochlear Limited (based in Australia), have resulted in an aim to provide electronically stable models (3). Studies are currently underway to determine the consequences and clinical affectability in having these devices in the field of therapeutic radiation. Included in this effort are attempts to design models capable of withstanding high-energy x-ray beams from a clinically used high-energy particle accelerator (4–6). Currently, only 4 published articles have focused on the influence of radiation therapy on cochlear implant function in vitro (4,7–9). In parallel to the interest, researchers are interested in determining and documenting the converse effect to the radiation beam itself, where scattering and attenuation is hypo-thesized but unknown. This marks the first therapeutic radiation simulation and measurement publication involving cochlear implants at clinically relevant megavoltage x-ray energies.

## MATERIALS AND METHODS

Five Cochlear Nucleus implants are included in this study. These are the models Nucleus 22, Nucleus 24, Nucleus 24 ABI, Nucleus 24 k, and Nucleus CI512. These are illustrated in Figure 1. As one can tell, these devices come in a variety of sizes, each with a distinct construction and shape. The cochlear implant has a flexible silicone overmold that contains a magnet in the center and a receiver coil that surrounds it along the inner rim of the silicone overmold. The receiver coil connects to the receiver-stimulator, which houses complex electronics. These include capacitors, resistors, transformers, and an application-specific integrated circuit placed on a printed circuit board. Hermetic connections between the electronics, receiver coil, and electrode array also are located here. An example of the construction is provided in Figure 2 for the Cochlear Nucleus CI512 specifically (the latest implant marketed by Cochlear, part of the Nucleus 5 system).

Many of these structures are metallic in composition. For Nucleus CI512, electrode wires and the receiver coil are made of platinum alloys. Specifically, the coil is refined platinum, whereas electrode wires are platinum-iridium. The housing (receiver/stimulator package) is made of titanium. The feedthrough is mostly aluminum. All Cochlear Nucleus implants use rare-earth magnets. Other models may have varying changes in materials of the application-specific integrated circuit on the printed circuit board or geometric changes in shape, but the
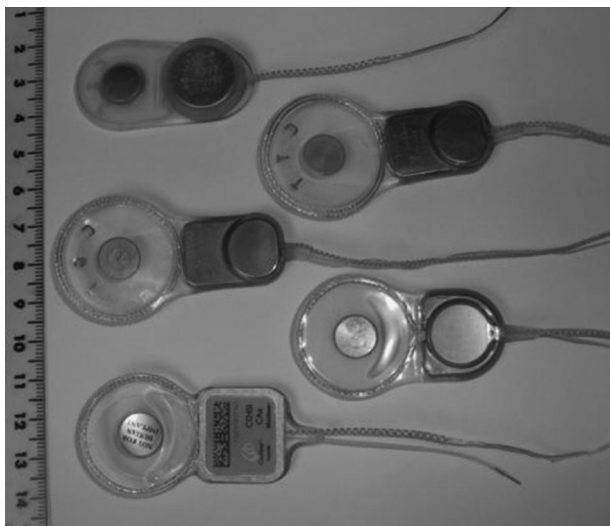
**FIG. 1.** Photograph of the Cochlear Nucleus implant models used in this study from top down: Nucleus 22, Nucleus 24, Nucleus 24 ABI, Nucleus 24 k, and Nucleus Cl512 (scale in centimeters).

basic components remain constant. These objects stated are of considerable importance for study here because they tend to scatter x-ray radiation and attenuate it, causing computed tomographic (CT) scan artifacts and can further induce dose delivery changes in radiation therapy beams, even in the megavoltage x-ray range.

### CT Scan Acquisition Process

A phantom was used to simulate a patient having a cochlear implant. The phantom design included a water tank and water-equivalent plastic media plates. Two CIRS (Norfolk, VA, USA) model PW-3050 Plastic Water pieces, each having an area of $30 \times 30$ cm$^2$ $\times$ 5 cm thickness, were first positioned on the couch of the CT scanner for backscatter. A CNMC Company, Inc. (Nashville, TN, USA) model WP-3040 water tank with dimensions 40 cm wide $\times$ 40 cm long $\times$ 38 cm high was then placed on top. The phantom is constructed with each side having 1-cm acrylic. The total amount of backscatter material is then 11 cm. The water tank was filled to a depth of 4 cm so that each implant could be entirely submerged. One by one, a cochlear implant was submerged and affixed to the bottom of the acrylic water tank for independent, consecutive scanning. Laser cross hairs assured that the center of all scans was at the centroid of the cochlear implant magnet, with stimulating electrodes exiting in the transverse plane as shown in Figure 1.

The LightSpeed RT CT scanner (General Electric, Fairfield, CT, USA) was provided in all imaging requirements. A helical mode stereotactic radiosurgery protocol was programmed for scanning each cochlear implant. The technique included an x-ray beam of 120 kVp, nominally 221 mA in 87.3 seconds, and a couch increment of 1.25 mm per slice. A field of view at 50 cm was used. All images were autogenerated using a commissioned extended Hounsfield unit CT range. Following all scanning, a total of 1,284 slices in 5 scans were networked for computer treatment modeling.

### Dose Simulation Process

Three-dimensional conformal radiation therapy treatment planning was performed using the Eclipse build, version 8.6

software (Varian Medical Systems, Inc., Palo Alto, CA, USA). Artifacts were identified around each implant, which occur as a consequence to improper sampling of attenuation coefficients for metals. Data processing was then conducted to avoid incorrect dose calculations as a result of these streaking artifacts seen on the image for metals observed within each cochlear implant. Contours of each implant model were first iteratively made using known factory specifications on the outer dimensions of each. Then, a Boolean operator was incorporated whereby all streaks and object material around the implant were removed and fixed to a Hounsfield unit value of 0, resembling waterlike media. Each cochlear implant maintained the Hounsfield unit values that were specific to the density of each component designed in their constructed. Although the Boolean operator was used to remove illusory scatter contributions from artifacts around the implant, scatter from the implant will still be evident and important.

The Anisotropic Analytical Algorithm, version 8.6.15 (Varian Medical Systems, Inc.) was the software used for treatment simulation modeling. The software makes use of measured radiation output data from a particle accelerator to calculate dose through the CT scan images. This software was commissioned to provide accurate dose assessment resembling treatments from 21EX high-energy particle accelerator (Varian Medical Systems, Inc.). Photon energies of 6 and 18 MV are specifically addressed here. The software reflects the accelerator calibration for dose output capable of running at 600 cGy per minute at the depth of maximum dose, while at the center of machine rotation. For calibration purposes, the depth in water providing maximum dose was nominally 1.5 cm at 6 MV and 2.5 cm at 18 MV. Varian standard scaling geometry was used, where the gantry angle, couch angle, and collimator were each at 180 degrees, aiming the beam straight down to the phantom. The beam size for calibration was a square $10 \times 10$ cm$^2$ field.

Treatment plan modeling was conducted here specifically with the CT data from previous phantom scanning. A unique simulation plan was created for each implant because CT data are specific for each implant. For each of the 5 unique plans, a single anterior field with a $30 \times 30$ cm$^2$ aperture was assigned and aligned to pass centrally through the magnet. A small dose calculation grid of 1.25 mm was assigned. The 4-cm water depth in the phantom ensured adequate buildup to maximum dose as calibrated.

Calculation points of interest were placed throughout the coronal plane (magnet-electrode plane) of each cochlear implant anterior and posterior by 1 cm away. Careful attention was made to ensure points were positioned below and above the locations
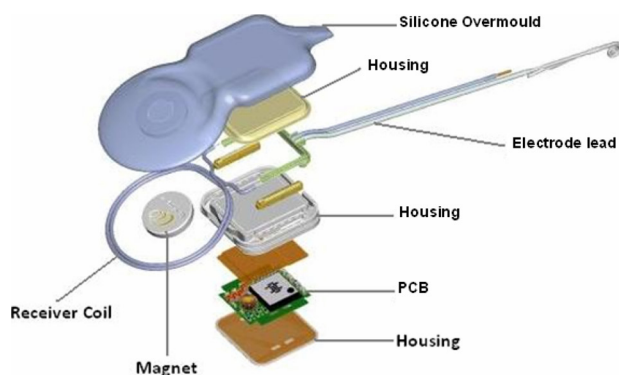


**FIG. 2.** Exploded view of the Cochlear Nucleus model Cl512.

**TABLE 1.** *Computerized treatment simulation results for all Cochlear Nucleus implants*

Computer simulation

| Model | Magnet side (%) 6-MV attenuation | Magnet side (%) 18-MV attenuation | Feed-through side (%) 6-MV attenuation | Feed-through side (%) 18-MV attenuation |
|---|---|---|---|---|
| Nucleus 22 | −7.5 | −3.9 | −4.4 | −2.2 |
| Nucleus 24 | −6.6 | −4.0 | −5.4 | −3.0 |
| Nucleus 24 ABI | −6.6 | −4.0 | −5.0 | −2.9 |
| Nucleus 24 k | −6.9 | −4.4 | −7.3 | −4.2 |
| Nucleus Cl512 | −6.5 | −4.1 | −2.5 | −1.4 |

determined from CT data to be of high-density metal. Points of interest also were set laterally next to the magnet casing, extracochlear electrode, and electrode connection terminal at 1 cm for each. Dose reduction (attenuation) is expected as radiation passes through the metals in the implant. Dose increase is expected anterior to the implant (backscattering) and lateral to the implant (side scattering) as a result of x-rays being deflected from the metal. Therefore, the calculation points were set to establish the level of change exhibited in each process.

Software should be used for studies such as these that allow the user to determine the dose to the medium being irradiated by either accounting for density changes (heterogeneity correction) or assuming the medium is entirely water (homogeneity). The results from each dose calculation method reveal the consequence of having an object, such as a cochlear implant, in the radiation field. Strictly, the ratio of the results from these 2 plans was sought after for data analysis. This method is consistent with published research and American Association of Physicists in Medicine formal guidance for simulating dose involving high–atomic number materials (10,11). The axial, coronal, and sagittal planes of the 3-dimensional software detail the shape the resulting dose distribution with incremental isodose lines that are color coded.

A total of 4 plans were created for each cochlear implant model. Two plans were created with and without heterogeneity correction in the way of 6-MV x-rays as well as identically for 18-MV x-rays. Five total cochlear implants were studied. Therefore, 20 simulation plans were created for analysis.

### Radiation Measurement Process

The phantom setup used for CT scan acquisition was duplicated for measurement. The detection system included a 192 electrometer (Capintec, Inc., Ramsey, NJ, USA) and a water-resistant TN31014 miniature thimble ionization chamber (PTW, Freiburg, Germany) having a 0.015 cm$^3$ sensitive volume. The chamber center pin was equilibrated to nominally +300 V for highest ionization collection efficiency. Attenuation measurements require the detector to be placed posterior to the implant at the location simulated to create the most change. With the detector placed at the bottom of the tank, an MT-CB-410S bolus (CIVCO Medical Solutions, Kalona, IA, USA) with a thickness of 30 × 30 cm$^2$ × 1.0 cm was placed directly on top of it. Then, the cochlear implant was positioned immediately on top of the rubber bolus material. Backscatter requires the detector to rest 1 cm anterior to the implant. For this exercise, the chamber was first inserted into the tank, with the bolus and finally the detector to follow. Side-scatter measurements were conducted similarly, with the bolus and detector at the side of the implant. Again, these specific points were prediscovered from the simulation modeling on a computer. Numerous measurements were taken

at each location. Radiation measurements were conducted consecutively at 6 and 18 MV and for each implant model studied.

## RESULTS

Computerized simulation of treatment revealed no discernable difference in dose when the extracochlear electrodes were in the path of radiation for any of the physical processes. No side scatter or backscatter was seen surrounding the device at any location. However, considerable changes were discovered because of the process of attenuation, both on the magnet side and on the feed-through side. These results are provided in Table 1.

Radiation measurements concluded that no discernable change in dose was exhibited at either x-ray energy as a result of extracochlear electrodes being in the field of radiation. No side-scattering dose variance around each implant was determined measurable. Contrary to the results simulated, increased dose change was noted for backscatter anterior to some, but only remarkable to levels of maximally 0.6% at 6 MV and 0.3% at 18 MV. These dose changes were achievable only for the side containing the magnet. No measureable differences were determined on the feed-through side as a consequence of it causing increased scatter anteriorly. Consistent with simulated results, a considerable amount of attenuation dose change was detected. Results for measured attenuation levels are presented in Table 2.

## DISCUSSION

The methods described may be used as a guide for the medical physicist in modeling external beam radiation therapy treatments on computer for patients referred with cochlear stimulator already implanted. These data can be used to directly assist in the understanding of dose perturbations that have proven here to be observed consequential to the interaction of incident radiation with such high-density devices. With a flat water phantom being used along with flat and symmetric beam of radiation orthogonally directed to it, the impact of such a dense implant in the path of the beam is now appreciated. Figure 3 provides an illustration of the clinical significance of dose distribution shifting when the device was placed in the medium. Inhomogeneities in the dose

**TABLE 2.** *Radiation measurement results for all Cochlear Nucleus implant models*

Radiation measurements

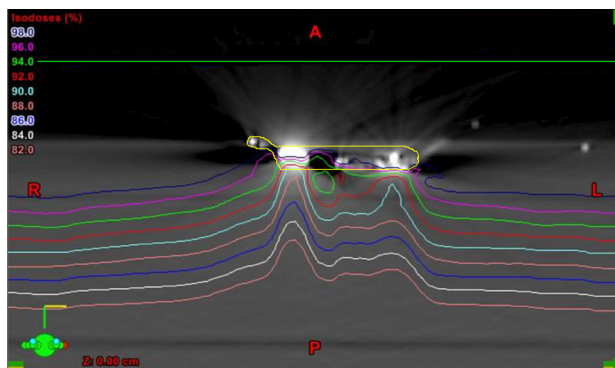| Model | Magnet side (%) 6-MV attenuation | Magnet side (%) 18-MV attenuation | Feed-through side (%) 6-MV attenuation | Feed-through side (%) 18-MV attenuation |
|---|---|---|---|---|
| Nucleus 22 | −8.8 | −2.7 | −5.1 | −1.1 |
| Nucleus 24 | −6.1 | −4.2 | −4.5 | −2.0 |
| Nucleus 24 ABI | −6.7 | −3.8 | −4.8 | −1.8 |
| Nucleus 24 k | −6.7 | −2.9 | −6.6 | −3.0 |
| Nucleus Cl512 | −5.5 | −2.5 | −2.3 | −0.5 |

**FIG. 3.** Cochlear implant model 24 k dose simulation for 6-MV x-rays in the axial plane.

distribution may compromise local control (11). There is always a cancer or target structure intended for treatment that exists in the patient's anatomy. A simulation plan for treating the patient would ideally cover this target structure, while optimally limiting the dose to local organs at risk well below the criteria designed by the radiation oncologist. Cochlear implants have shown to greatly alter these isodose distributions, making it much more difficult to iteratively create the optimal plan. Noncoplanar beam arrangements to completely or partially avoid the implant have been proposed (12). Although these techniques often are the preferred solution, sometimes, these techniques are not the best choice because the dose to organs now directed to be at risk may be unacceptably high (13). Therefore, there are advantages and disadvantages of modifying treatment techniques within ordinary beam geometries to avoid excessive dose complications to organs at risk. Conformally planning with heterogeneity correction for cochlear implant patients should be technically managed by the medical physicist and carefully evaluated with the radiation oncologist before using the simulation model for actual therapy.

Continual research is needed for cochlear implant stimulators. Given that there is no formal guidance provided by the American Association of Physicists in Medicine or any other nationally accepted society currently, future investigations should involve the direct testing of devices for their operative ability to withstand radiation. Testing should include devices that represent each of the various families of cochlear implant models available. Clinical therapeutic environments for testing are highly recommended. In radiation oncology, particle accelerators are generally operated to emit 6- to 18-MV x-ray energies and at dose rates of up to 1,000 cGy/min. Therefore, cochlear devices should be tested in the range of dose rates from 300 to 1,000 cGy/min as used clinically, and at both bremsstrahlung energies. Likewise, the electronic integrity of the devices should be tested in each of these research arms at various levels of cumulative dose. Similar studies already published for heart rhythm devices may be impactful for developing benchmark data processes for cochlear implants (14).

## CONCLUSION

We have shown that computerized modeling can be used to qualitatively and quantitatively assist the radiation oncologist and medical physicist in recognizing attenuation effects from Cochlear Nucleus implant models in therapeutic x-ray beams from a medical accelerator. Dose changes were observed on computer within ±1.3% of results measured from an ionization chamber at all points and from both 6- and 18-MV x-ray beams. Dose change by attenuation was observed asymmetrically across the magnet-electrode plane of the cochlear implant. Greater dose change was seen to be generally caused by the magnet than by the electronic feed-through location.

We have shown that attenuation consequences are dependent on the model of Nucleus implant studied and specifically dependent on the material composition of these devices. Data suggests that a thinner implant design, such as that of the Nucleus CI512 is ideal because it induces less dose change. The maximum attenuated dose change for the study was found to be −8.8% for 6 MV and −6.6% for 18 MV.

Recommendations: 1) only use CT scanning that was commissioned for the extended Hounsfield unit range with computer-modeled dose delivery simulation calculations (15), 2) incorporate known implant device outer dimension information provided by the manufacturer in assisting manual contouring of the device in simulation software, 3) remove all streaking artifacts local to the cochlear implant device that may cause the resulting calculation to be inaccurate, 4) position reference points for computer prescriptions away from all high-density areas, 5) use algorithms for calculation that minimally have 3-dimensional convolution superposition capability for most accuracy in determining dose with high-density implants (16), 6) facilitate both homogeneous and heterogeneous calculations while comparing the differences observed for a better understanding of the magnitude and direction of profile shifts in the isodose distribution, and 7) finally, extrapolating computer results to correlate with the results here, which prove that simulation modeling can underestimate the true effect in comparison to radiation measurements.

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