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Thesis Outline

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In Chapter 1, “Introduction”, the current status of dosimetry for boron neutron capture therapy (BNCT), as well as the current status of BNCT and the trend towards accelerator-based BNCT is briefly discussed. Notably, there is currently no online dosimetry method for BNCT, and the current methods of dosimetry rely very heavily on Monte Carlo simulation, with the only measurement being a gold-wire activation to which the Monte Carlo simulation is normalized.

Prompt gamma single photon emission computed tomography, PG-SPECT, is a proposed online dosimetry imaging system for BNCT, which works on the principle of single photon emission computed tomography (SPECT), a common medical imaging technique in the X-ray energy region, being applied to measure the 478 keV prompt γ-ray produced by the $^{10}B(n,\alpha\gamma)^{7}Li$ reaction, the reaction which is the primary source of dose in a BNCT irradiation.

There are two primary goals in our research. First, we would like to experimentally prove proof-of-concept of PG-SPECT by creating an experimental version of PG-SPECT. Secondly, we would like to predict the performance of the system under clinical settings.

In accordance with our goals, our methodology is as follows: First, a hypothetical clinical design was created, and an experimental test version of that hypothetical design was constructed. Second, using the Monte Carlo particle simulation program PHITS, the probability of a 478 keV photon emitted in a given position depositing 100% of its energy in each detector element was calculated, for both conditions matching our experimental irradiation, as well as for conditions matching clinical settings. This matrix is hereby referred to as the C matrix, and is equivalent to the absolute efficiency for every position and every detector element. Third, a second Monte Carlo simulation, the “neutron irradiation simulation”, was conducted, which calculated the distribution of neutrons in the phantom, as well as where 478 keV photons are being produced in the phantom. Fourth, an experimental irradiation of the experimental test version was conducted, and the number of net counts in the 478 keV ROI, as well as the number of background counts in the 478 keV ROI were calculated for each detector position and angle. The number of net counts in the 478 keV ROI was compared to the predicted

number from the previous PHITS calculation. Using only the experimental data of the number of net counts in the 478 keV ROI, image reconstruction was conducted, proving proof-of-concept. Fifth, we combined the expected number of net 478 keV counts from $^{10}B(n,\alpha\gamma)^{7}Li$ from the Monte Carlo simulation in clinical settings with the expected number of background counts in the 478 keV ROI from the experimental data. Combining the expected net number of counts and background counts in the clinical setting, we were able to calculate the expected statistical fluctuations due to counting statistics, under clinical settings. Then, we plugged in random numbers according to the calculated expected statistical distribution, and used those to conduct image reconstruction under clinical settings. In this way, we were able to predict the performance of the system under clinical settings.

In Chapter 2, "Design", we discuss the theoretical PG-SPECT system for clinical use, as well as the experimental test version, and the design of the phantom. Both the theoretical clinical design as well as the experimental test design use a primary CdZnTe detector with a secondary BGO Compton-suppression anticoincidence detector, using 19 collimators per angle, lead as the collimator material and photon shielding, and polyethylene and ^{6}LiF as neutron shielding. The primary differences, however, are that in the clinical setting, there are 9 angles, each with 19 CdZnTe/BGO detector pairs, while in the experimental system, there is only a single CdZnTe/BGO detector pair, which is moved to 19 different positions to simulate having 19 detectors, and 4 angles are simulated by rotating the phantom. In the experimental test case, we used a phantom with a healthy tissue region made out of pure water contained in acrylic, with 5% ^{nat}B-borated polyethylene for the tumor region. For the theoretical clinical case, we used 10 ppm ^{10}B-borated water in the healthy tissue region, and 30 ppm ^{10}B-borated water in the tumor region, typical values in an actual BNCT procedure using current drugs.

In Chapter 3, "Experiment", we discuss the experimental irradiation of the system, as well as the layout of the system, and the electronics linking the detectors to the counting software. We discuss the procedure of conducting the experiment, as well as the neutron source, the $^{7}Li(p,n)$ reaction, using protons near threshold energy, a source that is expected to be common for accelerator-based BNCT in the near future. We set the phantom to the desired angle, and the detector pair to the desired position, and then measure the pulse-height spectrum for about 1 hour. Irradiation time slightly varied depending on detector position, angle, and allotted time remaining for the experiment. Notably, we measured the number of net 478 keV counts, as well as the number of background counts in the 478 keV ROI for each detector position and angle. Additionally,

we measured the number of $^{6}Li(n,\alpha)^{3}H$ counts in our neutron monitor, for the purposes of data normalization.

In Chapter 4, "Experimental Data Analysis", we discuss how the experimental data was analyzed, how the detectors were calibrated, how to properly convert the data from different runs into a single comparable format through rebinning, making corrections for BGO veto-time, and corrections for the number of counts on our neutron monitor. We show many graphs of various pulse-height spectra for different points, showing that the 478 keV photopeak is cleanly separated from the 511 keV annihilation peak. We also show the number of normalized net 478 keV counts for each detector position and angle, and the number of normalized background counts in the 478 keV ROI for each detector position and angle.

In Chapter 5, "Monte Carlo Simulations", we discuss the two separate Monte Carlo simulations, the C matrix calculation simulation, and the neutron irradiation simulation. In the C matrix calculation, the probability of a 478 keV photon produced in a given position depositing 100% of its energy in each detector element is calculated. This was done by placing an isotropic 478 keV photon source in the detector, and calculating the photon flux at every position around the phantom, and then using symmetries and normalization to convert this to the probability of a 478 keV photon emitted in a given position depositing 100% of its energy in a given detector position. Additionally, the reliability of this calculation through symmetry is verified.

In the neutron irradiation simulation, we calculate the distribution of neutrons and the distribution of 478 keV photon production in the phantom, in both the experimental and theoretical clinical case. By multiplying the 478 keV photon production by the C matrix, we can calculate the expected value for the number of net counts in the 478 keV ROI for each detector position and angle.

In Chapter 6, "Analysis", we compare the experimentally measured number of net counts in the 478 keV ROI with the calculated number of 478 keV full-energy deposition events from Monte Carlo. We reconstruct an image of 478 keV production in the experiment using only the experimentally measured number of net counts in the 478 keV ROI and the calculated C matrix. We finally combine the calculated number of 478 keV full-energy deposition events in the clinical case from Monte Carlo, and combine that with the expected number of background counts in the 478 keV ROI from Monte Carlo, to calculate the expected statistical distribution associated with measuring the number of net counts in the 478 keV under clinical settings, and then use a random number generation in conjunction with the calculated expected statistical distribution to show the effect of statistical fluctuations on the performance of image reconstruction

under clinical settings.

In Chapter 7, “Discussion”, we discuss the calculated amount of error associated with image reconstruction, the possible problem of BGO veto-time under clinical settings, and the effect changing the depth of the viewed slice.

In Chapter 8, “Conclusion”, we briefly summarize the results of our research, which is that we have experimentally proven proof-of-concept, and that we predict the performance of the PG-SPECT system to be accurate to within 9.2% under clinical settings.