EVALUATING AND IMPROVING PATIENT-SPECIFIC QA FOR IMRT DELIVERY

By

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To my beloved wife, Xiuyao and daughters, Maggie and Sarah
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Modern radiation therapy techniques such as intensity-modulated radiation therapy (IMRT) and newly-emerging volumetric modulated arc therapy (VMAT) aim to deliver highly conformal radiation dose to the target volume while sparing nearby critical organs as much as possible with the complex motion of multi-leaf collimator (MLC) leaves. Pre-treatment patient specific quality assurance (QA) has become an essential part of IMRT in making sure the delivered dose distributions agree with the planned ones. This dissertation evaluates the performance of current patient-specific QA process and proposes solutions to improve its sensitivity, accuracy and efficiency.

In step and shoot IMRT, the study on the sensitivity of patient-specific QA to minor MLC errors reveals tighter criterion such as 2%/2mm must be employed to detect systematic MLC positioning errors of 2 mm. However, such criterion results in low average passing rate which leads to excessive false alarms, mainly due to inadequate treatment planning system (TPS) beam modeling on beam penumbra. An analytical deconvolution approach is proposed to recover true photon beam profiles to obtain a true beam model which significantly improves agreement.
between calculated and measured dose distributions. Thus a tighter criterion could be employed to enhance the sensitivity of patient-specific QA to minor errors in the delivery system.

Measurement based patient-specific IMRT QA is a time-consuming process. A fast and accurate independent planar dose calculation algorithm is proposed to replace measurement based QA. The algorithm analytically models photons coming out from the accelerator and computes dose distribution from first principles. Accuracy of the algorithm is validated against 2D diode array measurements. The algorithm is found to be fast and accurate enough to replace time consuming measurement based QA.

Patient-specific QA for VMAT differs significantly from step and shoot IMRT due to the increased use of dynamic components (dynamic gantry and collimator, dynamic MLC and variable dose rate) in VMAT. A novel four dimensional (4D) diode array is developed by Sun Nuclear Corp for patient-specific VMAT QA. This work develops effective calibration procedures for this novel device by accounting for diode sensitivity and angular response dependence. A real time algorithm to derive gantry angle is developed to interpolate corresponding angular correction factors. Clinical applications of the diode array are demonstrated with IMRT as well as VMAT plans. Excellent agreement (>95% passing rate with 1%/2mm criterion) is achieved between diode array measurement and TPS calculation. The 4D diode array is proved to be a valuable tool for both IMRT and VMAT patient specific QA.
CHAPTER 1
INTRODUCTION

General Introduction

Radiation therapy uses ionizing radiation as part of cancer treatment to control malignant cells. It destroys or kills cancer cells in the targeted area ("tumor") by damaging their genetic materials and preventing them from growing and dividing. Although radiation can destroy both cancer cells and normal cells, most normal cells will recover from radiation damage due to regrowth process. It is crucial to deliver adequate dose to the tumor cells to achieve cancer killing while avoiding surrounding healthy tissue to minimize side effects.

Historically, maximum dose prescribed to the tumor is limited by the sensitivity and limits of nearby organ at risk (OAR) and thus the effectiveness of radiation therapy is reduced. A number of recent refinements and technologies have dramatically increased the effectiveness of external radiation therapy by shaping radiation to the target. Three-dimensional conformal radiation therapy (3D-CRT) and intensity modulated radiation therapy (IMRT) are among those new technologies. Three dimensional CRT utilizes modern computed tomography (CT) techniques to acquire a 3D image of the patient which allows the physician to delineate the tumor and critical organs precisely. Radiation beams are optimized to conform to the targets with jaws and multi-leaf collimators (MLC). Sufficiently high radiation dose could be delivered to the tumor while surrounding healthy tissues are largely spared. The effectiveness of 3D-CRT is significantly improved compared to traditional two dimensional (2D) treatments where only one or a few axial slices of the patient anatomy were acquired for planning. Three dimensional CRT is a dramatic change from traditional practice in that it utilizes tumor and critical structures delineated on multiple transverse CT images, radiation fields designed in beam’s eye view and volumetric dose calculation and evaluation tools.\(^1\)
IMRT is a further extension of 3D-CRT in that non-uniform fluences generated by the superposition of multiple MLC segments or dynamic MLC movements are used to improve dose conformity and OAR sparing. The modulated fluences are obtained from inverse planning and computer optimization with the guidance of clinical goals and objectives. It inherits all the new tools and methodologies from 3D-CRT. Previous radiation therapy techniques including 3D-CRT do not provide a method for sparing critical structures that either push into the target or are tightly surrounded by targets. IMRT does allow for highly conformal dose distribution with sharp dose gradient for complex target volumes with concave surfaces. This is accomplished by the complex motion of MLC leaves (Figure 1-1 a) equipped on a medical linear accelerator (LINAC). This advantage has been utilized to escalate dose to the tumor. As a consequence, each IMRT field has many small, irregular and asymmetric subfields defined by MLC leaves and are characterized by high dose gradient.

In IMRT delivery, the mechanical components that can be varied are gantry positions, collimator positions, MLC configurations, couch positions and dose rates. IMRT delivery techniques based on MLC can be typically classified into two categories: fixed gantry IMRT and dynamic gantry IMRT. If the gantry remains stationary when radiation beam is on, the delivery is referred to as fixed gantry IMRT. Otherwise, it is referred to as dynamic gantry IMRT. Fixed gantry IMRT includes sliding window techniques and segmental IMRT. Sliding window techniques allow MLC motion when beam is turned on while segmental IMRT turns off the beam when MLC is in motion (so called step and shoot delivery). In both sliding window techniques and segmental IMRT, gantry angles are usually selected manually with the guidance of beam’s-eye-view for optimal target coverage and OAR sparing (Figure 1-1 b). Both techniques in the fixed gantry IMRT category have gained wide application in clinical practice.
for more than a decade. In the category of dynamic gantry IMRT, a special type called volumetric-modulated arc therapy (VMAT) is rapidly proliferating recently in both academic and community practice due to its shorter treatment time and less monitor units than fixed gantry IMRT. It utilizes all the dynamic components (dynamic gantry and collimator, dynamic MLC and variable dose rate) to achieve these advantages.

IMRT represents one of the most exciting technology advancements in radiation therapy since the introduction of CT imaging in radiation treatment planning. It is a new treatment paradigm to deliver high doses of radiation to tumor while providing maximum sparing for nearby OAR. The obvious dosimetric advantage and increased reimbursement have fueled the speed of clinical dissemination of IMRT. A recent survey shows IMRT treatment was used by 73% of the responding radiation oncologists while the number was 32% in 2002. It has been estimated that 30%~60% of the cancer patients in the United States are currently being treated with IMRT. Improved local control and reduced complications as compared with 3D-CRT has been observed for all common anatomical sites with the largest number of applications for prostate and head and neck treatment.

However, the advancement in IMRT delivery doesn’t come without a risk. The clinical efficacy of IMRT relies on the ability of the planning system and the delivery system to accurately deliver planned dose to the target. The proximity of critical organ to tumor leads to high dose gradient which puts stringent requirements on modeling radiation beam penumbra as well as radiation dose outside beam opening. The complicated motion of MLC leaves to modulate beam segments makes leaf positioning accuracy more critical than 3D-CRT. Patient delivery quality assurance (QA) has become an integral part of IMRT treatment process. Increased effort has to be made to understand IMRT planning and delivery process and its
associated QA procedures compared with 3D-CRT. The AAPM guidance document on IMRT points out that IMRT QA consists of three tasks: commissioning and testing of the treatment planning and delivery systems, routine QA of the delivery system, and patient-specific validation of treatment plans. The first task concerns with the integrity of the planning system in modeling the delivery system. The second task deals with the mechanical and dosimetric accuracy of the delivery system (gantry and collimator rotation, MLC positioning, etc). The third task is to ensure safe and accurate treatment of a patient. In general, IMRT is a complex technique including patient simulation, treatment planning, leaf sequencing, plan transfer, patient positioning, online verification and treatment delivery. Patient specific IMRT QA, as a safe guard and total system check, plays an essential role in making sure IMRT delivery is carried out as prescribed and planned.

**Patient-specific IMRT QA**

The clinical efficacy of IMRT treatment depends on the ability of the delivery system to faithfully deliver the planned dose distribution to the desired location. IMRT process as described above has multiple steps with potential errors arising from each step. Patient-specific IMRT QA, as a total system check, provides a unique opportunity to identify these potential sources of errors and plays an essential role in ensuring the safe and accurate delivery of IMRT.

Typical patient-specific IMRT QA involves the measurement of a point dose or 2D dose distributions in a homogeneous phantom and compared with the treatment planning system (TPS) calculation. The whole delivery sequence of an IMRT treatment is transferred onto the CT image of the phantom and calculated with TPS. At the same time, the entire delivery sequence is delivered on the phantom and a planar dose at a certain depth is measured with some dosimeters. The measured and calculated planar dose distributions are compared and analyzed with dose difference, distance to agreement or gamma index, etc. The percentage of points, area or
volume passing a pre-selected criterion is used to indicate the quality of the whole planning and delivery procedure.

IMRT planning and delivery systems are changing rapidly and there have been no well established standards on measuring devices, criteria or acceptance levels for patient-specific IMRT QA. Different dosimeters have been employed or developed for IMRT QA measurement. Ion chamber combined with film is the early popular choice. An ion chamber could be placed in a high dose and low dose gradient region for absolute point dose measurement. Film can be irradiated to measure a relative dose distribution. An absolute planar dose map can be obtained by combining ion chamber and film measurement. Accuracy of the measurement depends on the selection of the measuring point. This is a time consuming process. Film is gradually replaced by online 2D detectors such as diode arrays and ion chamber arrays. Absolute planar dose distribution could be obtained during a single delivery which makes measurement more accurate and efficient. The downside of these online 2D detectors is their limited spatial resolution compared with film. Portal dosimetry based on electronic portal imaging device (EPID) is another alternative for IMRT QA measurement. It has excellent resolution with immediate readout. However, designed as an imaging device, EPID requires a lot of correction factors to achieve accurate dosimetry which may limit its wide application in IMRT QA. Just like there are a lot of different choices of dosimeters for QA measurement, there is no consensus on what criteria one should use in evaluating the agreement between measured and calculated dose distribution. A recent survey shows the majority responding clinical institutions use 3%/3mm criterion in their practice. At the same time, stricter criterion such as 2%/2mm and looser criterion such as 5%/5mm are both used at different institutions. With the same comparison criterion (for example, 3%/3mm), there have been very different achievable agreements (or
action levels) reported in literature. The survey conducted by Nelms et al shows majority of the community employing 3%/3mm criterion for IMRT QA analysis can achieve 90%-95% passing rates between measurement and calculation. Basran et al reported that greater than 95% agreement could be achieved for non-head and neck IMRT with same criterion, while the number for head and neck was only 88%. The AAPM summer school proceedings provide general guidance on IMRT QA, tolerance limits and action levels. However, it doesn’t seem practical to build standards across all institutions with all these diversities in the IMRT practice. It is recommended that each facility offering IMRT should develop its own guidelines and criteria for the acceptance and QA of IMRT planning and delivery. This will inevitably raise the question of how one should evaluate the performance of their own IMRT QA process in making sure the delivered dose distribution agrees with the prescribed and planned dose distribution.

In 2008, Radiological Physics Center (RPC) reported that 30% of 250 head and neck IMRT irradiations of an anthropomorphic phantom performed by academic institutions applying for credentialing failed to agree with their own treatment plan to within 7% difference and 4 mm DTA accuracy criteria. In other words, the irradiation of an anthropomorphic phantom demonstrated failings in their own QA procedures usually performed well on homogeneous virtual water phantom with a tighter criterion (such as 3%/3mm). These institutions were those who had confidence in their dosimetry accuracy and expected to pass the credentialing tests. The RPC credentialing test results clearly show the current IMRT QA process in these institutions fail to achieve their goal, i.e., identifying sources of errors in the system.

**Study Aims**

This dissertation consists of two parts. The first part evaluates the ability of current IMRT QA practice for fixed gantry IMRT in catching minor errors in the delivery system and proposes solutions to improve its sensitivity to errors and its efficiency. The second part focuses on
calibrating a novel four-dimensional dosimeter for dynamic gantry IMRT (VMAT) patient specific QA.

Specifically, the first part of this dissertation focuses on these aspects of patient specific QA for fixed gantry IMRT:

(1) Sensitivity to errors in delivery system. Patient specific IMRT QA is expected to catch both gross errors and minor errors in the planning and delivery system. Minor MLC positioning errors are of particular interest since MLC positioning reproducibility and accuracy are critical to the success delivery of IMRT. It is not uncommon to observe these errors and they typically lead to large dosimetric impact. Can the current IMRT QA process effectively catch these errors when they are present in the system? Our study shows that with either film or online 2D diode arrays the QA process fails to identify systematical MLC errors of 2 mm using popular 3%/3mm criterion.

(2) Accuracy. Our results show that patient-specific IMRT QA with a stricter criterion (such as 2%/2mm) exhibits stronger sensitivity to minor errors in delivery system. However, not all institutions can achieve acceptable passing rate with such strict criterion which leads to large number of false alarms. One of the main reasons causing large discrepancy between calculation and measurement is the inadequate beam modeling by using beam profiles measured with finite-sized ion chambers which suffer from volume averaging effect. In this dissertation, we develop algorithms to extract true photon beam profiles by removing volume averaging effect to improve treatment planning system (TPS) beam modeling. The improved TPS beam model significantly improves dose calculation accuracy. The improved agreement between calculated and measured dose distributions enables the use of stricter criterion in patient specific QA which enhances its sensitivity to minor errors in the delivery system.
(3) Efficiency. IMRT QA with offline detectors (film, gel dosimeter) is extremely time-consuming. With fast online electronic dosimeters (diode or IC arrays), it still takes 30 min on average for each patient QA. It has been pointed out that patient-specific verification could be reduced once confidence in dosimetric accuracy has been accumulated\textsuperscript{19}. In this dissertation, we develop a fast and accurate independent planar dose calculation algorithm to replace measurement based patient specific IMRT QA.\textsuperscript{20} The independent planar dose calculation is validated against diode array measurement. When performing patient-specific QA, TPS dose calculation is compared with the fast independent dose calculation instead of time-consuming delivery and measurement after clinic hours.

The second part of this dissertation deals with the calibration and testing of a novel device for patient-specific QA for the newly emerging VMAT. Patient-specific QA for VMAT differs significantly from IMRT due to its increased use of dynamic components (dynamic gantry and collimator, dynamic MLC and variable dose rate).\textsuperscript{21, 22} Ideally, patient-specific QA for VMAT should be performed in arc mode with an isotropic 3D dosimeter to check dosimetric impact of the interplay of these dynamic components. At present, there has been no such isotropic 3D dosimeter available in the market. Sun Nuclear Corp developed the first isotropic 3D dosimeter (ArcCHECK\textsuperscript{TM}) for patient-specific QA for VMAT. However, effective calibration procedure, dosimetric characterization and clinical application of this device are yet to be established. In this dissertation, we develop calibration procedures for this novel device and demonstrate its clinical application for both IMRT and VMAT patient-specific QA.

In summary, this dissertation is organized with five specific aims (Figure 1-2):

**Specific aim 1**: Investigate the sensitivity of current patient-specific IMRT QA to MLC leaf positioning errors (Chapter 2).
**Specific aim 2**: Develop deconvolution algorithms to recover true beam profiles to improve Pinnacle beam modeling and dose calculation accuracy. Tighter criteria could thus be employed in IMRT QA for enhanced sensitivity (Chapter 3).

**Specific aim 3**: Develop independent planar dose calculation algorithms to replace measurement based IMRT QA to improve its efficiency (Chapter 4).

**Specific aim 4**: Develop effective calibration procedure for ArcCHECK for its use in VMAT QA (Chapter 5).

**Specific aim 5**: Study ArcCHECK dosimetric characteristics and validate its clinical application for both IMRT and VMAT QA (Chapter 6).
Figure 1-1. (a) Multi-leaf Collimator designed by Varian Medical Systems. Typical MLC leaf has a width of 1 cm and each leaf can be moved independently to modulate the beam. (b) A patient undergoing IMRT. The LINAC can rotate around the patient to shoot the beam from the best angles.
Figure 1-2. Flow chart of the dissertation. The purpose of this work is to propose solutions for existing issues with patient specific IMRT QA. Major work towards improving QA for fixed gantry IMRT includes the evaluation of its sensitivity (SA 1), development of deconvolution algorithm to extract true photon beam profiles for TPS commissioning (SA 2) and an independent planar dose calculation (SA 3) aiming to replace measurement based IMRT QA. For VMAT, a novel 4D diode array is characterized and calibrated (SA 4). Algorithms for its clinical application in both IMRT and VMAT QA are developed (SA 5).
CHAPTER 2
ON THE SENSITIVITY OF PATIENT-SPECIFIC IMRT QA TO MLC POSITIONING ERRORS

Introduction

Intensity-modulated radiation therapy (IMRT) has become the treatment technique of choice for many types of cancers receiving radiation therapy. The clinical efficacy of IMRT relies on dose escalation to the tumor while avoiding toxicity to the surrounding critical structures. Accurate multi-leaf collimator (MLC) leaf positioning plays a crucial role in the effective implementation of MLC-based IMRT\(^\text{13}\). Tolerance limits for leaf position accuracy and reproducibility have been suggested for IMRT which are more stringent than for conventional radiation therapy\(^\text{23}\).

Several authors have studied the dosimetric effect of leaf positioning errors\(^\text{16, 24-27}\). Luo et al studied the correlation between leaf position errors and dosimetric impact in prostate cancer treatment\(^\text{24}\). They found a linear correlation between the target dose error and the average MLC position error, with 1% target dose change arising from 0.2 mm systematic leaf position errors. LoSasso et al also reported that a 0.2 mm gap variation leads to 1% dose variation with an average gap width of 2 cm with dynamic beam delivery\(^\text{28}\). Mu et al studied the dosimetric effect of leaf position errors on head and neck patients by deliberately introducing random (uniformly sampled from 0 mm, ±1 mm and ±2 mm) and systematic (±0.5 mm or ±1 mm) leaf positioning errors into the plan\(^\text{26}\). They found no significant dosimetric effect (<2% dose change to both target and critical organs) introduced by random leaf position errors up to 2 mm, while significant effects (8% change in \(D_{95}\%\) and ~12% in \(D_{0.1\text{cc}}\) to critical organs) were observed by 1 mm systematic leaf position errors in complex IMRT plans. Zygmanski et al studied the dosimetric effect of truncated Gaussian (with 0.1 cm standard deviation) shaped random leaf position errors\(^\text{27}\). They found that although the average composite dose to the target of a nine
field IMRT plan was changed only by 3%, fluence change resulted from each single field was commonly > 10%. Woo et al found that leaf position uncertainty could lead to dose variations of up to 13% when positioning the ion chamber on the field edge. All these studies emphasized the importance of the MLC positioning accuracy and reproducibility.

Several authors have reported excellent accuracy of MLC leaf position by analyzing MLC log files for both dynamic MLC and static MLC. For dynamic MLC, Zygmanski et al reported <0.05 cm leaf position error, while LoSasso et al found that the average leaf gap error was much smaller than 0.02 cm. For static MLC, Luo et al reported average leaf position errors of ~0.05 cm based on the analysis of MLC log files. On the other hand, by using a fast video-based electronic portal imaging device, Zeidan et al observed a maximum unplanned leaf movement of 3 mm during static MLC delivery. The same group’s study, based on MLC log file analysis, reported that in approximately 80% of the total segment deliveries, at least one collimator leaf had unplanned movement of at least 1 mm (projected at isocenter) during segment delivery. Significant dosimetric impact could arise from these MLC position errors which suggests that periodic MLC quality assurance (QA) should be performed to ensure the accuracy and reproducibility of MLC leaf positions.

It is both challenging and time-consuming to check the position accuracy of every single MLC leaf pair at all possible off axis positions. In practice, dedicated MLC QA is conducted bi-weekly, or even less frequently, whereas patient-specific IMRT QA is usually done for every new patient before the start of IMRT treatment. The aim of this work is to assess the sensitivity of patient-specific IMRT QA to leaf position errors. A common method of patient-specific QA is to re-calculate the treatment plan in a QA phantom with all beams at 0° gantry angle (IEC convention) and normal to the phantom surface. The planar dose distribution is measured under
the same geometry using film or two-dimensional (2D) detector arrays. The agreement between the calculated and measured planar dose distributions is then quantified using parameters like the Gamma index\textsuperscript{10}, percent dose difference (%Diff) or distance-to-agreement (DTA), etc. Patient-specific IMRT QA is required for every new IMRT patient to ensure the accuracy of the treatment plan. Childress \textit{et al} have investigated the feasibility of using a 2D dosimetric system to detect gross errors such as beam energy change, wrong patient’s beam data, one beam collimator angle change of 90°, gantry angle change of 10° and omitting the delivery of one beam\textsuperscript{34}. However, the sensitivity of commonly used patient-specific IMRT QA systems to subtle MLC position errors has not been fully examined. Sastre-Padro \textit{et al} studied the consequences of subtle leaf positioning errors on IMRT delivery but whether the errors could be detected by patient-specific IMRT QA systems was not answered\textsuperscript{35}. In this work, random MLC position errors up to 2 mm as well as systematic errors (±1 mm, ±2 mm) were introduced into treatment plans for 8 H&N IMRT patients (totaling 53 fields). Planar dose distributions calculated before and after introducing errors were compared to dose distributions measured with both film and 2D diode arrays using gamma index as well as DTA criterion on a field-by-field basis. The change in the passing rate distribution was used as an indicator to study the sensitivity of the IMRT QA procedure to MLC position errors.

\textbf{Materials and Methods}

\textbf{Patient Plan Selection}

Eight head and neck IMRT plans (total of 53 fields) previously used for patient treatment with 6 MV photon beams of a linear accelerator (Trilogy, Varian Medical Systems, Palo Alto, CA) with 120-leaf Millennium MLC were randomly selected for this study. The step-and-shoot treatment plans were generated with a commercial treatment planning system (TPS, Pinnacle\textsuperscript{3}, version 8.0d, Philips Medical Systems, Madison, WI) with direct machine parameter
optimization (DMPO) option which directly optimizes the shape and weight of each MLC segment. Minimum segment area and minimum segment MU were set to 4 cm² and 3 MU, respectively. The adaptive convolution dose calculation algorithm with inhomogeneity correction was used for all the plans.

**Leaf Positioning Errors Simulation**

Random and systematic leaf positioning errors were simulated in the same manner as in Mu et al. The original treatment plans were exported from Pinnacle using the Pinnacle scripting language. Leaf position errors were introduced into the plans by directly modifying the leaf positions in the exported treatment plans. Random errors were uniformly sampled from [0 mm, ±1 mm, ±2 mm] for each leaf from both leaf banks. Negative and positive errors make the leaf end-to-end distance smaller and larger, respectively. When leaf collisions happen, the leaves were assume closed and they were assigned the same position. When introducing systematic errors, the position of each open leaf was changed by the same amount (±1 mm and ±2 mm). A total of 6 plans were generated for each patient: the original plan without leaf positioning error, the plan with random errors and four plans with different amount of systematic errors. The modified treatment plans were imported back into Pinnacle for planar dose calculation. Planar dose distribution was calculated for each field at 10 cm depth and 90 cm source-to-surface distance (SSD) in a rectangular water phantom with 0° gantry angle (IEC convention) and normal to the phantom surface. All the calculation was done using a 1.0 x 1.0 mm² dose grid.

**Dose Distribution Measurement**

In standard practice for patient-specific IMRT QA, both film and 2D detector arrays are widely used. In this work, both radiochromic films and a 2D diode array were employed to measure the planar dose distribution under the same geometry as in calculation. Radiochromic films (Gafchromic EBT film, ISP Corp., Wayne, NJ) were sandwiched between solid water
pieces at 10 cm depth with a source-to-film distance of 100 cm. Small marks were placed on the edges of the films for geometric registration. After irradiation, the films were left for 24 hours before analysis as recommended by Niroomand-Rad et al\textsuperscript{36} to minimize post irradiation coloration effects. A commercial flatbed scanner (EPSON 1680) was used to digitize the film at a resolution of 100 μm/pixel and was sampled at a spatial resolution of 1 mm at both directions using commercial data analysis software (Matlab v7.0, Mathworks, Inc.). An adaptive 2D wiener filter of 4x4 pixel region was used to reduce the inherent noise. Film images were acquired from the red CCD channel only. Sensitometric curves were obtained by delivering varying amounts of radiation which covered the range of the doses of interest. When irradiating the radiochromic films, all doses were rescaled such that the maximum dose was around 100 cGy to avoid the uncertainty at low doses\textsuperscript{37}.

Convenient 2D detector arrays are replacing film for patient-specific IMRT QA. A 2D diode matrix (MapCHECK 1175, Sun Nuclear Corp., Melbourne, FL) was also used to measure the planar dose distribution under the same geometry as the film measurement. The MapCHECK device has been demonstrated to be an excellent tool for the verification of IMRT fields with little energy or dose rate dependence\textsuperscript{38,39}. An intercomparison between film and MapCHECK showed that MapCHECK can effectively replace film dosimetry in routine IMRT QA, even though it has a limited spatial resolution\textsuperscript{40}. The MapCHECK device was calibrated for absolute dose measurement.

The original clinical treatment plans without MLC positioning errors were delivered for measurement. Each of the 53 IMRT fields was measured with the diode array and 10 of the fields were measured with the radiochromic films.
Planar Dose Comparison

Planar dose distributions without and with the leaf position errors were calculated for each of the 53 fields. The calculated planar dose distributions were imported into the MapCHECK analysis software and compared with either MapCHECK measurement or film measurement. When calculation was compared with film measurement, film measurement was loaded into MapCHECK software and used as reference. Each of the MapCHECK (or film) measured points above 10% of the maximum dose level was compared with calculation using absolute distance-to-agreement (DTA) comparison as well as $\gamma$ index in the absolute dose comparison mode. Two sets of criteria, 2%/2mm and 3%/3mm, were employed, and the percentage of the points passing the acceptance criteria was evaluated.

Criteria for Identifying Errors

The decision regarding whether the procedure could identify the MLC errors was made based on the comparison between passing rate distributions before and after the errors were introduced. In this work, the change in average passing rates and a significance test were used to make the decision. A sudden drop of the average passing rate from the baseline value serves as a first indication of errors in the system. In this work, the threshold of average passing rate decrease was arbitrarily set at 5%. A non-parametric significance test, Wilcoxon rank-sum test, was employed to compare the distribution of the passing rates before and after the introduction of the MLC position errors. The Wilcoxon rank-sum test was chosen over the popular student t-test because the passing rates didn’t follow the normal distribution and the standard deviation varied significantly. The p-value of the test gives the probability the two compared variables are from the same distribution with equal median. The significance level of the test was set to 0.01. In other words, the difference was significant only when the p-value of the test was less than 0.01. In summary, the introduced MLC position errors would be considered “identified”
when (1) the drop in average passing rates was larger than 5% and (2) the p-value from the Wilcoxon rank-sum test was smaller than 0.01.

Results

The average passing rates and the standard deviations using both films and the diode array are displayed in Figure 2-1 to 2-2 and Figure 2-3 to 2-4, respectively, for different MLC positioning errors, and the drop of average passing rate caused by MLC positioning errors is shown in Figure 2-5 to 2-6. The p-value from the Wilcoxon rank-sum test is summarized in Table 2-1. For film measurements, the average passing rate with “DTA, 3%/3mm” criterion was 93±5% (1 standard deviation) without MLC errors. With random leaf position errors of up to 2 mm and systematic errors on the order of 1 mm, the average passing rate dropped by less than 5% and was still around 90%. For 2 mm systematic errors, an asymmetric behavior was observed with 5% drop in average passing rate for positive errors but ~15% drop for negative errors. The p-values from the rank-sum test indicated significant difference only for negative 2 mm systematic error, with a p-value of 1.10E-03. In this case, we conclude that only the -2 mm systematic MLC positioning error could be identified by patient-specific IMRT QA procedure when using EBT film and “DTA, 3%/3mm”.

When using “DTA, 2%/2mm” criterion in film result analysis, both the -1 mm and ±2 mm systematic errors would cause the average passing rates to drop by more than 5%. But rank-sum test showed that only -2 mm error caused significant difference in passing rate distribution with a p-value of 6.22E-04. For the other cases, there is significant overlap between the distributions. Based on our criteria for identifying errors, only the -2 mm errors could be identified. The same can be concluded when employing γ test for film result analysis.
Figure 2-3 to 2-4 show the results when using MapCHECK as the detector. The drop of average passing rates is displayed in Figure 6 and the p-values from Wilcoxon rank-sum test are summarized in Table 2-1. The p-values indicated that both ±1 mm errors and ±2 mm errors caused significant difference in the passing rate distributions for both 2%/2mm and 3%/3mm criterion, using either DTA comparison or γ analysis. However, Figure 2-6 shows that larger than 5% drop in average passing rate was observed only for ±2 mm error. Thus we conclude that with MapCHECK, the IMRT QA procedure could identify systematic errors on the order of 2 mm.

**Discussion**

Due to the complexity of planning and delivery of IMRT, patient-specific QA is recommended for every new patient before the start of IMRT delivery. This could potentially detect any gross errors such as the wrong beam energy, wrong patient plan, or any data transfer errors from the TPS to the delivery system. However, the ability of patient-specific QA in detecting subtle errors, such as TPS calculation error or MLC positioning error, is more difficult to quantify. In practice, after commissioning of the IMRT planning and delivery system, baseline values for patient-specific QA acceptance criteria, which might be site-dependent, could be established over a period of time. A sudden drop in passing rates would signify deviations in one or more of the IMRT planning and delivery components and would warrant further investigation. Therefore, whether the commonly-used QA method of field-by-field planar dose comparison is sensitive to MLC positioning errors on the order that would significantly affect the treatment is of practical interest.

Both radiochromic film and MapCHECK device were used in this study. When no MLC positioning errors were introduced, the passing rate distribution from MapCHECK device had a less variation than radiochromic film: 1% vs. 5% with the 3%/3mm criterion and 3% vs. 9% with
the 2%/2mm criterion. With larger MLC positioning errors, the variation associated with MapCHECK passing rate distribution gradually increased to around 6% with 3%/3mm and 9% with 2%/2mm, while it stayed constant for radiochromic film. If the average passing rates didn’t decrease significantly when the MLC errors were introduced, the passing rate distribution would overlap the one without errors. Figure 2-1 to 2-4 showed that the overlapping was larger for film than for MapCHECK device due to the larger variation associated with film. This fact should explain the larger p-value from rank-sum test with radiochromic film than with MapCHECK. From this perspective, MapCHECK device showed larger sensitivity than radiochromic film to MLC positioning errors. The difference in variation associated with passing rate distribution was mainly due to different detector response. The variation of film sensitivity is on the order of 1.5%~4% depending on dose level\textsuperscript{37}, while MapCHECK has variation of only 0.15%\textsuperscript{38,39}.

Larger variation in the passing rate distribution was found when using 2%/2mm than 3%/3mm. At the same time, a larger drop in the average passing rate was noticed with 2%/2mm than with 3%/3mm as shown in Figure 2-5 to 2-6. Table 2-1 showed smaller p-values with 2%/2mm than with 3%/3mm in general, which meant a higher probability of distinguishing the passing rate distributions with 2%/2mm. Therefore, a tighter criterion shows more sensitivity to MLC position errors. With 2%/2mm, the average passing rate was ~93% and ~80% with MapCHECK and radiochromic film, respectively. It was relatively difficult to achieve acceptable passing rates using film with tighter criterion\textsuperscript{37}.

Both the DTA analysis and $\gamma$ index were studied in this work. DTA analysis is more stringent than $\gamma$ index and on average gives 2~3% lower passing rates. However, no significant difference was found in the decrease of average passing rate and the p-values from rank-sum test
were in general on the same order, which suggested that the choice of using DTA analysis or $\gamma$ index didn’t affect the sensitivity of the IMRT QA to MLC position errors.

Asymmetric response to +2 mm and -2 mm systematic MLC positioning errors was noticed with both MapCHECK and film. The decrease in average passing rate was ~3% larger with -2 mm systematic errors than with +2 mm errors when using MapCHECK as the detector. When using film, a 10% larger decrease in average passing rate was noticed with -2 mm errors than with +2 mm errors (25% vs. 10% when using 2%/2mm criterion for both DTA analysis and $\gamma$ index). To investigate the asymmetric behavior with respect to systematic MLC positioning errors, the following simulation was conducted to reveal the cause of such behavior. The planar dose distribution calculated from plan without MLC errors was re-sampled and only those points corresponding to a detector position on the MapCHECK device were kept. The re-sampled dose distribution was compared to other four dose distributions from plans with systematic MLC errors (±1mm and ±2mm) using 2%/2mm DTA criterion. The average passing rates of 15 IMRT plans were shown in Figure 2-7. An obvious asymmetric behavior was noticed. Compared to an error free plan, plans with -1 mm symmetric MLC errors had ~3% lower average passing rates than plans with +1 mm symmetric MLC errors (93% vs. 96%). While for 2 mm systematic MLC errors, 17% lower average passing rates were observed for plans with negative errors than plans with positive errors (66% vs. 83%). A similar trend was observed when comparing error-free plans to plans with errors in full resolution (i.e., no re-sampling). This simulation exercise showed that the asymmetric behavior is not caused by MLC calibration error or TPS modeling errors, but rather indicated that the dose distribution is more sensitive to negative systematic MLC errors (which makes the MLC segment opening smaller) than positive errors. The
magnitudes of the passing rate differences between the negative and positive MLC errors are very similar to those observed in Figure 2-1 to 2-4.

Childress et al investigated the feasibility of automatically detecting both gross delivery errors and small delivery errors (5 mm lateral shift of isocenter, etc) with patient-specific IMRT QA procedures. They found that most scalar dose comparison metrics, including $\gamma$ index, failed in identifying isocenter shift error of 5 mm. Our results indicated that the studied IMRT QA procedures could detect systematic MLC positioning errors of 2 mm. A major difference in their work was that they compared composite dose distribution while our analysis was done on a field-by-field basis. One reason for poor detection ability when using composite dose distribution in comparison could be attributed to the fact that when composite dose distribution was analyzed, errors from individual fields could compensate each other.

As TPS beam modeling, dose calculation and dose measurement become more accurate, tighter criterion could be employed in the analysis and still achieve good agreement between calculated and measured dose distribution, which will lead to better and earlier IMRT delivery error detection, especially with MapCHECK which provides enhanced sensitivity compared with EBT film. However, considering the dosimetric impact of such errors, patient-specific IMRT QA should be combined with a periodic MLC positioning check program in order to guarantee the accuracy of IMRT delivery.

Conclusions

In this work, we investigated the sensitivity of patient-specific IMRT QA procedure to MLC positioning errors. Random errors of up to 2 mm and systematic errors of $\pm 1$ mm and $\pm 2$ mm were simulated in the treatment plans. Planar dose distributions calculated with plans containing errors were compared to dose measurement with both Gafchromic EBT films and
MapCHECK device, acquired while irradiating the original error-free plan. DTA analysis and $\gamma$ index with 2%/2mm or 3%/3mm criteria were employed to evaluate the agreement between calculation and measurement. The change of passing rates distribution was used as an indicator of the sensitivity to simulated MLC positioning errors. It was found that the studied patient-specific IMRT QA procedure with both radiochromic film and MapCHECK device was just able to detect systematic errors on the order of 2 mm. Significantly larger sensitivity to MLC position errors was found with the 2%/2mm criterion than with the 3%/3mm criterion. Since clinical acceptable agreement between calculation and measurement using film dosimetry with 2%/2mm is relatively difficult to achieve, MapCHECK is more appropriate to be employed with a tight criterion. As the accuracy of dose calculation and measurement is further improved, the sensitivity of IMRT QA procedure to MLC positioning errors will be greatly enhanced. However, considering the dosimetric impact of such errors, patient-specific IMRT QA should be combined with a periodic MLC QA program in order to guarantee the accuracy of IMRT delivery.
Table 2-1. P-values from two-tailed Wilcoxon rank-sum test. The test compared the distribution of passing rates before and after the introduction of MLC position errors.

<table>
<thead>
<tr>
<th></th>
<th>p-Value</th>
<th>Random</th>
<th>1 mm</th>
<th>-1 mm</th>
<th>2 mm</th>
<th>-2 mm</th>
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<tr>
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<tr>
<td>DTA</td>
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<td>1.05E-01</td>
<td>2.81E-02</td>
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<tr>
<td></td>
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<td>1.61E-01</td>
<td>7.74E-02</td>
<td>1.10E-03</td>
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<tr>
<td>Gamma</td>
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<td>9.81E-01</td>
<td>1.30E-01</td>
<td>2.81E-02</td>
<td>6.22E-04</td>
</tr>
<tr>
<td></td>
<td>3%/3mm</td>
<td>6.29E-01</td>
<td>1.00E+00</td>
<td>3.14E-01</td>
<td>9.95E-02</td>
<td>1.90E-03</td>
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<tr>
<td>DTA</td>
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<td>2.35E-11</td>
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<td>3.85E-06</td>
<td>3.57E-04</td>
<td>4.33E-12</td>
<td>9.20E-10</td>
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</tbody>
</table>

Figure 2-1. Average passing rates when comparing Pinnacle calculated planar dose distribution and Gafchromic EBT film measurement using 2%/2mm and 3%/3mm criterion with DTA analysis for a total of 53 IMRT fields. The x-axis, from left to right, stands for no MLC positioning error, random errors and systematic errors of ±1 mm and ±2 mm, respectively. The error bar indicates one standard deviation.
Figure 2-2. Average passing rates when comparing Pinnacle calculated planar dose distribution and Gafchometric EBT film measurement using 2%/2mm and 3%/3mm criterion with Gamma index for a total of 53 IMRT fields. The x-axis, from left to right, stands for no MLC positioning error, random errors and systematic errors of ±1 mm and ±2 mm, respectively. The error bar indicates one standard deviation.

Figure 2-3. Average passing rates when comparing Pinnacle calculated planar dose distribution and MapCHECK measurement using 2%/2mm and 3%/3mm criterion with DTA analysis for a total of 53 IMRT fields. The x-axis, from left to right, stands for no MLC positioning error, random errors and systematic errors of ±1 mm and ±2 mm, respectively. The error bar indicates one standard deviation.
Figure 2-4. Average passing rates when comparing Pinnacle calculated planar dose distribution and MapCHECK measurement using 2%/2mm and 3%/3mm criterion with Gamma index analysis for a total of 53 IMRT fields. The x-axis, from left to right, stands for no MLC positioning error, random errors and systematic errors of ±1 mm and ±2 mm, respectively. The error bar indicates one standard deviation.

Figure 2-5. Drop of average passing rates with respect to the clinical plan (no MLC positioning errors) with the introduction of random and systematic errors. The comparison was between Pinnacle calculated planar dose distribution and film measurement using DTA 2%/2mm and 3%/3mm as well as $\gamma$ 2%/2mm and 3%/3mm criterion. The x-axis, from left to right, stands for no MLC positioning error, random errors and systematic errors of ±1 mm and ±2 mm, respectively.
Figure 2-6. Drop of average passing rates with respect to the clinical plan (no MLC positioning errors) with the introduction of random and systematic errors. The comparison was between Pinnacle calculated planar dose distribution and MapCHECK measurement using DTA 2%/2mm and 3%/3mm as well as Gamma 2%/2mm and 3%/3mm criterion. The x-axis, from left to right, stands for no MLC positioning error, random errors and systematic errors of ±1 mm and ±2 mm, respectively.

Figure 2-7. Average passing rates when comparing error free plans (re-sampled according to MapCHECK detector positions) to plans with +/- 1 mm and +/- 2 mm systematic MLC errors using DTA 2%/2mm criterion. The error bar stands for one standard deviation.
CHAPTER 3
THE EXTRACTION OF TRUE PHOTON BEAM PROFILE FOR TPS COMMISSIONING

Introduction

The aim of intensity modulated radiation therapy (IMRT) is to deliver highly conformal radiation dose to the planning target volume while avoiding critical structures as much as possible. Complicated IMRT fields are characterized by many small segments and steep dose gradients which require treatment verification. The main purpose of patient-specific quality assurance (QA) is to detect potential errors which could arise in the chain of IMRT process: treatment planning system (TPS), treatment planning, dose calculation, plan parameter transmission and treatment delivery.\(^2\)

A common method of patient specific QA is to re-calculate the treatment plan in a QA phantom with all beams at 0° gantry angle and normal to the phantom surface. The planar dose distribution is measured under same geometry using film or two-dimensional (2D) detector arrays. The agreement between the calculated and measured planar dose distribution is then quantified using parameters like the Gamma index,\(^{10,11,42,43}\) percent dose difference (%Diff), or distance-to-agreement (DTA), etc. The existence of various sources of errors in the chain of the IMRT process makes it difficult to analyze unsatisfactory IMRT QA results. Large clinical effort has been put in identifying the sources of discrepancy. If necessary, the plan is changed to achieve acceptable IMRT QA analysis statistics\(^{11}\). Studies\(^{44,45}\) have shown that the main source of discrepancy is detector size effect in TPS commissioning data, which makes it even more difficult to detect errors due to mechanical inaccuracy or TPS dose calculation error.

An essential issue in the beam modeling of TPS is to ensure the calculated dose matches the dose delivered to the patient. A key component in successful TPS commissioning is the accurate measurement of beam profiles\(^1\), especially the penumbra\(^{44}\). Measurement error in the
beam profile will propagate to TPS dose calculation directly. A well-known source of error during cross beam profile measurement is the volume averaging effect caused by the finite size of the measuring detectors, which causes the artificial broadening of the profile penumbra. Depending on the specific detector and radiation field, the commonly used standard ionization chambers with volumes on the order of 0.1~0.2 cm$^3$ can cause penumbra increase by 0.2~0.3 cm\textsuperscript{,44,46,47} which is obviously large enough to influence the IMRT dose calculation accuracy. In the effort to rectify the measurement, two approaches have been attempted to recover the “true” profiles in the literatures\textsuperscript{46}: (1) perform the measurement with variable size detectors and extrapolate to zero size detector\textsuperscript{46-50} (2) deconvolve the detector response kernel from the measured profiles.\textsuperscript{46,51-56} The first approach utilizes detectors of variable sizes to measure the same beam profiles. The relationship between penumbra width and detector sizes can be established and the “true” penumbra can be recovered by extrapolating down to zero size detector. The disadvantage of this method includes: (1) several different sized detectors are required; (2) it is time-consuming. The second approach either calculates or measures the detector response kernel and its effect on penumbra broadening can then be eliminated by numerical or analytical deconvolution. Numerical deconvolution suffers from measurement noise and becomes very unstable.\textsuperscript{51,52,54,55,57} Analytical deconvolution produces stable outcome but it depends on the choice of analytical functions which would fit both the profiles and the kernel and the deconvolution can be solved analytically.\textsuperscript{51,57}

Although the detector size effect on the profile measurement has been well understood, its clinical impact on IMRT QA has not been quantified. Laub and Wong pointed out the importance of the detector size effect in small field dosimetry in IMRT but the effect on patient specific IMRT QA was not studied.\textsuperscript{48} Arnfield \textit{et al} used film dosimetry to improve beam profile
penumbra measurement and observed improved agreement between calculation and measurement. Esch et al studied patient specific IMRT QA results among five European departments and concluded that the detector size effect was one of the main sources for discrepancy. But the degree of influence was not determined. A recent survey showed that the majority of the community which uses planar dose verification method employs 3%/3mm criteria for IMRT QA analysis with the number of points passing the criteria (passing rates) of 90%-95% as acceptable. Yang et al proposed an acceptable criterion of ≥85% passing rates with ±3% and ±3 mm DTA, which are considered achievable even with the most complicated cases. Given that the leaf positioning accuracy for most MLC systems can achieve better than 1 mm accuracy, the use of 3 mm as distance to agreement does not seem justified. Is detector size effect the limiting factor? Could more stringent criteria be applied if detector size effect is removed? The impact of the detector size effect on the results of IMRT planar dose verification is the focus of this work.

The aim of this work is two fold: (1) to develop a method of extracting the “true” cross beam profiles, free of volume averaging effect, using analytic fitting/deconvolution; and (2) to investigate the clinical impact of detector size effect on patient specific IMRT QA. Photon beam profiles were obtained with a 6 mm and a 4 mm diameter ion chambers, respectively. An analytic function was used to fit only the high dose gradient regions of the profiles (penumbra regions) and the “true” profile set was recovered via an analytic deconvolution approach. Each of these beam profile sets was used to commission a photon beam model. Planar dose distributions were calculated for 53 IMRT fields using each of the three beam models. The agreement between calculated and measured planar dose distribution was evaluated using %Diff and DTA.
Significantly better agreement between calculated and measured planar dose distributions was achieved with the beam model commissioned using the deconvolved profiles.

**Methods and Materials**

**Beam Data Measurement**

A commercial treatment planning system (TPS, Pinnacle3, version 8.0d, Philips Medical Systems, Madison, WI) which employs a convolution-superposition based dose calculation algorithm was commissioned for the 6 MV photon beam of a linear accelerator (Trilogy, Varian Medical Systems, Palo Alto, CA) with 120-leaf Millennium MLC. Data requirement for the TPS commissioning includes depth dose curves at different field sizes, cross beam dose profiles for collimator- and MLC-defined fields at different field sizes and depths, calibration and relative output factors, block and tray transmission factors. Depth dose and cross beam profiles were acquired with two scanning ionization chambers: one with 6 mm diameter, 5.8 mm length and 0.13 cm³ volume (CC13, Scanditronix Wellhofer, Bartlett, TN), and the other with 4 mm diameter, 3.6 mm length and 0.04 cm³ volume (CC04, Scanditronix Wellhofer) in a 48x48x48 cm³ scanning water tank (Wellhofer Dosimetrie, Shwarzenbruck, Germany). All beam profiles were collected at a source-to-surface distance (SSD) of 90 cm.

**Beam Profile Deconvolution**

The effect of the finite size of the detector on beam profile measurement is described by a convolution of the detector response function $K(x)$ to the true profile $P_t(x)$\(^{51, 52, 55}\)

$$P_m(x) = \int P_t(u) \cdot K(x-u) \cdot du$$

(3-1)

where $P_m(x)$ is the measured profile. The true profile can be derived by using the convolution theorem,

$$F[P_m(x)] = F[P_t(x)] \cdot F[K(x)]$$

(3-2)
and inverse Fourier transform,

\[ P_f(x) = F^{-1} \left[ \frac{F[P_m(x)]}{F[K(x)]} \right] \]  

(3-3)

Where \( F[ \ ] \) and \( F^{-1}[ \ ] \) denote Fourier transform and inverse Fourier transform, respectively. So there are two steps to recover the true profile.

1). Determine the detector response function \( K(x) \).

2). Perform de-convolution on measured profile.

As mentioned above, discrete deconvolution suffers from measurement noise and numerical errors. In practice, it is very complex and the results depend on the noise level, the measurement uncertainty and the cut-off frequency chosen to filter the inherent noise.\(^{51, 57}\) This approach becomes impractical in TPS beam modeling where a large number of profiles are needed.\(^{59-61}\) Garcia-Vicente \( et \ al \)\(^{51}\) provided an exact analytic solution to the deconvolution problem. The solution was limited to small field size profiles. In this work, we extended and applied their method to profile of arbitrary field size and depth for the purpose of TPS beam modeling.

**Detector response function**

The detector response function for arbitrary detector could be derived from Equation 3-1 and Equation 3-2 if the “true” profile is known. A common approach to derive the “true” profile is to measure the same profile using variable size detectors and extrapolate down to zero size detector.\(^{46-50}\) The downside of this approach is that measurements with a large number of variable size detectors are required in order to get good confidence in the extrapolation. Pappas \( et \ al \)\(^{57}\) studied the relationship between detector size and penumbra width using gel dosimeter and concluded that detector sizes of 0.5 mm to 0.7 mm were adequate for relative dose
measurements in high gradient regions of a 5 mm diameter photon beam. In this study, we used a small diode detector (Edge Detector, Sun Nuclear Corp., Melbourne, FL) with 0.8x0.8 mm² effective detection area to collect the “true” profile, \( P_t(x) \), for small field sizes. Comparison between the profiles measured using the diode and radiochromic films showed excellent agreement for small field sizes\(^{62}\). The same profiles measured with CC13 were denoted as \( P_m(x) \).

The detector response function is assumed to have a Gaussian shape,\(^{51}\)

\[
K(x) \approx e^{-\pi \cdot x^2 / \sigma_k^2}
\]

(3-4)

where \( \sigma_k \) is the shape parameter. \( \sigma_k \) is related to the sigma of a standard Gaussian function form \( \sigma \) by multiplying \( \sigma_k \) by a factor of 0.4. The optimal \( \sigma_k \) was obtained by minimizing the difference between \( P_m(x) \) and the convolution of \( P_t(x) \) and \( K(x) \) in a least square sense. The modified Levenberg-Marquardt algorithm was used to solve this nonlinear problem.\(^{63}\)

Cross beam profiles of collimator defined field sizes of 4x4 cm² and 10x10 cm² were measured with the Edge Detector at SSD of 90 cm and at depths of 1.5 cm, 10 cm, and 25 cm to derive the detector response function parameters. This procedure was restricted to small (\( \leq 10 \times 10 \) cm²) field sizes since diode measured profiles are known to deviate from chamber measured profiles at large field size due to the energy dependence of its response.\(^{48,64-66}\) The dependence of the detector response function on field size and depth was also investigated.

**Profile deconvolution**

The essential part of analytical deconvolution is to find an analytic function that can fit the measured beam profiles well and Equation 3-3 can be solved in a closed form. Garcia-Vicente et al\(^{51}\) used the difference of error functions, originally proposed by Cho et al,\(^{67}\) to fit small photon fields,

\[
F_m(x) = 1/2\{\text{erf}[(\pi)^{1/2}(L-2x)/(2\sigma_m)] + \text{erf}[(\pi)^{1/2}(L+2x)/(2\sigma_m)]\} + c
\]

(3-5)
where $L$ is the field width, $\sigma_m$ the shape parameter and $\text{erf}(x)$ the error function,

$$\text{erf}(x) = \frac{2}{\sqrt{\pi}} \int_0^x e^{-t^2} dt$$  \hspace{1cm} (3-6)

They worked out an analytic solution to Equation 3-3 which has the same functional shape as Equation 3-5 but with a new shape parameter $\sigma_t$,

$$\sigma_t^2 = \sigma_m^2 - \sigma_k^2$$ \hspace{1cm} (3-7)

Thus, if the measured beam profile could be fit reasonably well using Equation 3-5, the true profile could be recovered once the shape parameter $\sigma_k$ for the detector response function was determined and deconvolution becomes a simple replacement of the shape parameter with the new one. It was demonstrated that good fits could be achieved for small photon beams using Equation 3-5.\textsuperscript{51, 67} However, large field photon beams are characterized by the well known “horn” effect in the high dose region due to the flattening filter. The difference of error functions will fail in reproducing this feature which is present in all large field photon beams at various depths.

Computer simulation\textsuperscript{68} showed that the effect of volume averaging on low dose gradient regions (including the tail region and “horn”-shaped high dose region) was negligible; only the high dose gradient region was affected significantly. This motivated us to exclude the low dose gradient region and only keep the high dose gradient region in consideration when rectifying measured profiles. Meanwhile, the function used to fit the measured profile could be extended as following,

$$\overline{F_m}(x) = \sum_{n=1}^{N} F_n(x)$$  \hspace{1cm} (3-8)

where each $F_n(x)$ is the difference of error functions as expressed in Equation 3-5. We found that the use of $N = 5$ gave the best fit for all the profiles with no improvement when more
terms were used. The selection of high dose gradient region could be either gradient based or distance based. In gradient based selection, only the region where local gradient is above 10% of the gradient at the inflection point (the point where the derivative of the gradient changes signs) is included. In distance based approach, the region falling in between 80% and 120% of field width is included. Both methods were implemented and results showed little difference. In this study, the gradient based selection was used. Once the measured profile is fit using Equation 3-8, the deconvolution is simply to adjust the shape parameter of each $F_n(x)$ according to Equation 3-7. In-plane and cross-plane beam profiles for symmetric fields with field sizes ranging from 4x4 cm$^2$ to 40x40 cm$^2$ at depths of 1.5 cm, 10 cm, and 25 cm were deconvolved using this approach. Asymmetric profiles were divided into two half profiles at the field center and each half was deconvolved separately and then combined together. The deconvolved profiles for the two smaller field sizes of 4x4 cm$^2$ and 10x10 cm$^2$ were validated using the Edge Detector measured profiles.

**TPS Commissioning**

External photon beam commissioning for Pinnacle$^3$ has been well documented in the literature.$^{59-61}$ The convolution-superposition algorithm employed in the TPS computes dose distributions from first principles. Photon fluence exiting the head of the linac was characterized by a set of photon beam model parameters. Different regions of the depth dose curves and dose profiles were used to adjust the parameters which characterize the beam.$^{59}$ The rounded leaf end effect of the MLC was characterized by a set of parameters which modeled the penumbral difference between the jaw-defined and MLC-defined profiles. These parameters were iteratively adjusted during the modeling process such that the calculated profiles match the measured ones as much as possible.
Three different beam data sets (CC13-measured, CC04-measured and deconvolved beam data) were used to commission Pinnacle\(^3\) for the 6 MV photon beam of the Trilogy and three photon beam models (BM6, BM4 and BM08, respectively) were created. A 2x2x2 mm\(^3\) voxel size was used during the commissioning. Task Group report 53 (TG-53) of The Radiation Therapy Committee of the American Association of Physicists in Medicine\(^69\) was followed during the commissioning process.

**Patient Specific IMRT QA Comparison**

Five head & neck (H&N) IMRT plans and three prostate IMRT plans (for a total of 53 fields) previously used for patient treatment on the Trilogy were randomly selected for this study. The plans were generated using the DMPO (Direct Machine Parameter Optimization) option in the TPS, which optimized the shape and weight of each MLC segment directly. The three beam models BM6, BM4 and BM08 were used to calculate planar dose distribution for each field at 10 cm depth and 90 cm SSD in a rectangular water phantom with 0° gantry angle (IEC convention). All the calculation was done using a 1.0 x 1.0 mm\(^2\) dose grid. The measurement was performed in a solid water phantom using a 2-dimensional (2D) diode array (MapCHECK 1175, Sun Nuclear Corp., Melbourne, FL) under the same geometry. The MapCHECK device has been demonstrated to be an excellent tool for the verification of IMRT fields with little energy or dose rate dependence.\(^{38,39}\) The small size of its detectors (0.8x0.8 mm\(^2\)) ensures minimal volume averaging effect. An intercomparison between film and MapCHECK showed that MapCHECK can replace film dosimetry in routine IMRT QA.\(^{40}\) The MapCHECK device was calibrated for dose distribution measurement based on the manufacturer-recommended procedures. The calculated planar dose distributions were imported into the MapCHECK analysis software. Each of the MapCHECK measured points above 10% of the maximum dose level was compared with calculation using %Diff and DTA acceptance.
criteria in the absolute dose comparison mode. Two sets of criteria, 2%/2mm and 3%/3mm, were employed, and the percentage of the points passing the acceptance criteria was evaluated.

**Results**

Figure 3-1 shows the comparison of the cross beam profiles measured with the Edge Detector, CC04, and CC13 for a 10x10 cm² field at 10 cm depth in water phantom with SSD=90 cm. The 80%-20% penumbra widths are 0.40 cm, 0.61 cm and 0.72 cm for the diode, CC04 and CC13, respectively. The penumbra was broadened by 0.2 ~ 0.3 cm using the commonly used ion chambers. The difference is consistent with published results. This necessitates the need to deconvolve the profiles measured with finite size ion chambers.

The beam modeling process is as important as the beam data accuracy, especially for MLC-delimited small field sizes. Figure 3-2 (a) and (b) show diode measured beam profiles and TPS calculated beam profiles with three different beam models for MLC-delimited 2x2 cm² and 4x4 cm², respectively. The agreement in penumbra (80% to 20%) between diode measured data and beam model BM08 calculated data was within 1 mm, which met the criteria recommended by TG-53. Similar agreement was achieved between ion chamber measured data and TPS calculation with the corresponding beam models.

The derived parameters for the detector response function for CC13 are summarized in Table 3-1 and demonstrated in Figure 3-3 for the two field sizes and at three depths. These were determined using Equation 3-1 in a least squares fit and assuming that the diode measured profiles are the “true” cross beam profiles. Figure 3-3(a) shows the results for 4x4 cm² field size. The convolution between the diode measured data and the best fit Gaussian function at each depth agrees well with the CC13 measured profile. The Gaussian function parameters \( \sigma_k \) obtained at depths of 1.5 cm, 10 cm and 25 cm were 0.68 cm, 0.68 cm and 0.70 cm, respectively.
When converted to the sigma of a standard Gaussian function ($\sigma_s$) by multiplying $\sigma_k$ by a factor of 0.4, they correspond to 0.27 cm, 0.27 cm and 0.28 cm, respectively. Figure 3-3(b) shows the results for the 10x10 cm$^2$ field size. The optimal parameters $\sigma_k$ obtained were 0.68 cm, 0.72 cm and 0.69 cm at depths of 1.5 cm, 10 cm and 25 cm, respectively, which correspond to 0.27 cm, 0.28 cm and 0.28 cm when converted to $\sigma_s$. These are very close to those obtained with the 4x4 cm$^2$ field size. The results indicate that the detector response function is independent of both field size and depth in the ranges studied. The derived $\sigma_s$ is also very close to the radius (0.3 cm) of the CC13 chamber, which agrees with the findings of other works.$^{46,54}$

The analytic fit using Equation 3-8 to the CC13 measured cross beam profiles for field sizes 4x4 cm$^2$ and 10x10 cm$^2$ at the three different depths are shown in Figure 3-4. By restricting the analytic fit to high gradient regions of the profiles and extending the analytic function from Equation 3-5 to Equation 3-8, excellent fit can be achieved at all depths (1.5 cm, 10 cm and 25 cm). The deconvolved profiles, obtained by deconvolving the analytically fitted profiles with the detector response function obtained above, agree well with the diode measured profiles at each depth. The advantage of using the analytic functional form of Equation 3-8 to fit the measured profiles is that deconvolution can be solved analytically and is simply a replacement of the shape parameter $\sigma_m$ with the new one based on Equation 3-7. The results validated the procedures of extracting the true profiles with the analytic fit/deconvolution method.

The results of applying the above method to extract the true profiles for larger field sizes and assuming that the detector response function is independent of field size and depth are illustrated in Figure 3-5, where results for field size 30x30 cm$^2$ at three depths are shown. Excellent analytic fit was also obtained for field sizes 20x20 cm$^2$ and 40x40 cm$^2$ at all depths (not shown). Figure 3-6 shows the results of extracting the true profiles for MLC-defined
asymmetric fields. Asymmetric profiles were divided into two halves along the center of the field and were treated separately. The deconvolved profiles were then merged together to form the complete profiles. Figure 3-6 shows that the same functional form that fit well the collimator defined fields was equally effective in fitting the MLC defined fields. By restricting the analytic fit to high dose gradient regions of the profiles, we were able to apply the analytic fitting method to larger field sizes which would otherwise fail in trying to reproduce the “horn” effect.

The passing rates between the measured and calculated planar dose distributions for the 8 clinical cases (53 treatment fields) using both the 2%/2mm and 3%/3mm criteria are summarized in Figure 3-7, where the passing rates were binned into 5% steps for the three different beam models. Table 3-2 lists the average passing rates for all the cases along with their associated standard deviations. The data demonstrates a clear trend where beam models commissioned using profiles obtained with smaller detectors have higher passing rates than those commissioned using profiles obtained with larger detectors. The deconvolved beam model BM08 gave the best passing rates among the three beam models with an average passing rate of 96.8% when evaluated with the 2%/2mm criterion, compared with average passing rates of 92.9% and 81.6% for beam models BM4 and BM6, respectively. For beam model BM08, all the cases achieved passing rates of > 90% and ~76% of which were above 95%. For beam model BM4, 81% of the cases had passing rates greater than 90% and 31% of which were above 95%; while for beam model BM6, only 10% of which were above 90% and none of which was above 95%. When the criterion of 3%/3mm was employed, both BM4 and BM08 achieved excellent results with passing rates for all the cases greater than 95%. The average passing rates for both beam models were above 98%. However, for beam model BM6, although above 86% of all the cases had
passing rates greater than 90%, only 37% of which were above 95%. The average passing rate for BM6 was only 93.8% with the criterion of 3%/3mm.

Figure 3-8 shows an example of calculated planar dose distributions using three different beam models and their comparison to MapCHECK measurement using 2%/2mm criterion. In this case, the passing rates were 85%, 96% and 98% for BM6, BM4 and BM08, respectively. The failing points were indicated on each calculated dose distribution with stars and squares representing hot spots and cold spots, respectively. It was clearly shown on Figure 3-8(a) that most failing points occurred around the high gradient region. Also plotted in Figure 3-8(d) was a comparison among calculated in-plane dose profiles using three different beam models and the MapCHECK measurement. The calculated profiles were taken from the same position of each calculated planar dose distribution and the position was indicated by the arrows. The calculated beam profile with BM6 was characterized with more rounded peaks and valleys and slower gradient compared with calculated profiles with the other two beam models due to detector size effect. The detector size effect was less pronounced with BM4. Similar dose values were achieved with all three beam models in low dose and high dose but low gradient regions.

**Discussion**

There has been a great interest in extracting the “true” profiles from profiles measured with finite sized detectors. In this study, an analytic deconvolution procedure was developed based on the work of Garcia-Vicente.\(^{51}\) The analytic function was extended and a gradient based selecting approach was introduced to achieve good fit to the measured profiles.

The “true” profiles for small field sizes (4x4 cm\(^2\) and 10x10 cm\(^2\)) were collected using a small diode detector. In literature, it was usually obtained by profile measurement using various size detectors and extrapolation down to zero size detector.\(^{46-50}\) This process requires many detectors and is very time consuming. In addition, error arises around zero size region where the
linear relationship assumed during the extrapolation may not hold. Other options are film and diode detector. Compared to ion chamber, film has excellent spatial resolution and has been used in the determination of the detector response function. However, accurate film dosimetry is time consuming and it has the problem of over-response to low energy photons for radiographic films. Diode detectors have over-response for broad beams (over 10×10 cm²) due to energy and directional dependence. But the influence is minimal when the diode detector is used to measure the penumbra of small field sizes (≤ 10×10 cm²). The active measuring area of the diode detector is small enough to resolve the volume averaging effects. It has been shown that the penumbra widths measured by silicon diodes are very close to those measured by film and diamond detectors.44, 48, 71

When the “true” profile is obtained, numerical procedures such as deconvolution or inverse Fourier transform can be used to derive the parameter of the detector response function. As mentioned above, numerical procedures suffer from the measurement noise or uncertainty and the results could be disconcerting. In this work, a curve fitting procedure was used to derive the Gaussian parameter for the detector response function. The true profiles were compared with the CC13 measured profiles after convolution with the Gaussian function in a least squares sense. The influence of the measurement noise was minimized. Very reliable and consistent Gaussian parameters were achieved throughout different geometries (field sizes of 4×4 cm² and 10×10 cm² and depths of 1.5 cm, 10 cm, and 25 cm, see Table 3-1). The sigma of the corresponding standard Gaussian function falls in 0.27~0.29 cm, which is very close to the radius of CC13 ion chamber (0.3 cm), similar to the findings of previous works. Bloch and Wallace found that the broadening of the penumbra was independent of the depths of scan in phantom. The results of this work are in agreement with this finding. In addition, we found no change in the detector
response function from field sizes $4 \times 4 \text{ cm}^2$ to $10 \times 10 \text{ cm}^2$. We therefore assumed that the detector response function derived from small field sizes can be applied to large field sizes.

Gaussian function and the difference of error functions have been used to fit photon beam profiles. However, they are limited to small field sizes and can’t reproduce asymmetric or “horn” shaped profiles. In this work, the extended difference of error functions was used to fit only the high dose gradient regions of an arbitrary profile. The gradient based selection approach was motivated from the observation that only high dose gradient regions of the profiles suffer from volume averaging. The high dose, low gradient region, characterized by unflatness or horns, is excluded in the fitting process. Asymmetric profiles were divided into two half profiles along the field center and treated separately. In this way, the analytic fitting can be performed for profiles of arbitrary field sizes.

The purpose of IMRT patient specific QA is to detect any errors in the chain of IMRT planning and delivery process. When the errors introduced by the volume averaging effect of the beam commissioning data are significant, the QA results will be less sensitive to any other dosimetric errors. This would defeat the purpose of patient specific QA. When 3%/3mm criterion was used, the averaging passing rates for 57 IMRT fields were increased by 5% from $\sim94\%$ to $\sim99\%$ when the detector size was decreased from 0.6 cm to less than 0.4 cm. When using 2%/2mm criterion, 10% increase in averaging passing rates was achieved with detector size decrease from 0.6 cm to 0.4 cm. A further increase of 4% was achieved from the CC04-measured beam model to the deconvolved beam model. These results confirmed the conclusion of Esch et al that the main source of discrepancy between calculated and measured dose distribution was volume averaging effect caused by the finite size of the detectors used for collecting the data for TPS beam modeling. Unsatisfactory IMRT QA results caused mainly by
detector size effect could lead to wrong clinical decisions\cite{48} and probably a waste of clinical resources (e.g., in trying to identify the sources of error for each questionable IMRT field, redo the IMRT plan\cite{11}, etc). On the other hand, when the detector size effect is removed, a stringent acceptance criteria (such as 2%/2mm) in IMRT patient specific QA could be employed. Thus the QA procedure would be more sensitive to systematic uncertainty in the system. For example, a recent study showed that a 1 mm systematic error in MLC positioning could lead to clinically significant dosimetric errors.\cite{26} The deconvolved beam mode combined with more stringent acceptance criteria such as 2%/2mm would be more likely to detect the MLC positioning error. This is the subject of our future study.

**Conclusions**

A simple analytic approach has been developed to deconvolve ion chamber measured profiles and extract the “true” profiles at arbitrary field size and depth. A series of difference of error functions and a gradient based fitting approach has been used to fit the measured profiles. The deconvolution procedure has been demonstrated to be effective under different geometries. CC04 and CC13 measured beam profiles as well as the deconvolved profiles were used to commission a commercial TPS. Agreement between calculated and diode array measured planar dose distribution were evaluated using percent dose difference and DTA criteria. The passing rates were increased significantly with the decrease in the detector size used to obtain the commissioning beam profiles, with an average passing rate of ~96% using the 2%/2mm criterion for the beam model using the deconvolved beam profiles. These results confirm the hypothesis that detector size effect is the main source of discrepancy in IMRT patient specific QA. When detector size effect was removed from the measured beam profiles for commissioning, more stringent criteria can be employed clinically which will likely result in higher chances of detecting dosimetric errors caused by TPS or MLC positioning or other mechanical errors.
Table 3-1. Best-fit shape parameter $\sigma_k$ of the detector response function for CC13 at depths of 1.5 cm, 10.0 cm and 25.0 cm for field sizes of 4x4 cm$^2$ and 10x10 cm$^2$. The sigma of the corresponding standard Gaussian function ($\sigma_s$) is obtained by multiplying $\sigma_k$ by a factor of 0.4.

<table>
<thead>
<tr>
<th>Depth</th>
<th>1.5 cm</th>
<th>10 cm</th>
<th>25 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$\sigma_k$ (cm)</td>
<td>$\sigma_s$ (cm)</td>
<td>$\sigma_k$ (cm)</td>
</tr>
<tr>
<td>4x4 cm$^2$</td>
<td>0.68</td>
<td>0.27</td>
<td>0.68</td>
</tr>
<tr>
<td>10x10 cm$^2$</td>
<td>0.68</td>
<td>0.27</td>
<td>0.72</td>
</tr>
</tbody>
</table>

Table 3-2. Average passing rate and standard deviation of the IMRT QA results for 53 fields for the three photon beam models. The measurement was performed with the MapCHECK device. IMRT plan was generated using the BM4 beam model for each patient. The same beam parameters (MUs, segment weights, etc) were then employed to calculate planar dose distributions with each of the three beam models. Percent dose difference (%Diff) and distance-to-agreement (DTA) acceptance criteria were used to evaluate the agreement between calculated and measured dose distribution.

<table>
<thead>
<tr>
<th></th>
<th>CC13 model (BM6)</th>
<th>CC04 model (BM4)</th>
<th>Deconvolved model (BM08)</th>
</tr>
</thead>
<tbody>
<tr>
<td>2%/2mm</td>
<td>81.6±7.8</td>
<td>92.6±3.3</td>
<td>96.8±2.3</td>
</tr>
<tr>
<td>3%/3mm</td>
<td>93.8±4.1</td>
<td>98.9±1.0</td>
<td>99.4±0.6</td>
</tr>
</tbody>
</table>
Figure 3-1. Comparison of cross beam profiles measured with the Edge Detector, CC04, and CC13 for a 10x10 cm² field of a 6 MV photon beam from Varian Trilogy at 10 cm depth and 90 cm SSD. The 80%–20% penumbra distance is 4.0 mm, 6.1 mm and 7.2 mm, respectively.
Figure 3-2. Diode measured beam profiles and the corresponding TPS calculated beam profiles using three different beam models (“BM6”, “BM4” and “BM08”) for (a) MLC-delimited 2x2 cm² and (b) MLC-delimited 4x4 cm².
Figure 3-3. Comparison of the diode measured (“diode”) and CC13 measured (“CC13”) cross beam profiles at three depths (1.5 cm, 10.0 cm, and 25.0 cm) for field sizes of 4×4 cm² (a) and 10×10 cm² (b) at 90 cm SSD. The convolved profiles (“convolved”), obtained by convolving the diode measured profiles with the best-fit detector response function, are also shown. The optimal Gaussian Sigmas obtained at three different depths for the two field sizes are listed in Table 4-1.
Figure 3-4. Analytic fit using Equation 3-8 to the CC13 measured cross beam profiles (“CC13”) for field sizes of 4x4 cm$^2$ (a) and 10x10 cm$^2$ (b) at depths 1.5 cm, 10.0 cm, and 25.0 cm. Also plotted are the deconvolved profiles from the analytic fit and the diode measured profiles (“diode”).
Figure 3-5. Analytic fit using Equation 3-8 to the CC13 measured cross beam profiles ("CC13") for field size 30x30 cm$^2$ at depths 1.5 cm, 10.0 cm, and 25.0 cm. Also plotted are the deconvolved profiles from the analytic fit using the detector response function obtained from the smaller field sizes of 4x4 cm$^2$ and 10x10 cm$^2$.

Figure 3-6. Analytic fit using Equation 3-8 to the CC13 measured cross beam profiles ("CC13") at 10 cm depth for MLC-defined asymmetric fields. One MLC bank was fixed at 15.0 cm and the other one was placed at 5.0 cm, 0 cm, and -5.0 cm in the cross-plane direction. The profiles were measured in the cross-plane direction, which is parallel to the direction of the MLC leaf travel. Also plotted are the deconvolved profiles from the analytic fit using the detector response function obtained from the collimator defined field sizes of 4x4 cm$^2$ and 10x10 cm$^2$. 
Figure 3-7. Comparison of the passing rates between the MapCHECK measured and calculated planar dose distributions for 8 clinical cases (53 treatment fields) using three different beam models ("BM6", "BM4" and "BM08"). Comparison was done using the MapCHECK analysis software (version 2.01) in the “AD” mode with both (a) 2%/2mm and (b) 3%/3mm DTA criteria. The passing rates were binned into 5% steps.
Figure 3-8. Example of calculated planar dose distributions using three different beam models and the comparison with MapCHECK measurement using 2%/2mm criterion (a) BM6, 85% (b) BM4, 96% and (c) BM08, 98%. Hot spots and cold spots were indicated by stars and squares respectively. Also plotted in (d) were calculated in-plane profiles compared with MapCHECK measurement. The profiles were taken from the same position of each calculated planar dose distribution and the position was indicated by arrows.
CHAPTER 4
INDEPENDENT PLANAR DOSE CALCULATION FOR IMRT QA

Introduction

Patient-specific quality assurance in intensity-modulated radiation therapy (IMRT) involves the verification of treatment parameters generated from the treatment planning system. In most places, IMRT plans are verified with independent measurement in phantom. This has been done with ionization chambers, radiographic or radiographic films, thermo-luminescent dosimeter (TLD) chips, electronic portal imaging devices or other two dimensional detector arrays. These time-consuming measurements require actual beam delivery and thus are usually performed after hours in a busy clinic. Alternatively, IMRT plan verification can be conducted through independent computational algorithms. Algorithms for both stereotactic radiosurgery and large size multi-leaf collimator (MLC) shaped fields have been reported. In these planar dose calculation algorithms, dose output factor can be separated into two components: a head scatter factor and a phantom scatter factor. Head scatter factor accounts for radiation scattered by the machine head structure reaching the point of calculation in-air while phantom scatter factor accounts for the scattered dose deposition within the phantom. Typically, phantom scatter factors are calculated through the convolution between in-air fluence distribution and a dose deposition kernel which is either generated via Monte Carlo simulation or an analytic kernel with parameters fitted from measured profiles.

Recently, Mihaylov et al compared the accuracy of dose calculation using both analytic methods and Monte Carlo method to predict fluence modulation. Although using analytic methods for fluence modulation calculation was found to be not as accurate as when Monte Carlo method was employed, analytic methods are easy to implement and can potentially achieve the desired accuracy for independent MU and fluence verification for IMRT plans. Many
different analytic photon source models for head scatter factor calculation have been proposed. Most of the source models consist of a primary (focal) point source at the X-ray target and one or more extra-focal sub-sources representing scattered photons from the primary collimator, flattening filter and/or other head structures. Some source models explicitly model the radiation backscattered into monitor chamber from the collimator jaws. The main focus of this paper is to compare three different source models regarding their ability to model head scatter factors using a comprehensive set of data measured on a linear accelerator (LINAC) as well as facilitate planar dose calculation.

**Methods and Materials**

**Head Scatter Source Models**

Head scatter source models use empirical models to parameterize the head scatter in LINAC head structure. Head structure components in the beam line of a typical LINAC include the X-ray target, primary collimator, exit window, flattening filter, monitor chamber, mirror and collimator jaws. Results of Monte Carlo simulations showed that more than 90% of the scattered photons are produced in the primary collimator and the flattening filter. Dose contribution to the isocenter from other scattering components including the exit window, monitor chamber, mirror and collimator jaws is negligible. Thus, when developing a head scatter source model, we could model the contribution from the X-ray target, primary collimator and flattening filter only and ignore other components. Ding et al also showed the backscatter from the collimator jaws to the monitor chamber plays a significant role in the beam output factors. So it is likely one has to model the backscatter effect explicitly in order to achieve a very accurate head scatter source model. The three different source models compared in this work are the three-source model, the dual-source model and a single source model, all modeled for one linac manufacturer (Varian Medical Systems). The three-source model has three
sub-source components which are based on the results of Monte Carlo simulation\textsuperscript{77}. The dual-source model has two sub-source components and it models the backscatter radiation explicitly. The single-source model has only one source component. Our objective is to achieve faster head scatter calculation while still maintaining acceptable accuracy for IMRT dose calculation. In this section, we briefly describe these three different source models.

Three-source model. The three-source model proposed by Yang \textit{et al}\textsuperscript{78} used one point source for the primary photons coming from the X-ray target, and two extra-focal photon sources for photons scattered from the primary collimator and the flattening filter, respectively. There are three contributing factors to the total fluence at a point of calculation: (1) an isotropic point source at the X-ray target with intensity $C_p$ (for all source intensities in this paper, the unit 1/MU, i.e., per monitor unit, is assumed); (2) a planar annulus source about 4.0 cm downwards from the target with 0.2 cm inner radii and 1.4 cm outer radii, the intensity of which is constant ($C_{sp}$) inside the annulus and zero elsewhere; (3) a planar disk source about 12.5 cm downwards from the target and the source intensity $I_{sf}$ could be expressed analytically as

$$I_{sf}(r) = \frac{A_0}{r} \exp(-kr)$$

(4-1)

Where $A_0$ and $k$ are constants and $r$ is the radial distance from a point on the source plane to the center of the source. The shape, location and size of each sub source component are motivated from Monte Carlo simulation\textsuperscript{77,83}. Backscatter radiation from the upper surface of the jaws into the monitor chamber is not separated or modeled directly. Instead, the effect is taken into account when fitting model parameters using head scatter factors of different square field sizes.

The fluence at the point of calculation was derived by integrating the visible source portion of each sub-source plane from the detector’s eye view. The output factor is simply the fluence.
normalized to the fluence under reference condition. When the calculation point is on the iso-center plane, the head scatter factor is simply

\[ S_c = C_p + \int \int I \sp{sp}(r)d\sigma + \int \int I \sp{sf}(r)d\sigma \]  \hspace{1cm} (4-2)

where \( \sigma \sp{sp} \) and \( \sigma \sp{sf} \) are the visible back projected portions by the jaw openings on the annulus source and disk source, respectively. The fact that the fluence is derived by integrating each source distribution eliminates the singularity problem posed by Equation 4-1 automatically since its integral is always finite.

**Dual-source model.** The dual-source model proposed by Jiang et al\(^{79}\) consists of the primary source and one extra-focal radiation source. The primary photons are modeled by a point source located at the position of the X-ray target. The extra-focal radiation originates only from the flattening filter and it is approximated by the sum of a series of Gaussian functions at the bottom of the flattening filter. The backscatter radiation into the monitor chamber is modeled explicitly. A linear relationship between the backscatter radiation and the irradiated area on the jaw’s upper surface is assumed. So the in-air output factor \( S_c \) for a certain field size \( f_s \) could be calculated as the product of two terms,

\[ S_c (f_s) = (1 + F \sp{efs}(f_s)) \cdot (1 - F \sp{mbs}(f_s)) \]  \hspace{1cm} (4-3)

where the first term describes contribution from scattering source \( F \sp{efs} \) and second term accounts for decrease due to monitor backscatter effect \( F \sp{mbs} \). The scattering contribution is derived by back-projection through detector’s eye view and integration in a similar fashion as in three source model.

**Single-source model.** A numerical single source model was used by Chui et al\(^{75}\) for IMRT planar dose calculation. All the primary and scattering photons were considered to originate from the source plane at the X-ray target for the sake of simplicity. The rotationally symmetric source
distribution peaks at the central axis and monotonically decreases with off-axis distance. The source distribution was represented by a numerical form and samples from the source distribution were obtained by fitting the head scatter factors of square fields with different field sizes. The full source distribution was interpolated from these samples. Backscattered radiation was not considered in this model. The head scatter factor is simply calculated by back-projection and integration on the visible source portion through detector’s eye view.

It is noticed that the numerical source is very sensitive to the error in output factor measurement. To make the source model more stable, we propose an analytic function to approximate the numerical one as follows,

\[
I(r) = \frac{1}{r^a} + \frac{c}{r^b} + d
\]

(4-4)

where \( I(r) \) is the source intensity at a point with radial distance \( r \) from the central axis, and \( a, b, c \) and \( d \) are free parameters.

In the original implementation of the three-source and single-source models, the effect of monitor chamber back scatter was not modeled explicitly. This effect could be incorporated for possibly improved fitting to the measurement data. Therefore, we have also attempted to modify the original three-source and single-source models by explicitly incorporating the effect of monitor chamber back scatter using Equation 4-3.

**Planar Dose Calculation Algorithm**

The algorithm is based on a generalized dose calculation procedure. The flow chart is illustrated in Figure 4-1. Since planar dose distribution at isocenter plane with 10 cm of water-equivalent buildup is used for IMRT quality assurance at our center, we aim at calculating the 40x40 cm\(^2\) dose distribution with 1 mm resolution under the same setting. The parameters that defined the source models were determined from fits to measured Sc data. Back-projection was
done for each point of calculation through the combination of jaw opening and MLC opening from detector’s eye view. In-air fluence distribution was obtained by integrating the visible portion of each sub-source component. Correction to “horn” effect caused by the flattening filter was then applied to the fluence distribution. A pencil beam kernel, generated from Monte Carlo simulation but modified to fit the in-phantom cross-beam profiles for several regular fields, was convolved with the in-air fluence distribution. The resulting relative dose distribution was converted to the absolute dose distribution through the effective field size method. The algorithm is similar to the one described by Chui et al\textsuperscript{75}. Several modifications were applied:

1. In-air fluence distribution is derived based not only on the MLC opening, but the combined openings of jaws and MLC.

2. Off-axis fluence variation due to flattening filter is corrected by using cross-beam in-air scans with appropriate buildup.

3. Pencil beam kernel is expressed as the sum of three Gaussian functions and the free parameters were obtained by fitting the in-phantom beam profiles of several regular square field sizes at the same plane as the dose calculation plane.

The input data for commissioning the algorithm include the following: in-air output factors, in-air scan along the diagonal for the largest field size, square field beam profiles in water and output factors in water at the depth of interest.

**Calculation of fluence distribution.** In model-based dose calculation algorithms, the sub-source components are considered iso-tropical within a certain solid angle\textsuperscript{79}. Thus, the fluence at any point of calculation can be calculated as the integration over the visible portion of each sub-source components from the calculation point’s eye view as described before. In step and shoot
mode, the total in-air fluence $\Phi(x, y)$ at point $(x, y)$ for an IMRT field is the sum of the contribution from each segment, and is given by

$$\phi(x, y) = MU \cdot \sum_{k=1}^{K} f_k \phi_k(x, y)$$  \hspace{1cm} (4-5)$$

where $MU$ is the total monitor unit delivered for the field, $K$ is the total number of segments within this field, $f_k$ is the fractional monitor unit for segment $k$ and $\Phi_k(x, y)$ is the fluence delivered in segment $k$, given by

$$\phi_k(x, y) = \sum_{s=1}^{S} \int_{\sigma_s} I_s(\sigma) d\sigma$$  \hspace{1cm} (4-6)$$

where $S$ is the total number of sub-source components in the head scatter source model, $\sigma_s$ is the visible back projected portion on sub-source $s$ from point $(x, y)$, and $I_s$ the source intensity distribution on sub source $s$. When determining the visible portion from the point of calculation, we have to consider the combined opening of the jaws and MLC when doing back-projection onto the source plane. One way of achieving this is to back project the MLC and jaw openings separately onto the source plane as $\sigma_{jaw}$ and $\sigma_{MLC}$, respectively, and then find their intersection. Thus $\sigma_s$ can be expressed as

$$\sigma_s = \sigma_{jaw} \cap \sigma_{MLC}$$  \hspace{1cm} (4-7)$$

This process is computationally time-consuming. Thus the more source planes (sub source components) there are in the source model, the more computer time is needed to calculate delivered fluence distribution. The inter-leaf and mid-leaf transmission, tongue and groove effect, rounded leaf end of MLC and jaw transmission were taken into account while determining $\sigma_{jaw}$ and $\sigma_{MLC}$, using the methods described by Chui et al.\textsuperscript{75}
Horn effect correction. It is well known that the flattening filter causes “horn” effect on the beam profile perpendicular to the beam central axis. The central portion of the beam suffers more attenuation which results in an increased number of photons at lower energy with off axis distance. The combined effect is the “horn” effect on the cross beam profiles. To correct for this effect, an in-air scan with appropriate buildup was done along the diagonal of the largest open field (40x40 cm^2 for the Varian Trilogy). The “horn” effect correction factor, or the off-axis ratio (OAR), is the ratio between this in-air scan $\Phi_{scan}$ and the corresponding calculated $\Phi_{calc}$ from the source model (ref Figure 4-2), and is expressed as

$$OAR(r) = \frac{\phi_{scan}(r)}{\phi_{calc}(r)}$$  \hspace{1cm} (4-8)

where $r$ is the radial distance from the point of calculation to the beam central axis. This correction factor was applied to in-air fluence calculated for any IMRT field.

Pencil beam kernel generation. The dose distribution is calculated by pencil beam convolution

$$D(x, y) = \phi(x, y) \ast h(x, y) = \int \int \phi(x', y')h(x-x', y-y')dx'dy'$$  \hspace{1cm} (4-9)

where $\Phi(x, y)$ is the delivered fluence distribution and $h(x, y)$ the pencil beam kernel. The pencil beam kernels are typically obtained from Monte Carlo simulation for different nominal energy beams. Monte Carlo simulation considering the full energy spectrum of the nominal energy beam will result in a polyenergetic pencil beam kernel. Usually, photon beams with the energy spectrum of the initial bremsstrahlung photons emitted from X-ray target are employed in the Monte Carlo simulation. However, the energy spectrum of treatment photon beams undergone multiple scattering differ from their initial status. Thus the pencil beam kernel derived from
Monte Carlo simulation has to be adjusted or customized to account for the spectrum change. The sum of three Gaussian functions was used to approximate the pencil-beam kernel,

\[
h(x, y) = \sum_{i=1}^{3} \frac{A_i}{2\pi\sigma_i^2} e^{-\frac{(x^2 + y^2)}{2\sigma_i^2}}
\]

(4-10)

where \(A_i\) and \(\sigma_i\) are the amplitude and standard deviation for the \(i^{th}\) Gaussian, respectively.

Monte Carlo generated pencil beam kernel was used to acquire the initial values for the free parameters, \(A_i\) and \(\sigma_i\), which were then fine-tuned by fitting the model to the in-plane and cross-plane beam profiles of several square fields (e.g., 10x10, 20x20, 30x30 and 40x40 cm\(^2\)) on the isocenter plane. The beam profiles were measured with a 0.2 mm radius ion chamber (CC13) and deconvolved using the algorithm in Chapter 3. The free parameters in the formula were adjusted such that the calculated profiles (via convolution) matched the corresponding measurements in a least-square sense.

**Absolute dose conversion.** Using Equation 4-5 and 4-9, the calculated relative dose distribution \(D_c(\sigma)\) could be expressed as

\[
D_c(\sigma) = \left[ MU \cdot \sum_{k=1}^{K} f_k \phi(\sigma_k) \right] \otimes h = \sum_{k=1}^{K} \left[ MU \cdot f_k \cdot (\phi(\sigma_k) \otimes h) \right]
\]

(4-11)

It is the summation of the contribution from each segment within one IMRT field. If we denote the isocenter dose measured and calculated with field opening \(S_e\) by \(D_m(S_e)\) and \(D_c(S_e)\), respectively, and denote measured and calculated relative output factor with field opening \(S_e\) by \(S_{cpm}(S_e)\) and \(S_{cp}(S_e)\), respectively, the calculated relative dose distribution \(D_c(\sigma)\) could be converted to absolute dose distribution \(D_a(\sigma)\) as following,
\[
D_d(\sigma) = \sum_{k=1}^{K} \left[ \text{MU} \cdot f_k \cdot (\phi(\sigma_k) \otimes h) \cdot \frac{D_m(s_e^k)}{D_c(s_e^k)} \right]
\]

\[
= \sum_{k=1}^{K} \left[ \text{MU} \cdot f_k \cdot (\phi(\sigma_k) \otimes h) \cdot \frac{D_m(s_e^{ref}) \cdot S_{cp}^m(s_e^k)}{D_c(s_e^{ref}) \cdot S_{cp}^c(s_e^k)} \right]
\]

\[
= \sum_{k=1}^{K} \left[ \left( \text{MU} \cdot \frac{D_m(s_e^{ref})}{D_c(s_e^{ref})} \right) \left( f_k \cdot \frac{S_{cp}^m(s_e^k)}{S_{cp}^c(s_e^k)} \right) \cdot (\phi(\sigma_k) \otimes h) \right]
\]

\[
= \left[ \text{MU}' \cdot \sum_{k=1}^{K} f_k \cdot \phi(\sigma_k) \right] \otimes h
\]

(4-12)

where \( s_e^k \) and \( s_e^{ref} \) are the effective square field sizes formed by the MLC shape \( \sigma_k \) and the reference field respectively. They are calculated according to a method described by Dong et al.\(^7^3\) We use \( \text{MU}' \) and \( f_k' \) to denote the modified total and fractional monitor units, respectively. The last step in Equation 4-12 shows the conversion can be done by modifying monitor units and the contribution to in-air fluence from each segment can still be added up and only one convolution (i.e., the one between total in-air fluence and the pencil beam kernel) is needed for each IMRT field.

**Comparison for Head Scatter Calculation**

The three different source models were implemented and were first applied to their original data to ensure correct implementation. The models were then applied to a comprehensive set of \( S_c \) measurement data with a 6 MV photon beam collected on a LINAC with the 120-leaf Millennium MLC (Trilogy, Varian Oncology Systems, Palo Alto, CA). A cylindrical ion chamber (IC-10, Wellhofer Dosimetrie, Germany) in a 4-cm diameter cylindrical mini-phantom at 10 cm depth was used for the measurement. The measurement geometries were very similar to those of Jiang et al.\(^7^6\) Two groups of data were measured for the 6 MV photon beams at isocenter. All calculated and measured in-air outputs were normalized to the reference condition, i.e., at isocenter with 10x10 cm\(^2\) field size. The first group included: (1) in-air output factor \( S_c \) for
square fields from 4x4 cm² to 40x40 cm²; (2) \( S_c \) for symmetric rectangular fields with 10 cm X-jaw (lower jaw in the cross-plane direction) setting and Y-jaw settings from 4 to 40 cm; (3) \( S_c \) for symmetric rectangular fields with 10 cm Y-jaw setting and X-jaw settings from 4 to 40 cm. The entire group data was used to obtain the fitting coefficients for the dual-source model. For the other two source models, only square field \( S_c \) was needed during fitting and the rest of the \( S_c \) was used for verification purpose. The second group was used for additional verification for each of the source models and included two sets of asymmetric rectangular fields, i.e., X jaws set to symmetric 10 cm and 40 cm and Y2 jaw set to 20 cm while changing Y1 jaw from 2 cm to 20 cm.

**Comparison for IMRT Dose Calculation**

The main goal of this work is to compare different source models for the purpose of planar dose calculation for independent IMRT plan verification. Each of the source models was plugged in the planar dose calculation algorithm described above. The calculated planar dose distributions were compared with measurements done with a 2-dimensional (2D) diode array (MapCHECK 1175, Sun Nuclear Corp., Melbourne, FL). The MapCHECK device has been demonstrated to be an accurate and reliable tool for IMRT treatment plan verification\(^{38, 88}\). An intercomparison between film and MapCHECK showed that MapCHECK can effectively replace film dosimetry in routine IMRT QA\(^ {40}\). Seven clinical cases (3 brain and 4 H&N cases for a total of 53 treatment fields) were selected for comparison, all delivered on the Trilogy with 6 MV nominal photon beams and with gantry angle set to 0° (IEC convention). Planar dose distributions were calculated with each of the different source models and compared to measurements done in a rectangular solid water phantom at 10 cm depth and 100 cm source-to-detector distance (SDD). All the calculation was done using a 1.0x1.0 mm² dose grid. The MapCHECK device was calibrated for absolute dose distribution measurement based on the
manufacturer-recommended procedures. Quantitative comparison between the calculation and MapCHECK measurements was done using the MapCHECK analysis software (version 2.01) in the absolute dose comparison mode. The calculated planar dose distributions were imported into the MapCHECK analysis software. Each of the MapCHECK points above 5% of the maximum dose level was compared with calculation using percent dose difference (%Diff) and distance-to-agreement (DTA) acceptance criteria. Two sets of criteria, 2%/2mm and 3%/3mm, were employed, and the percentage of the points passing the acceptance criteria was evaluated.

Results

Comparison of Head Scatter Factors

The results of the three different source models in their fit to the head scatter factor data from a Varian Trilogy LINAC is presented in Figure 4-3, Figure 4-4 and Figure 4-5. The dual-source model achieved the best fitting with standard deviations of 0.05%, 0.08%, and 0.30% for the three data sets in Figure 4-3 (a)–(c), respectively. The original three-source and single-source models without explicit modeling of the effect of the monitor back scatter were less satisfactory in fitting the measured $S_c$ data, with standard deviations of 0.14%, 0.23%, and 0.54% for the three data sets in Figure 4-4 (a)–(c), respectively, for the three-source model and 0.10%, 0.22%, and 0.48% for the three data sets in Figure 4-5 (a)–(c), respectively, for the single-source model. These two models underestimated the head scatter factors for certain symmetric rectangular fields and asymmetric fields. The discrepancy is especially large for symmetric rectangular fields with variable upper jaws and fixed lower jaw opening at X= 4 cm, with deviations about 0.4% at large field sizes (Figure 4-4 (b) and Figure 4-5 (b)). For asymmetric fields, the calculation is systematically lower with average deviations of 0.5%. However, these two models were improved significantly when the effect of the monitor back scatter were modeled explicitly as in the dual-source model, with standard deviations of 0.07%, 0.12%, and 0.33% for the three data
sets in Figure 4-4 (d)–(f), respectively, for the three-source model and 0.07%, 0.21%, and 0.30% for the three data sets in Figure 4-5 (d)–(f), respectively, for the single-source model.

**Comparison of Dose Calculation**

Figure 4-6 shows the comparison between measured and calculated dose profiles for a 10x10 cm² field. The calculation was done using the single-source model with either a Monte Carlo-generated kernel or an experimentally-determined kernel. Using the experimentally-determined kernel noticeably improved the agreement between calculated and measured profiles, especially around the shoulder and foot region. Similar results were also achieved with the other two source models.

The passing rates between the measured and calculated planar dose distributions for the 7 clinical cases (53 treatment fields) using both the 2%/2mm and 3%/3mm criteria are summarized in Figure 4-7, where the passing rates were binned into 5% steps for the three different source-models, and Table 4-1 lists the average passing rates for all the cases. The single-source model gave better passing rates than the other two source models when evaluated with the 2%/2mm criterion, with an average passing rate of 96.2% compared with average passing rates of 90.8% and 89.7% for the dual-source and three-source models, respectively. For the single source model, almost all the cases achieved passing rates of > 90%, while about half of the cases had passing rates less than 90% for the other two models. However, comparable average passing rates (>97%, see Table 4-1) were achieved for all the source models if the criterion of 3%/3mm was employed. Figure 4-8 shows an example of the calculated planar dose distribution using the single-source model and a comparison between the measured and calculated profiles with different source models. In this case, the passing rates were 94.5%, 83.9%, and 83.5% using the 2%/2mm criterion and 97.5%, 96.5%, and 96.2% using the 3%/3mm criterion for the single-source, dual-source, and three-source models, respectively.
Computation Time

Computation time is another concern for any dose calculation algorithm to be used in clinic. As described above, the energy deposition process is isolated from the fluence determination in the dose calculation algorithm. For each of the source models, the energy deposition process takes the same amount of computation time, i.e., the time to do a convolution between pencil beam kernel and a two dimensional fluence distribution. However the time spent on fluence determination through back-projection and integration varies from one source model to the other depending on how many sub source planes the source model has. The three-source model takes twice as much time as the other two source models (4 seconds vs. 2 seconds on a Pentium 1.66 GHz laptop computer with 2 GB of RAM) to calculate fluence distribution in air on average for one IMRT field due to its additional sub source component. The dose deposition, described by the pencil beam convolution, takes the same amount of time (2 seconds) while implemented using Fast Fourier Transform. The average total time for each IMRT field is about 6 seconds for the three-source model and 4 seconds for the other two models.

Discussion

Both single-source model and three-source model underestimate the head scatter factors in some cases as shown in Figure 4-4 b & c and Figure 4-5 b & c, especially when the lower jaws (X jaws) are fixed and upper jaws increase to a maximum opening. This could be explained by the shape of the source distribution (see Figure 4-9 a). Each of the three analytical source models uses either exponentially decaying functions or fast decaying reciprocal function to model the extra-focal source distribution. When the jaw opening keeps increasing above a certain field size (10 cm in Figure 4-4 b), the increase in contribution from integrating larger visible area around the tail region of the source distribution becomes insignificant. This becomes worse when fixing the lower jaw opening and increasing the upper jaw opening, where the output factors increase
much faster (see Figure 4-4 b) while large portion of the source is blocked by the lower jaws. From this reasoning, all three source models would fail in predicting the head scatter factors in those cases. However, the major difference between the dual-source model and the other two source models is that the former isolates and models the monitor chamber back scatter effect separately while the other two don’t. The reduced back scatter contribution at large field sizes, which increases the linac output, makes up the fast decaying source. A linear relationship between the backscatter radiation and the irradiated area on the jaw’s upper surface is appropriate if one notices the approximately linear increase in $S_c$ curve with large upper jaw opening (see Figure 4-4 (b)). By modeling this effect explicitly, the dual-source model achieved better agreement between measured and calculated head scatter factors than the other two. On the other hand, significant improvement is observed when one incorporates this relationship into the other two source models. These facts show it is desirable to isolate the monitor chamber back scatter effect when using analytical functions to model head scatter factors. This conclusion is consistent with the finding in Ding\textsuperscript{84}. Fast decaying analytical functions can predict the fast increasing in $S_c$ at small field size while nearly linear increase of $S_c$ at large field size is better modeled by modeling the monitor chamber back scatter explicitly. Since the machine output varies from one LINAC model to another, even from the same manufacturer, the behavior of $S_c$ at large field size differ significantly. For example, when fixing the lower jaw to 4 cm and moving the upper jaw to an opening larger than 30 cm, the head scatter factor at isocenter from Varian high energy machine (Clinac 2100/2300 and Trilogy) is about 1% larger than Varian low energy machine (Clinac 600 C/D).\textsuperscript{78} The $S_c$ curve from the low energy machines tends to flatten out at large field sizes, in which case it can be fit equally well with or without the explicit modeling of the monitor back scatter.
IMRT fields are characterized by many small field segments and sharp gradients. All three photon source models studied in this work have comparable performance in modeling output factors for small fields. Thus the performance in modeling penumbra region dictates the IMRT dose calculation accuracy. In IMRT patient-specific QA with either diode array or film, most of the failing points occur in the high dose gradient regions where the modeling of the penumbra is less accurate. In some dose calculation algorithms, calculated fluence or dose profiles are convolved with a Gaussian function to account for the finite source size. The width of the Gaussian is independent of the off-axis position. This causes problems since the contribution from the extra-focal source varies with the off-axis positions. In our algorithm, the contribution to the penumbra dose from scattering source is calculated from the extra-focal source based on the off-axis position. The calculated penumbra is fit to the corresponding measurement at different off-axis positions by fine-tuning the pencil beam kernel. Figure 4-9 (b) shows the comparison between the measured dose profile for a 10x10 cm² open field and the “horn” effect-corrected in-air fluence profiles calculated from each of the three analytic source models. The convolution between the fluence profiles and the pencil beam kernel results in the calculated dose profiles. The faster decaying source distributions of the dual-source model and three-source model produce sharper corners in the penumbra region, while the calculated fluence profile from the single-source model has rounded shoulder and foot regions which, when convolved with the kernel, agree better with the measured dose profile. This probably explains the better agreement between calculated and measured dose distributions with the single-source model than with the dual- or three-source models.

Patient-specific IMRT plan QA using independent MU calculation algorithms have been reported by many groups and are commercially available. Most of these algorithms
verify dose to one or a few points. Due to the dynamic nature of IMRT delivery, the correctness of the dose at one spatial point does not warrant the correctness of dose at other points. It is therefore necessary to verify fluence/dose distribution for each IMRT field to ensure the derived time sequence of MLC from the TPS is correct\textsuperscript{75,90,95}. Recently, patient-specific IMRT QA using 2D detector arrays, such as diode arrays or ion chamber arrays, has become increasingly popular due to their ease of use and immediate readout of the results. Measurements are usually done in regular phantom with radiation beam orthogonal to phantom surface. Such measurements can potentially be replaced using independent calculation algorithms such as the one proposed in this work. Using computational algorithms to replace phantom measurement as patient-specific IMRT QA has the drawback that it does not verify the functionality of the MLC delivery system for each field, and a supplemental QA of the MLC delivery system needs to be in place. It has been proposed that the QA of the MLC leaf sequences (fluence maps) of a patient treatment and the dynamic MLC delivery system can be separated into two different tasks\textsuperscript{89,95}. A rigorous QA procedure to examine the mechanical and electronic reliability of the MLC delivery system must be established and their dosimetric implications must be well understood. The frequency and extent of the QA of the MLC delivery system, however, are independent issues and should be determined by the need to maintain the normal operation of dynamic delivery system. This may not require an actual delivery check for every IMRT treatment field, similar to that we do not check the dynamic jaw movement for every treatment field involved with a dynamic wedge in conventional radiotherapy treatment. The verification of the dose distribution, on the other hand, should be done for every field in a patient treatment to ensure the accuracy of the planar dose distribution. This separation of QA tasks allows us to use computer simulation to
examine the functionality of the MLC leaf sequences and the delivered fluence/dose map based on a validated calculation algorithm.

The other important aspect that needs to be verified if an independent MU verification algorithm is to be used in place of phantom measurement is the integrity of the data transfer from the TPS to the record-and-verify (R&V) system which controls the linac delivery. In our implementation, the algorithm read in the exported plan files from our R&V system (MOSAIQ version 1.2018, IMPAC Medical Systems, Inc., Mountain View, CA), rather than from the TPS. In this way, treatment parameters that are stored in the R&V system and are used directly for patient treatment are verified and any data transfer errors between the TPS and R&V could be detected.

**Conclusions**

The performance of three different source models regarding their ability to model head scatter factor and facilitate planar dose calculation for IMRT has been evaluated. The comparison was done on an extensive set of measurement data from a 6 MV photon beam of a Varian Trilogy and showed that the dual-source model predicted the head scatter factor most accurately. The original three-source model and single-source model underestimated head scatter factors for asymmetric fields and certain symmetric rectangular fields, but similar good agreement could be achieved when monitor back scatter effect was incorporated explicitly. Each of the source models was used to calculate fluence distribution in-air, which when convolved with an experimentally-determined kernel, resulted in the planar dose distribution in a homogeneous phantom. The comparison on passing rates using percent dose difference and distance-to-agreement criteria showed the single source model achieved the best agreement with measurement when using the 2%/2mm criterion. When the criterion was relaxed to 3%/3mm, all source models achieved comparable passing rates (>97%). Due to the simplicity of the models,
single-source and dual-source models required less computational time on in-air fluence
distribution and overall calculation time.
Table 4-1. The comparison of average passing rates for the 53 IMRT fields while using different source models in the planar dose calculation algorithms. The standard deviation is also given for each case.

<table>
<thead>
<tr>
<th>Criterion</th>
<th>Single-source model (%)</th>
<th>Dual-source model (%)</th>
<th>Three-source model (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>2 %/2 mm</td>
<td>96.2±2.8</td>
<td>90.8±4.4</td>
<td>89.7±4.7</td>
</tr>
<tr>
<td>3 %/3 mm</td>
<td>99.0±1.2</td>
<td>97.7±2.1</td>
<td>97.4±2.4</td>
</tr>
</tbody>
</table>

Figure 4-1. Flowchart of the proposed IMRT planar dose calculation algorithm.
Figure 4-2. Comparison of the calculated fluence as a function of off-axis distance before correcting for the “horn” effect, the measured fluence from in-air scan, and the calculated OAR factor, which is the ratio between the in-air scan along the diagonal direction and the corresponding calculated fluence profile.
Figure 4-3. Comparison between the measured and calculated head scatter factors using the dual-source model for a Varian Trilogy 6 MV photon beam for three different sets of fields: (a) square fields from 4x4 cm² to 40x40 cm² (“Square field”), symmetric rectangular fields with lower jaw opening fixed to 10 cm while upper jaws changed from 4 cm to 40 cm (“X=10 cm”), and symmetric rectangular fields with upper jaw opening fixed to 10 cm while lower jaws moving from 4 cm to 40 cm (“Y=10 cm”). These data were used to fit the model parameters; (b) symmetric rectangular fields with lower jaw opening fixed to 4 cm while upper jaws changed from 4 cm to 40 cm (“X=4 cm”), and symmetric rectangular fields with upper jaw opening fixed to 4 cm while lower jaws moving from 4 cm to 40 cm (“Y=4 cm”); (c) asymmetric rectangular fields with symmetric X jaws set to either 10 cm or 40 cm and Y2 jaw fixed at 20 cm while changing Y1 jaw from 2 cm to 20 cm. All the data were normalized to the results at 10x10 cm² field size.
Figure 4-4. Comparison between the measured and calculated head scatter factors using the original three-source model (a-c), and the modified three-source model to incorporate the monitor back scatter effect explicitly (d-f). The data sets were the same as in Figure 3. In the original three-source model (a-c), only the square field data in (a) were used for the model fitting, while with the modified three-source model, both the square field data and the rectangular field data in (d) were used for the model fitting.
Figure 4-5. Comparison between the measured and calculated head scatter factors using the single-source model without explicit incorporation of the monitor back scatter effect (a-c), and the single-source model with the monitor back scatter effect incorporated explicitly (d-f). The data sets were the same as in Figure 4-3. In the first case (a-c), only the square field data in (a) were used for the model fitting, while in the second case (d-f), both the square field data and the rectangular field data in (d) were used for the model fitting.
Figure 4-6. Comparison between the measured and calculated cross beam profiles at 100 cm SDD and 10 cm water-equivalent depth for a 10x10 cm² field. The calculation was done using the single-source model combined with either a Monte Carlo generated kernel or an experimentally-fitted kernel.

Figure 4-7. Comparison of the passing rates between the MapCHECK-measured and calculated planar dose distributions for 7 clinical cases (53 treatment fields) using the single-source model (“SSM”), dual-source model (“DSM”) and three-source model (“TSM”). Comparison was done using the MapCHECK analysis software (version 2.01) in the “AD” mode with both the 2%/2mm (a) and 3%/3mm (b) DTA criteria. The passing rates were binned into 5% steps.
Figure 4-8. Example of a calculated planar dose distribution using the single-source model (a) and its in-plane and cross-plane profiles compared with the profiles from MapCHECK-measured and calculated profiles using the dual-source model (DSM) and the three-source model (TSM) (b). The positions where the profiles were taken are indicated by the arrows. The cross-plane (horizontal arrow) profile was scaled by a factor of 1.5 to make the two profiles more distinguishable.

Figure 4-9. (a) Source intensity distribution of the modified single-source model (“SSM”), the dual-source model (“DSM”) and the three-source model (“TSM-1” and “TSM-2”) modeled for Varian Trilogy. TSM-1 and TSM-2 represent the annulus source for the primary collimator and the disk source for the flattening filter, respectively (b) Comparison between measured dose profile (“Dose”) in a water phantom for a 10x10 cm² field at 10 cm depth and “horn” effect-corrected in-air fluence profiles calculated using the single-source model (“SSM”), the dual-source model (“DSM”) and the three-source model (“TSM”). In-air fluence profiles for DSM and TSM overlap each other. The relatively slower decaying source of the single-source model results in rounded profile corner.
CHAPTER 5
THE CALIBRATION OF A NOVEL 4D DIODE ARRAY

Introduction

It has become standard practice to use planar two dimensional (2-D) detectors (film, diode array, ion chamber array and electronic portal imager) for IMRT QA. Due to the planar design of those detectors, IMRT beams are usually delivered with fixed gantry angle (0°, IEC convention) and beam incident direction perpendicular to the detector plane\(^3\). One obvious limitation of this QA solution is that potential dosimetric and multi-leaf collimators (MLC) positioning errors during gantry rotation can’t be detected. The emergence of innovative rotational radiotherapy delivery techniques such as dynamic arc therapy, helical TomoTherapy and volumetric modulated arc therapy (VMAT) poses greater challenge for an appropriate QA solution.

Rotational delivery modalities are characterized by many time-varying parameters, such as dynamic MLC, dynamic gantry speed and variable dose rate. In the pre-treatment verification of these radiotherapy modalities, dosimetric measurement would be ideally performed in arc mode. An ideal dosimeter for treatment verification would provide absolute dose distribution in a full three dimension (3-D) space\(^9\). Individual beam dosimetry should be available at the end of measurement to identify sources of error.

Gel dosimetry would be an appropriate candidate to fulfill the requirement for a full 3-D dose distribution. Very encouraging dosimetric measurements have been shown with Gel dosimetry\(^97-10^0\). However, the need of expensive magnetic resonance imaging system prevents the wide use of Gel dosimetry in practice. Film in combination with dedicated cylindrical phantom with spiral cavity has been used to sample and measure dose distribution from 3-D subspace\(^1^0^1\). Either radiographic or radiochromic film can be inserted in the spiral cavity. The design gives the system the advantage that beam always comes perpendicular to the film at any
gantry angle which makes it suitable to validate IMRT beams with planned gantry angle and rotational delivery modalities. On the other hand, film dosimetry is a very time-consuming procedure and great caution has to be taken to achieve accurate absolute dosimetry. Many user friendly on-line 2-D detector arrays have become commercially available with which absolute dosimetry can be conveniently performed. These detectors are designed to sample and measure dose within a single plane perpendicular to beam incident direction with fixed gantry. Dedicated phantom has been made to accommodate these 2-D detectors for the use of rotational radiotherapy QA. Usually, 2-D detector in combination with dedicated phantom requires special corrections and results in limited accuracy with oblique beam incidence. In addition, since film and most on-line detector arrays are integral dosimeters and only a composite dose distribution is available at the end of the measurement, correction for detector angular response difference is not possible with varying gantry angles. Individual beam dosimetry would be required to identify the sources of errors in the planning and delivery process especially for rotational delivery modalities with many time-varying parameters. When dosimetric error presents, integral dosimeters can’t provide adequate information for identifying the cause. Clearly, a desired dosimeter would be able to consistently sample dose distribution from 3-D subspace and capable of performing individual beam dosimetry to validate radiotherapy plans and separate sources of error.

A novel four dimensional (4-D) diode array has been developed for this purpose (ArcCHECK™, Sun Nuclear, Melbourne, FL). The prototype of the 4-D diode array contains 124 diodes embedded in a cylindrical acrylic phantom. The diodes are arranged to sample and measure dose distribution in 3-D subspace. Diode signal is updated every 50 ms (the 4th dimension) which makes both composite and per beam analysis possible for any complex
rotational delivery. The unique arrangement makes the diode array present a consistent detector image to beam’s eye view at arbitrary gantry angle. At any beam incident direction, entrance dose and exit dose in the cylindrical phantom will be measured simultaneously. The objective of this work is to calibrate the 4-D diode array for IMRT QA with planned beam angle and VMAT QA. Diode sensitivity and angular response dependence will be accounted for to achieve rotational isotropicity. Algorithms to derive instant beam angles will be developed based on detector signal to interpolate corresponding angular correction factors for each detector update. The calibration will be validated using water tank scanned beam profiles.

Methods and Materials

Description of the 4-D Diode Array

A prototype of the novel 4-D diode array is shown in Figure 5-1(a). The diode array consists of 124 diodes embedded in a cylindrical acrylic phantom with 13.7 cm height, 25.4 cm outer diameter and 15 cm inner diameter. The cylindrical phantom can accommodate a 15 cm diameter inner core composed of various homogeneous or heterogeneous materials. In this work, an inner core made from acrylic was inserted to make a homogeneous phantom. The 124 diodes are arranged on four rings of the same diameter (19.6 cm), giving each diode 3 cm intrinsic acrylic build up. Each ring has 31 evenly distributed diodes as can be easily identified on the CT image (Figure 5-1 b). The four rings are placed in the center with 1 cm spacing along the axis of the cylinder. Positions of diodes on adjacent rings are slightly offset to form a spiral configuration. N-type diodes with 0.8x0.8mm² active measuring area are used in the diode array. All the diodes are mounted on two flexible printed circuit boards (PCB). Diode signals are conducted from the two PCBs to two proprietary designed 64-channel analog-to-digital electrometers. Diode signal is updated every 50 ms. Collected charge in the capacitor is emptied during each signal update such that the detector will never saturate. The fact the diode array
samples and measures dose distribution in 3-D subspace and its capability of real time signal update make it a real 4-D detector.

Unless specified otherwise, the axis of the cylinder was always aligned with gantry rotation axis and center of the cylindrical phantom aligned with iso-center during all the measurements in this work. The source to surface distance (SSD) was 87.3 cm. Position of each diode was carefully marked on the outer surface of the cylinder and a plastic stand was made to support the phantom which provided setup reproducibility. A positive charge was applied to compensate the diode signal leakage during the measurement. Background signal was measured and monitored during each measurement session and subtracted from collected diode signal.

Diode Array Calibration

N-type silicon diodes are used in the diode array. A silicon diode is a p-n junction diode. A p-n junction is formed by combining P-type and N-type semiconductor materials in close contact (Figure 5-2). Diodes are referred as n-type diode or p-type diode depending on the base material. The opposite thinner layer material is referred as doping material. The close contact between these two materials is called depletion region. Radiation will create electron hole pairs inside the body of the diode. Those pairs within diffusion region will diffuse into depletion region. An electric field will be applied to collect those charges. Diode response is very sensitive to its construction. A small difference in the position of p-n junction could cause significant difference in its response. Due to the anisotropic design of p-n junction diode, radiation coming from different direction experiences different amount of attenuation and back scatter caused by the high-Z material surrounding the diode and metal contact used to collect charges. This effect is referred as diode directional response dependence. Diode response also depends on temperature, dose rate and radiation energy spectrum. However, diode sensitivity and directional response dependence are most dominating factors. In this chapter, we will discuss diode calibration,
focusing on correcting the effect caused by these two factors. A few notations will be introduced for this purpose.

**Relative Gantry Angle $\theta$ to diode $m$:** angle formed by two lines, one connecting isocenter and diode $m$, the other connecting isocenter and radiation source (Figure 5-3 a);

$R(m, \theta)$: response of diode $m$ at relative gantry angle $\theta$;

$SD(m)$: intrinsic diode sensitivity of diode $m$;

$SA(m, \theta)$: angular response sensitivity of diode $m$ at relative gantry angle $\theta$. It is normalized at $\theta = 0^\circ$,

$$SA(m,0) = 1$$

(5-1)

$D(\theta)$: absolute dose at the position of the diode of interest at relative gantry angle $\theta$;

Response of diode $m$ with relative gantry angle $\theta$ can be expressed as:

$$R(m, \theta) = SD(m) \cdot SA(m, \theta) \cdot D(\theta)$$

(5-2)

$CFD(m)$: correction factor for diode intrinsic sensitivity for diode $m$;

$CFA(m, \theta)$: correction factor for angular response sensitivity for diode $m$ at relative gantry angle $\theta$;

**Diode sensitivity correction factors**

Minor difference in individual diode construction and electrometer gain could cause significant variation in diode intrinsic sensitivity. A sensitivity correction factor is needed for each diode to account for this effect. A straightforward way to calibrate the diode array is to expose each individual diode under same beam condition and compare their responses. With Equation 5-2, if diode $m$ and a randomly selected reference diode $r$, are irradiated at $\theta = 0^\circ$ under same beam condition, their responses will be

$$R(m,0) = SD(m) \cdot SA(m,0) \cdot D(0)$$

(5-3)
and

\[ R(r,0) = SD(r) \cdot SA(r,0) \cdot D(0) \]  \hspace{1cm} (5-4)

The ratio of responses of diode \( r \) and \( m \) will be,

\[ \frac{R(r,0)}{R(m,0)} = \frac{SD(r) \cdot SA(r,0) \cdot D(0)}{SD(m) \cdot SA(m,0) \cdot D(0)} = \frac{SD(r) \cdot SA(r,0)}{SD(m) \cdot SA(m,0)} \]  \hspace{1cm} (5-5)

Since angular correction factor is normalized at \( \theta = 0^\circ \),

\[ \frac{R(r,0)}{R(m,0)} = \frac{SD(r) \cdot SA(r,0)}{SD(m) \cdot SA(m,0)} = \frac{SD(r)}{SD(m)} \]  \hspace{1cm} (5-6)

Thus, diode sensitivity correction factor for diode \( m \) with respect to reference diode \( r \) is given by,

\[ CFD(m) = \frac{SD(r)}{SD(m)} = \frac{R(r,0)}{R(m,0)} \]  \hspace{1cm} (5-7)

In calibration, the diode array was positioned on the plastic stand with its axis aligned with gantry rotation axis. Either the diode array or the gantry could be rotated such that each time one different diode was irradiated on beam central axis. Gantry rotation was chosen due to the lack of a reliable mechanism to rotate the diode array at the moment. Gantry rotation accuracy was checked before the calibration according to TG-40105. Radiation beams of field size of 8x30 cm\(^2\) and 100 MU were used in each radiation as shown in Figure 5-3 (b). The marker of each diode on the cylinder outer surface was used as reference to rotate gantry and position beam central axis right on top of each diode. CFD for all diodes was obtained with Equation 5-7 after a full gantry rotation.

**Diode directional response correction factors**

The different amount of backscatter with respect to beam incident direction caused by the metal contact and other high-Z packaging materials surrounding the diode partly explains diode directional response dependence. For each diode, directional correction factor is a function of
beam incident angle. Ideally, all diodes should exhibit same angular behavior and all diodes share the same set of angular correction factors. However, due to the difference in individual diode construction and their placement in the cylindrical phantom, different diodes exhibit different angular behavior. As a consequence, individual angular correction factors should be acquired for each diode.

An intuitive way to evaluate diode direction response dependence is to irradiate the diode with the same beam condition and build-up material from different beam direction and observe its response difference. However, it is not practical to pull out each of the 124 diodes and conduct such experiments. The 4-D diode array is designed to present an isotropic detector image in beams eye view. Thus the goal of the calibration is to remove response difference from different directions and achieve isotropic dosimeter response. In other words, the measured dose distribution should be independent of beam incident direction.

In previous section, a full gantry rotation is performed to obtain CFD for each diode. The beam opening (8x30 cm²) is large enough to cover all diodes during each irradiation. Response of diode \( m \) as a function of relative gantry angle \( \theta \) is

\[
R(m, \theta) = SD(m) \cdot SA(m, \theta) \cdot D(\theta)
\]  

as shown in Figure 5-3 (c). The normalized response \( \overline{R}(m, \theta) \) would be,

\[
\overline{R}(m, \theta) = \frac{R(m, \theta)}{R(m,0)} = \frac{SD(m) \cdot SA(m, \theta) \cdot D(\theta)}{SD(m) \cdot SA(m,0) \cdot D(0)} = \frac{SA(m, \theta)}{SA(m,0)} \cdot \frac{D(\theta)}{D(0)}
\]  

(5-9)

Diode sensitivity \( SD(m) \) cancels out since it is independent of gantry angle. Assume the true dose profile as a function of relative gantry angle is given by \( D(\theta) \). The normalized true dose profile would be,
\[
\overline{D}(\theta) = \frac{D(\theta)}{D(0)} \tag{5-10}
\]

Obviously, the difference between diode measured profile and true dose profile is attributable to diode angular response dependence as shown in Figure 5-3 (c). Thus the angular response sensitivity correction factor for diode m is defined as the ratio of true beam profile to the beam profile measured by diode m,

\[
CFA(m, \theta) = \frac{\overline{D}(\theta)}{R(m, \theta)} \tag{5-11}
\]

Plug Equation 5-9 and Equation 5-10 into Equation 5-11,

\[
CFA(m, \theta) = \frac{D(\theta)}{D(0)} \cdot \frac{SA(m,0)}{SA(m, \theta)} \cdot \frac{D(0)}{D(\theta)} = \frac{SA(m,0)}{SA(m, \theta)} \cdot \frac{D(\theta)}{D(0)} \cdot \frac{D(0)}{D(\theta)} \cdot \frac{SA(m,0)}{SA(m, \theta)} \tag{5-12}
\]

which says angular response sensitivity correction factor is the inverse of angular response sensitivity when normalized at \( \theta = 0^\circ \). This makes sense since diode response will be corrected by multiplying with CFA to remove angular response dependence.

In this work, the “true” dose profile will be approximated with treatment planning system (TPS) calculation. The accuracy of TPS calculation will be validated with ion chamber scanning in water tank. The normalized profile calculated with TPS is given by,

\[
\overline{D}_c(\theta) = \frac{D_c(\theta)}{D_c(0)} \tag{5-13}
\]

Plug Equation 5-9 and Equation 5-13 into Equation 5-11,

\[
CFA(m, \theta) = \frac{\overline{D}_c(\theta)}{R(m, \theta)} = \frac{D_c(\theta)}{D_c(0)} \cdot \frac{SA(m,0)}{SA(m, \theta)} \cdot \frac{D(0)}{D(\theta)} = \frac{SA(m,0)}{SA(m, \theta)} \cdot K(\theta) \tag{5-14}
\]

where \( K(\theta) \) is defined as
\[ K(\theta) = \frac{\frac{D_c(\theta)}{D(\theta)}}{\frac{D_c(0)}{D(0)}} = \frac{D_c(\theta)}{D(\theta)} \]  

(5-15)

which accounts for the uncertainty in TPS calculation. Ideally, \( K(\theta) = 1 \). Equation 5-14 will be used to estimate directional response sensitivity correction factor, CFA. Response of each diode as a function of gantry angle can be extracted from the measurement conducted in previous section when determining diode intrinsic sensitivity correction factor. In other words, both CFD and CFA could be obtained during one full gantry rotation.

**Independence of Calibration Factors on Field Sizes**

In this work, a beam of field size 8x30 cm\(^2\) was used to irradiate diodes to obtain diode sensitivity. Diode response is known to depend on beam field size and beam energy spectrum. If individual diode exhibits different response dependence on field sizes, the obtained diode correction factors would depend on the selected field size. Experiments were designed to check diode response dependence on field sizes for the diode array. Four diodes were randomly selected from the diode array. In the first experiment, each diode was irradiated by beams of 100 MU and 4 different field sizes, 4x4 cm\(^2\), 8x8 cm\(^2\), 8x16 cm\(^2\) and 8x30 cm\(^2\) with beam central axis normal to diode surface (Figure 5-4 a). The second experiment was the same as the first one except the beam central axis entered the diode at an oblique angle of about 45\(^\circ\) (Figure 5-4 b). The response of each diode was normalized to the response at field size 8x8 cm\(^2\). The variation among relative diode response of the four diodes was examined at both normal and oblique beam incidence.

**Gantry Angle Derivation Algorithm**

There is one more step in applying directional correction factors. The instant beam incident direction has to be determined to interpolate the corresponding directional correction factors. A
digital inclinometer may be used to monitor the beam angle. However, since gantry will keep moving during rotational therapy and diode signal will be updated every 50 ms, it would be difficult to register measured gantry angle with each signal update. An ideal approach would be to derive gantry angle from instant diode array signal. This section describes such an algorithm for gantry angle derivation.

The algorithm is depicted in Figure 5-5. In general, radiation beam on field edge intersects the ring of diodes twice as beam enters and exits the cylinder. Neighboring diodes around the intersection points supply information to determine exact locations of the intersection points (50% on beam edge). The radiation source should reside on the line connecting beam exit point and entrance point on the ring. The intersection of two beam lines from two sides of field edge was used to locate radiation source and hence beam incident direction. The detailed pseudo algorithm is listed in Table 5-1. Once a measurement is made, the first step is to correct for diode sensitivity which doesn’t depend on beam angle. The unfolded diode profile is shown in Figure 5-5 (left panel) Then Matlab code is used to look for two peaks, one locates on the beam entrance side (EntMax) and the other on the beam exit side (ExtMax) as in Figure 5-5. Starting from EntMax, looking for the first DownSlope. This DownSlope will include the point the beam edge intersects the ring of diodes. A polynomial fit is used to fit a curve along the DownSlope and the 50% point is interpolated on the curve, which is point B on Figure 5-5 (left panel). The search will continue to look for an UpSlope and its 50% point C. Once the search passes ExtMax, another DownSlope and its 50% point D will be located. When the search reaches the end, it circles around to the beginning of the ring and locates another UpSlope and its 50% point A. A straight line is connected from point D to point A. Another line is connected from C to B. The intersection of these two lines will be considered to be the radiation source position as in Figure
The algorithm can be applied to each of the four rings simultaneously to derive four gantry angles and take the average to improve the accuracy.

**Diode Array Calibration Validation**

An independently determined ("true") dose profile is needed to derive diode array directional correction factors. It could come from independent measurement with film or other dosimeters, Monte Carlo simulation or calculation using TPS. In this work, Pinnacle calculated beam profile is used. There are two things we need to validate, (1) Pinnacle calculated beam profile has to be validated independently; (2) Diode array correction factors have to be validated to make sure that isotropicity is achieved. This section describes the validation procedures.

The idea to validate Pinnacle dose calculation is to compare Pinnacle calculated dose profile to dose profile converted from ion chamber scanning in water tank. For this purpose, a rectangular acrylic phantom with dimensions of 45x35x10 cm$^3$ and a 25.4 cm diameter hole was machined to accommodate the diode array. Measurement was performed with the diode array inside the rectangular phantom. Radiation beam of field size 8x30 cm$^2$ and SSD 87.3 cm was chosen to cover all diodes. Dose calculation was performed by Pinnacle on the CT image of the whole phantom with same geometry. A 48x48x48 cm$^3$ water tank (Wellhofer Dosimetrie, Shwarzenbruck, Germany) was used to collect dose profile with the same setup. Cross beam profiles were scanned every half a centimeter along depth direction with a 0.04 cm$^3$ volume and 4 mm diameter ion chamber (CC04, Scanditronix Wellhofer). The cross beam profiles were interpolated to obtain a full 2-D dose plane in 1x1 mm$^2$ resolution. The 2-D dose distribution in water was then converted to dose distribution in acrylic according to TG-21106 by accounting for (1) density scaling from acrylic to water (2) inverse square law correction and (3) excessive scattering condition in acrylic (Figure 5-6). Suppose $D_{acrylic}(d', r, SSD)$ is the dose we want to
determine in acrylic where \(d'\) is the radiological path length in the medium, \(r\) is the field size at the surface, and SSD is the source-surface distance. The dose can be converted from dose measured in water using the following equation:\(^{106}\):

\[
D_{\text{acrylic}}(d', r, \text{SSD}) = D_{\text{water}}(d, r, \text{SSD}) \left( \frac{\mu_{\text{acrylic}}}{\rho_{\text{water}}} \right)^{\text{acrylic}} \left( \frac{ST + d}{ST + d'} \right)^2 \frac{\text{TAR}(r_d, d)}{\text{TAR}(r_d', d')}
\]

(5-16)

where \(ST\) is the radiological path length in air and \(d\) is given by

\[
d = d' \frac{\mu_{\text{acrylic}}}{\mu_{\text{water}}}
\]

(5-17)

and

\[
r_d = r \cdot \frac{ST + d}{ST} \quad \text{and} \quad r_d' = \frac{r_d}{\rho_e}
\]

(5-18)

\(\bar{\mu}\) is the mean linear attenuation coefficient of the beam and \(\rho_e\) is the electron density of acrylic relative to that of water. The last term in Equation 5-16 is the ratio of the tissue air ratio (TAR) of unscaled field size to that of scaled field size accounting for excess phantom scatter in acrylic relative to water. The converted dose distribution was used to validate the calculated dose distribution by Pinnacle and the diode array measurement corrected with both CFD and CFA.

Another experiment was designed to validate isotropicity has been achieved with the calibration. Two open beams with field sizes 8x8 cm\(^2\) and 8x30 cm\(^2\) were measured with three different beam angles (0\(^{\circ}\), 81.4\(^{\circ}\) and 278.6\(^{\circ}\)). Each time one different diode was right under beam center. The measured beam profiles were compared to each other to verify the isotropic detector response.

**Results**

Relative diode response as a function of field sizes for four randomly selected diodes with normal beam incidence is shown in Figure 5-7. Response of each diode is normalized to that of
field size 8x8 cm². Variation among four diodes is within 0.3% at both large field size (8x30 cm²) and small field size (4x4 cm²). Diode response with oblique beam incidence is shown in Figure 5-8 (normalized at 8x8 cm²). Variation at all field sizes is within 0.4%.

Figure 5-9 shows relative diode sensitivity. Relative diode sensitivity exhibits variation ranging from 0.7 to 1.55. Same experiment was repeated five times. It was found variation of diode sensitivity is within 0.6% which agrees with reproducibility results reported in Chapter 6.

Directional responses of four randomly picked diodes are shown in Figure 5-10 (a). Diode responses are normalized to that of normal beam incidence. All diodes exhibit similar pattern with higher sensitivity when irradiated at 90° and 270° and lower sensitivity when irradiated at 180°. Variation in directional response among diodes is on the order of 2%. A 13% variation can be observed from 90° to 180° which gives 0.14% change per degree. Average directional response of all 124 diodes as a function of beam incident angle is shown in Figure 5-10 (b). The error bars show ±1 standard deviation. The average and maximum standard deviations are 1.3% and 1.8%, respectively.

Results to validate diode array calibration and TPS dose calculation using a water tank scanning system are shown in Figure 5-11. Acrylic dose distribution converted from water tank scan agrees with TPS calculation within 0.5% which shows TPS can adequately model the cylindrical diode array. In general, diode array measurement after applying calibration factors and directional correction factors also agrees with converted acrylic dose from water tank measurement within 0.5%. The discrepancy in the low dose region is slightly higher (1%).

Open beam profiles measured at three different gantry angles for field size 8x8 cm² and 8x30 cm² are shown in Figure 5-12. Excellent agreement could be observed among three measurements except in the high gradient region. All three measured profiles agree with Pinnacle
calculation with all points passing 1\% difference and 1 mm DTA criteria. Same good comparison can be found for field size 8x30 cm$^2$.

Table 5-2 shows the results of beam angle derivation based on detector signal for 4 step and shoot IMRT beams delivered with nominal gantry angles. First column of Table 5-2 shows nominal gantry angle. Second column gives the number of valid signal updates (beam on time divided by 50 ms). A beam angle is derived for each signal update of each IMRT beam. The average and standard deviation of derived beam angles are shown in the table. Also shown is the difference between nominal beam angle and average of derived beam angle. The difference is within 0.9° with a standard deviation of ~1.4°.

**Discussion**

It has been realized the current IMRT QA practice is insufficient by checking all beams at fixed gantry angle. Dosimetric effect caused by gantry rotation error, gantry sagging and potential MLC movement error during gantry rotation has to be evaluated. It becomes more critical for rotational therapy which has more time varying parameters. Current IMRT QA dosimeters are limited by their planar design. When not mounted on the gantry, the planar dosimeters have to be placed parallel to the surface of the couch. Thus when gantry rotates, the dosimeter presents an inconsistent image to beam’s eye view which could potentially bias the QA results. The planar dosimeters could be mounted on the gantry and rotated with the gantry. In this way, the beam always sees the same detector image. However, the biggest problem of this solution is that it has no check on gantry rotation. In an extreme case, when the gantry gets stuck and doesn’t move at all, no alarm would be generated by the QA device.

The 4D diode array, ArcCHECK, provided by Sun Nuclear, is mainly designed for rotational therapy QA. The main idea is to maintain a consistent detector image to beam’s eye view. The goal is physically achieved by using a cylindrical phantom and a spiral arrangement of
diodes. Dosimetrically, the device has to be calibrated to present isotropic response. In this work, the calibration procedure consists of two parts, diode sensitivity correction and directional response dependence correction. Diode sensitivity correction factors are obtained by irradiating different diodes under same beam condition and comparing their responses. Direction response dependence correction factors are obtained based on the idea of isotropic response, i.e., measured dose distribution should be independent of beam incident direction.

When an accurate and reliable rotating mechanism is available, the diode array could be rotated such that each time one diode is placed under beam center and irradiated. It is equivalent to rotate the gantry and fix the diode array. In this work, due to the lack of a reliable rotating mechanism, gantry was rotated during the calibration. It is assumed the gantry rotation has minimum error and machine output is independent of gantry angle. The beam chosen to irradiate the diodes has a field size of 8x30 cm² which has a relatively large uniform region in beam center. Thus the calibration procedure is not sensitive to minor gantry rotation uncertainty. The span of the four rings of diodes is 3 cm in the beam center. The four diodes facing the gantry from each of the four rings are considered to receive the same irradiation. Thus all four rings of diodes are calibrated during one rotation.

The experiment designed to study diode response as a function of beam field sizes revealed that all four diodes responded in the same way when beam central axis was normal to diode surface. The variation of relative diode response was within 0.3%. This is within experiment uncertainty when considering machine output variation. This result shows diode sensitivity obtained in this work is independent of the beam field size used in calibration. In another experiment where beam entered diode at an oblique angle of 45°, relative diode response varied
within 0.4% which is also within experiment uncertainty. Thus selected beam field size during calibration shouldn’t affect diode directional correction factors.

Results in Figure 5-9 shows diode sensitivity changes about two folds. The difference in individual diode design partially accounts for the large variation. The exact geometry (shift and orientation) with which each diode is placed in its position (a small air space in the detector phantom) might cause difference also. The air space is about twice the size of the diode which is a big dosimetric concern for both TPS modeling and diode measurement.

The directional response dependence correction factors are derived based on the concept of a consistent detector image. When irradiated from different beam angles, the diode array is supposed to measure the same dose distribution. After diode sensitivity difference is removed, the difference left in measured dose distribution from different beam directions is attributed to diode directional response dependence. The measurement collected for the purpose of correcting diode sensitivity can be reused. All the correction factors are derived from same set of measurements. All diodes show similar pattern in directional response. It over-responds when irradiated from the side and under-responds when irradiated from back and front surfaces. The variation of diode directional responses is well within 2% as shown in Figure 5-10 (b). It might be reasonable to use the average directional response for each single diode. The feasibility will be evaluated in next chapter when discussing diode array clinical applications.

Pinnacle dose calculation, diode array measurement and dose profile converted from water tank scan come to very good agreement (<1%) as shown in Figure 5-11. This finding shows that Pinnacle can model the cylindrical diode array adequately. The corrected diode reading falls right on Pinnacle calculated dose profile which proves the accuracy of the calibration procedure. The measurements from three different beam angles for two different field sizes (8x8 cm² and
8x30 cm²) all fall on the Pinnacle calculation. They all agree each other except around the high gradient region where measurement is extremely sensitive to diode position. This shows the calibration has successfully achieved isotropicity.

The algorithm to derive instant gantry angle can achieve accuracy of within 1° with standard deviation of 1.4°. The purpose of this algorithm is to derive instant gantry angle to interpolate corresponding angular correction factors, not to check accuracy in gantry rotation. The diode directional response changes by only 0.14% per degree. Thus the uncertainty of this algorithm would cause at most 0.2% dosimetric change which is negligible. The full sized ArcCHECK would have 62 diodes on each ring which doubles the density of diodes. There will be more diodes on both the upslope and the downslope. The polynomial fit used to determine the 50% point would be more accurate. Thus it is expected the accuracy of the algorithm would be further increased with the full sized ArcCHECK. Its applicability for gantry rotation QA would be discussed when a full sized ArcCHECK is available. This algorithm relies on finding two beam edges intersecting the ring of diodes. When the beam is too large and go outside of the cylinder, the algorithm will fail to find the radiation source and hence the gantry angle. This would be a limitation of the algorithm. However, for practical IMRT beams and full sized ArcCHECK with more than 20 rings of diodes, it is expected at least one ring could capture both sides of the beam edges and provide enough information to determine beam incident direction.

The current calibration procedure takes about 30~40 mins. Recommended calibration procedure period is once a year. A more efficient calibration algorithm is still in search.

Conclusions

A 4D diode array provided by Sun Nuclear Corporation is described and calibrated in this chapter. Calibration procedure taking into account diode sensitivity and directional response dependence is developed. The calibration factors are validated against Pinnacle calculation as
well as water tank scan. The diode array presents isotropic response after applying calibration factors. A real time algorithm to derive instant beam angles is developed and validated. Corresponding directional correction factors could be interpolated based on derived beam angle and applied to correct instant dose response. The 4D diode array is calibrated and ready for clinical application. Next chapter will examine its dosimetric characteristics and demonstrate its clinical application for both IMRT and VMAT QA.
Table 5-1. Pseudo algorithm to derive instant beam angle based on diode array signal. Refer to Figure 5-5 for an illustration of the algorithm.

<table>
<thead>
<tr>
<th>Step</th>
</tr>
</thead>
<tbody>
<tr>
<td>Apply diode sensitivity correction factors</td>
</tr>
<tr>
<td>Find diode with maximum reading, $EntMax$;</td>
</tr>
<tr>
<td>Start from $EntMax$, look for another peak which is exit peak, $ExtMax$;</td>
</tr>
<tr>
<td>Start from $EntMax$, go through diode ring, look for diodes on downslope</td>
</tr>
<tr>
<td>Polynomial fit downslope and interpolate to get 50% point, $TopDown$;</td>
</tr>
<tr>
<td>Start from $TopDown$, look for upslope in between $TopDown$ and $ExtMax$;</td>
</tr>
<tr>
<td>Polynomial fit upslope and interpolate to get 50% point, $BottomUp$;</td>
</tr>
<tr>
<td>Start from $ExitMax$, look for downslope ;</td>
</tr>
<tr>
<td>Polynomial fit downslope and interpolate to get 50% point, $BottomDown$;</td>
</tr>
<tr>
<td>Start from $BottomDown$, look for upslope in between $BottomDown$ and $Entmax$;</td>
</tr>
<tr>
<td>Polynomial fit upslope and interpolate to get 50% point, $TopUp$;</td>
</tr>
<tr>
<td>Connect a line from $BottomDown$ to $TopUp$;</td>
</tr>
<tr>
<td>Connect a line from $BottomUp$ to $TopDown$;</td>
</tr>
<tr>
<td>Find the intersection of the two lines which is the source position;</td>
</tr>
</tbody>
</table>
Table 5-2. Beam incident angle derivation for four Step and shoot IMRT beams irradiated with nominal beam angles. Diode signal is updated every 50 ms. Beam angle is derived for each valid signal update during the course of beam irradiation. Average and standard deviation of derived beam angles are shown in the table. Also shown is the difference between nominal and derived beam angles.

<table>
<thead>
<tr>
<th>Nominal Beam Angle</th>
<th>Number of Signal Update</th>
<th>Derived Beam Angle</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Mean</td>
</tr>
<tr>
<td>0</td>
<td>159</td>
<td>0.1</td>
</tr>
<tr>
<td>70</td>
<td>164</td>
<td>70.0</td>
</tr>
<tr>
<td>220</td>
<td>220</td>
<td>219.6</td>
</tr>
<tr>
<td>290</td>
<td>133</td>
<td>289.1</td>
</tr>
</tbody>
</table>
Figure 5-1. (a) prototype of a 4D diode array. A plastic stand is machined to support the cylindrical phantom. An acrylic insert is made and put in place during all measurements. (b) A CT slice of the 4D diode array showing one ring of diodes.

Figure 5-2. Diagram of a p-n junction diode. The close contact between n-type silicon and p-type silicon is called depletion region. Electron-hole pairs created inside the diode body diffuse into the depletion region. Electronic field can be applied to collect the charge. Electron backscatter from metal contact and high-Z packaging materials surrounding the diode cause diode directional response dependence.
Figure 5-3. (A) Relative gantry angle $\theta$ is the angle between the norm passing a diode and the norm passing radiation source. $\alpha$ is the angle between Y-axis and the norm passing the diode. Nominal gantry angle $\beta = \alpha + \theta$. (B) Set up to calibrate the diode array. Diode array was rotated by $360^\circ$ with step size of $5.6^\circ$. Each time one diode was irradiated under beam center with a beam of field size $8 \times 30 \text{ cm}^2$ and 100 MU. (C) Normalized beam profile measured by diode as a function of gantry angle and normalized “true” dose profile. The difference between diode profile and true profile is caused by diode angular response dependence. Ratio of “true” dose profile to measured dose profile is used to correct diode angular response dependence.
Figure 5-4. (a) Four randomly selected diodes were irradiated with beam central axis (CAX) normal to diode surface. Each diode were irradiated with four beams of different field sizes (4x4 cm², 8x8 cm², 8x16 cm² and 8x30 cm²). (b) Four randomly selected diodes were irradiated with beam CAX oblique to diode surface (45°). Each diode was irradiated with five beams of different field sizes (4x4 cm², 8x8 cm², 8x16 cm², 8x24 cm² and 8x30 cm²).
Figure 5-5. Algorithm for deriving instant beam incident direction. Left panel shows the typical unfolded dose profile. The algorithm searches for four 50% points using polynomial fitting and interpolation. Right panel: Radiation source position is determined through back projection from two beam edges.

Figure 5-6. Convert dose measured in water to dose in acrylic according to TG-21. A diode was assumed to be at position A in acrylic phantom. Density scaling was performed from radiological path TA in acrylic to TB in water according to the relative electron concentration of acrylic to water. Inverse square law correction was applied along radiological path length (from SB back to SA). Excessive scattering in acrylic was also taken into account during conversion.
Figure 5-7. Relative diode response as a function of field size for four randomly selected diodes. Each diode was irradiated with four beams of different field sizes (4x4 cm$^2$, 8x8 cm$^2$, 8x16 cm$^2$ and 8x30 cm$^2$) with beam central axis normal to diode surface. Diode response was normalized to field size 8x8 cm$^2$.

Figure 5-8. Relative diode response as a function of field size for four randomly selected diodes with beam central axis oblique to diode surface (45°). Five beams of different field sizes (4x4 cm$^2$, 8x8 cm$^2$, 8x16 cm$^2$, 8x16 cm$^2$ and 8x30 cm$^2$) were used for this purpose. Diode response was normalized to 8x8 cm$^2$. 
Figure 5-9. Relative diode sensitivity of all diodes on the diode array.
Figure 5-10. (A) Directional response as a function of beam incident angle for four randomly selected diodes. It is normalized at 0°. (B) Average directional response of all 124 diodes as a function of beam incident angle. The error bars show ±1 standard deviation.

Figure 5-11. Normalized beam profile calculated by Pinnacle (8.0m), measured with diode array and converted from a water tank measurement for an 8x30 cm² beam. Diode array measurement was corrected for diode sensitivity and directional response dependence. Dose distribution measured in water was converted to dose in acrylic according to TG-21.
Figure 5-12. Relative beam profiles calculated by Pinnacle (8.0m) and measured with diode array for a beam of field size (a) 8x8 cm$^2$ and (b) 8x30 cm$^2$. Three measurements were performed with different beam angles (0°, 81.4° and 278.6°) with diode #3, #1 and #2 under beam center respectively.
CHAPTER 6  
CLINICAL APPLICATION OF A NOVEL 4D DIODE ARRAY FOR IMRT AND VMAT QA  

Introduction  

Two-dimensional (2D) dosimeters have been widely used for routine pre-treatment verification of intensity-modulated radiation therapy (IMRT) due to their ease of use and immediate readout. The purpose of patient specific IMRT QA is to detect potential errors in treatment planning and delivery. Sources of errors in treatment planning stage include inadequate beam modeling and dose calculation algorithm errors. Gantry rotation errors and multi-leaf collimator positioning errors with gantry rotation are among the sources of errors in the treatment delivery stage. It is all these complex dynamic components that make patient specific individual IMRT beam QA necessary. Due to the planar design of existing 2-D detectors, radiation beams are delivered with fixed gantry and beam central axis perpendicular to the plane of detectors during patient specific IMRT QA. The IMRT QA procedure is very likely to capture potential errors in the planning and plan transfer. However, errors in gantry rotation, gantry sagging and MLC movement errors with gantry rotation would remain undetected since IMRT beams are not delivered in planned beam angles during QA.  

Rotational therapy delivery techniques such as RapidArc and VMAT utilizes dynamic gantry angle, dynamic MLC and variable dose rate to achieve optimal dose distribution and shorter treatment time. Patient specific QA for these rotational therapy modalities has to be changed significantly from step and shoot IMRT. In rotational therapy delivery, patient specific QA has to be performed with gantry rotation. Existing 2D dosimeters are insufficient to meet the demand of rotational therapy QA.  

A novel 4-D diode array (Sun Nuclear, Melbourne, FL) was developed to meet the need of appropriate and efficient QA solution for both segmental IMRT and rotational radiotherapy. The
isotropic arrangement of diodes in a cylindrical phantom presents a consistent detector image in beams eye view at arbitrary gantry angles. The 50 ms update rate of the detector array makes per-beam analysis possible with any rotational delivery techniques. Calibration procedure was developed in Chapter 5 to correct for individual diode sensitivity difference as well as directional response dependence. An algorithm was developed to derive instant beam incident angle from diode array signal to apply corresponding directional correction factors.

In this chapter, dosimetric characteristics of the 4D diode array such as reproducibility of detector response, linearity as a function of delivered monitor units (MU), detector response dependence on pulse rate and dose rate will be evaluated. Its clinical application for MLC based IMRT and VMAT QA will be demonstrated. The sensitivity of the diode array to gantry rotation errors and MLC positioning errors at different gantry angles is also evaluated by introducing artificial errors.

Materials and Methods

Detector characterization

An Elekta Precise™ was used to characterize the 4-D diode array. Radiation beams with nominal energy of 6 MV were used. When commissioning beam model for this machine, a 0.04 cm³ volume with a diameter of 4 mm ion chamber (CC04, Scanditronix Wellhofer) was used to collect beam profiles. Absolute dose calibration of this machine was performed according to TG-51107 giving 1 cGy/MU at d_max=1.5 cm in water with 100 cm SAD and 10x10 cm² field size. Leaf positions of both leaf banks were calibrated to have accuracy better than 0.3 mm with an in-house procedure33. The 4-D diode array was supported by the plastic stand with the axis of the cylinder aligned with gantry rotation axis and its center coincident with machine isocenter.

The reproducibility of the detector was evaluated by irradiating the detector 20 times over a two hour period with the same setup and beam condition (100 MU and 8x30 cm² field size).
The field size was selected to cover all four rings of diodes and avoid irradiating electronic part of the detector. A 30 second background measurement was performed before each measurement and subtracted from the following measurement. A cylindrical ion chamber (IC-15, Wellhofer, Schwarzenbruck, Germany) was used to monitor machine output constancy. Ten diodes were randomly selected from the top, the bottom and off-axis positions from the cylinder. The diode array reproducibility was evaluated by calculating the relative standard deviations of the selected diodes during the 20 measurements.

Linearity of diode array response was evaluated by irradiating the detector with beams of 8x8 cm² field size and 1~1000 MU. Larger MU could be used since the diode array will never saturate but this range of MU will cover all clinical applications. Response of the top diode on beam central axis was studied. Other diodes were calibrated against the top diode to have same response using procedure described in Chapter 5.

Diodes are known to have instant dose rate dependence. As we know, photon beams are generated when high speed electrons hit target. In medical LINAC, high speed electrons come in the form of pulses as in Figure 6-1 (a). Instant pulse rate is defined as number of pulses delivered per unit time. It is also called pulse repetition rate. Pulse rate is inversely related to the gap between two continuous pulses. The pulse rate could be changed directly from Elekta console. Available pulse rate for Elekta Precise is [25 50 100 200 400] MU/min. Higher pulse rate means less time to deliver the same amount of monitor units or dose. Dose rate is a different concept. It’s the rate of dose deposited in the medium. Pulse rate is directly related to dose rate. Higher pulse rate results in higher dose rate. Another way to change dose rate is to change the distance from the point of interest to the radiation source. Dose rate changes according to inverse square law with the distance from radiation source to the medium. In this
way dose rate changes while pulse rate remains the same. In clinic, pulse rate and dose rate are
used interchangeably. Pulse rate and dose rate dependence of the diode array needs to be
investigated because of two reasons: (1) Diodes are distributed on a 3-D cylinder of 20 cm
diameter and bottom diodes on the cylinder are ~20 cm farther away from the radiation source
than diodes on the top. Dose rate at the top will be 1.5 times higher than that at the bottom. (2)
Rotational radiotherapy techniques such as RapidARC and VMAT could be delivered with
variable dose rate during rotation by changing pulse rate. In this study, beam of 100 MU and 8x8
cm² field size was delivered with pulse rate changing from 50 MU/min to 400 MU/min from
Elekta console to investigate diode pulse rate dependence. To evaluate diode dose rate
dependence, dose rate was changed by varying source to detector distance (SDD) from 70 cm to
130 cm as in Figure 6-1 (b). A farmer Ion Chamber was used as reference in both cases with 5
cm water equivalent build up. Response of the diode on the top of the cylinder was compared to
the farmer ion chamber to evaluate its pulse rate and dose rate dependence.

Clinical Application

The purpose of radiotherapy QA is to verify that the delivered dose distribution agrees
with prescribed dose distribution. Clinical application of the 4-D diode array follows the same
principles as in the application of 2-D detector arrays and software38. In this study, three head
and neck IMRT plans (totaling 17 beams) and one prostate conformal arc plan were verified with
the diode array on an Elekta Precise™ machine capable of volumetric modulated arc therapy.

For step and shoot IMRT, treatment plans were designed with Pinnacle³ and transferred
onto the CT images of the diode array for dose calculation with nominal beam angles. A 2x2x2
mm³ dose grid was used in dose calculation. Three dimensional dose distribution for each IMRT
beam was exported from Pinnacle³ using Radiation Therapy Oncology Group (RTOG) protocol.
All IMRT beams were delivered with planned beam angles and measured with the diode array.
Instant gantry angles were derived from detector signals using gantry angle derivation algorithms developed in Chapter 5. Diode sensitivity correction factors and corresponding angular correction factors were applied to diode raw reading. The reference diode was irradiated with an open beam of 8x8 cm² field size, 100 MU to obtain an absolute dose conversion factor which would convert relative diode response into absolute dose. Calculated 3D dose distributions were compared to the corrected diode array measurement in 3D space using percent difference (%diff) and distance to agreement (DTA) for each individual beam. A pass/fail test was performed for each diode. In this study, a 5% threshold of the maximum dose was employed to filter out low dose value which could bias the pass/fail analysis. Criteria of 1%/2mm and 1%/3mm were both used in the analysis. Percent difference was calculated based on local dose value which is more stringent than based on maximum dose in the whole field. The DTA search was performed in 3-D space.

For the prostate conformal arc plan, gantry went clockwise from 270° to 90°. Pinnacle treats conformal arc plan as a collection of static beams (control points) with certain angle separation. Dose calculation is performed for each control point and final dose distribution will be interpolated from the super imposed dose distribution. A smaller angular separation between control points leads to smoother final dose distribution. In this work, angular separations of 2° and 5° between control points were used when creating the conformal arc plan which resulted in 90 and 36 total control points, respectively. In delivery and measurement with the diode array, real time gantry angle was derived to interpolate corresponding angular correction factors to apply on raw diode reading. Relative diode response was converted into absolute dose in the same way as mentioned in IMRT measurement. Measured and calculated composite dose distributions were compared using %diff and DTA.
In Chapter 5, it was found all diodes exhibit similar patterns in directional response. The directional correction factors vary by less than 2% on average. The feasibility to use the averaged directional response correction factors for all diodes was investigated. The raw measurement for IMRT beams was corrected by individual directional correction factors as well as averaged directional correction factors. Both corrected measurements were compared with TPS calculation to examine the pass rate change. For a comparison, measurement without directional correction factors was also compared to TPS calculation.

**Sensitivity to errors**

To evaluate the diode array sensitivity to gantry rotation errors, three IMRT beams (with beam angles 20°, 70° and 290°) were selected to deliver without gantry rotation errors and with ±1° and ±2° errors in gantry rotation which resulted in five measurements for each beam. Positive and negative errors in gantry rotation cause gantry to over rotate and under rotate, respectively. These measurements were compared to TPS calculation and the percentage change of diodes passing the 1%/1mm and 1%/2mm criterion was used to evaluate the diode array sensitivity to minor errors in gantry rotation.

Three IMRT plans were selected to estimate diode array sensitivity to random as well as systematic MLC position errors. Wrong leaf bank calibration is among the causes of systematic MLC errors\textsuperscript{111}. Systematic errors were simulated by modifying both leaf banks by ±1 mm or ±2 mm as in Chapter 2. Positive and negative errors caused MLC apertures to contract and expand, respectively. A random leaf positioning error sampled from [0mm, ±1mm, ±2mm] was applied to change each leaf position to simulate random MLC position errors. IMRT plans were exported from Pinnacle using Pinnacle scripting language. Corresponding errors were applied to each leaf while avoiding leaf collisions. Modified plans were then imported back for delivery. Together
with the original error free plan, five variations (one with random error and four with systematic errors) were delivered and resulted in six measurements. These measurements were compared to the TPS calculation with error free plan to evaluate diode array sensitivity to MLC position errors.

**Results**

The short term reproducibility was evaluated by repeating the same measurement 20 times over 2 hour period. The relative standard deviation of 10 randomly selected diodes is 0.7% on average with a maximum of 0.9%. This result includes the machine output variation during the course of measurement. Machine output fluctuation is found to be 0.3% with a monitoring ion chamber. Short term diode variation is estimated to be ~0.6% after subtracting machine output fluctuation in quadratic form.

Diode array is found to have excellent linearity with a linear regression coefficient of 0.9997. Linear diode response is shown in Figure 6-2.

Figure 6-3(a) shows diode pulse rate dependence of the diode array. A nearly linear decrease is noticed in diode sensitivity with decreased pulse rate. Relative diode sensitivity decreases by 1.7% when pulse rate decreases from 400 MU/min to 50 MU/min. In comparison, the Farmer ion chamber shows a decrease of less than 0.5%. This result agrees with the finding of Létourneau *et al.* In their work, the behavior of 2-D diode array (MapCHECK™, Sun Nuclear, Melbourne, FL) was evaluated which uses the same type of diode according to the manufacturer. Diode dose rate dependence is shown in Figure 6-3(b). SDD was changed from 70 cm to 130 cm to change dose rate. A Farmer ion chamber was used to measure dose rate and normalized at 100 cm SDD. A nominal repetition rate of 400 MU/min was used during measurement. Dose rate was estimated to change from 2.4 Gy/min to 7.8 Gy/min when SSD changed from 130 cm to 70 cm. Within this dose rate range, diode sensitivity relative to response
of Farmer ion chamber varies from -0.5% to 1.1%. This result agrees with the findings of other works where the behavior of same type of diode was evaluated\textsuperscript{38,88}. A slight diode over response could be observed with increasing dose rate\textsuperscript{66,88}.

For the measurement of the 17 IMRT beams, 65 out of 124 diodes are above the 5% threshold on average. Two beams with beam angle 135° and 220° which crossed table top before entering detector have significantly lower average passing rates than the other 15 beams. The average passing rates with one standard deviation for the other 15 beams are 94.3±2.1% and 99.6±0.8% with 1%/2mm and 1%/3mm criteria respectively. While the average passing rates for the two beams crossing table top before the diode array are only 82.2% (1%/2mm) and 88.8% (1%/3mm). Further experiment and analysis were conducted to investigate the cause of lower passing rate for these two beams. Both the detector and beam incident direction were rotated 180° to avoid table top and to irradiate the same part of the diode array. The average passing rates for the two beams jumped to 95.9% with 1%/2mm and 98.9% with 1%/3mm. It could be concluded that table top attenuation was the cause. It was also noticed a 3.5% boost in MU would increase the passing rate to 93.0% with 1%/2mm and 98.2% with 1%/3mm when the couch was in the beams’ way.

Figure 6-4 (a~d) shows diode measurements on all 4 rings for a typical IMRT beam with beam angle 290°. Also shown is the TPS calculation at corresponding positions. Dose profiles are unfolded for displaying purpose with origin set at the top of the cylinder. This particular beam enters the detector at arc distance of 12 cm and exits at -18 cm. Two peaks (entrance and exit) could be clearly observed from measured dose profiles on all 4 rings. Diodes passing the analysis criteria (1%/2mm) are shown as plus signs while “cold” and “hot” diodes are shown as stars and diamonds respectively.
Table 6-1 shows the passing rate changes when using different directional correction factors. The average passing rates when using individual directional correction factors are 93.3% and 99.1% with 1%/2mm and 1%/3mm, respectively. The use of average directional correction factors caused negligible drop (1%) with both 1%/2mm and 1%/3mm. However, when no directional correction factors were applied, drops of average passing rate were noticed to be 13% with 1%/2mm and 9% with 1%/3mm.

Results for the prostate conformal arc plan are shown in Figure 6-5. For the diode array measurement, all 124 diode measurements are above the 5% threshold with 120 diodes passing the 1%/2mm criterion, giving a 97% passing rate. When using 1%/3mm criterion, the passing rate increases to 100%. The plan is calculated with both 36 control points (solid line) and 90 control points (dotted line). Dose profile calculated with 90 control points is much smoother than the other one. However, when compared to diode array measurement, they result in same agreement.

Figure 6-6 shows average passing rate change with gantry rotation error using both 1%/1mm and 1%/2mm. The standard deviation is shown as error bar. When using 1%/2mm, average passing rate drops from 95.9% (no error) to 88.8% (+1° gantry rotation error), 86.6% (-1°), 75.6% (+2°) and 70.6% (-2°), respectively. With tighter 1%/1mm, average passing rate decreases from 69.4% (no error) to 60.7% (+1°), 46.7% (-1°), 37.6% (+2°) and 33.6% (-2°). A slightly asymmetric behavior is noticed with larger passing rate drop for negative gantry rotation errors.

Sensitivity of the detector to MLC position errors is shown in Figure 6-7. When using 1%/2mm, average passing rate stays around 94.6% and shows negligible change with random MLC errors or +1 mm systematic errors. Around 9% decrease in passing rate is observed with
both -1 mm and +2 mm errors. Largest drop (15%) occurs with -2 mm error. The detector shows higher sensitivity with a tighter 1%/1mm criterion. The passing rate drops by more than 20% with ±2 mm error, ~6% with random error or +1 mm error and ~13% with -1 mm error. The detector is highly sensitive to systematic errors of 2 mm with either 1%/2mm or 1%/1mm. However, a tighter criterion is favorable to detect errors of 1 mm.

**Discussion**

In order to be useful as a radiation dosimeter, the diode array must possess several desirable characteristics. In this chapter, the diode array is characterized by its reproducibility, linearity, pulse rate dependence and dose rate dependence. Its application for IMRT and VMAT QA is also demonstrated. The diode array was found to have excellent linearity with delivered dose. A linear regression coefficient of 0.9997 was obtained for the MU range of 1 to 1000. The diode array uses a proprietary designed ASIC chip which empties the capacitor every 50 ms. The diode won’t saturate with higher MU. Excellent linearity can be expected from the diode array with MU higher than 1000 though it is beyond typical clinical range. The relative standard deviation during reproducibility study was found to be 0.6%. The same uncertainty has been reported for its counterpart, MapCHECK, which uses the same type of diodes\(^{39}\). Since the diode array uses N-type diodes, its dependence on instant pulse rate and dose rate is a concern. Figure 6-3 (a) shows 1.7% variation when instant pulse rate changes from 50 MU/min to 400 MU/min and 1.6% variation when dose rate changes from 2.4 Gy/Min to 7.8 Gy/Min. This variation is in agreement with the same type of diodes reported from MapCHECK\(^{38,39}\). The clinical effect of diode dose rate dependence on rotational delivery needs to be further evaluated.

For IMRT beams, the measurement after correcting for diode sensitivity and directional response dependence agrees with Pinnacle calculation pretty well as can be seen from Figure 6-
4. An entrance peak and an exit peak can be clearly identified. The exit peak is much shorter than the entrance peak due to the attenuation of the diode array itself as well as the plastic insert. For the measurement in Figure 6-4, several cold spots (measurement lower than calculation) could be observed on the exit peak. This might be explained by the results shown in Figure 6-3 (b) that diodes under response at lower dose rate. The dose rate at entrance peak is about 1.5 times higher than that at the bottom according to inverse square law. The pass/fail analysis was conducted in 3D space which is different from QA analysis based on 2D dosimeter. The 2D dosimeter samples a plane perpendicular to beam incident direction. Dose change along beam incident direction would be very slow. High gradient information would be only contained in the sampled 2D plane. However, the 4D diode array samples a 3D sub space. The distance to agreement has to be searched in all directions to capture gradient information. This can be seen from Figure 6-4 (d). No agreement for the hot spot (failed 1%/2mm) would be found in the displayed plane. However, when using 1%/3mm as the criterion, agreement would be found in the direction perpendicular to the displayed plane. It was also found the passing rate was dominated by the DTA criteria while percent difference had negligible effect. The diodes passing 1%/2mm and 3%/2mm were almost the same for the majority of the IMRT beams. Thus the positioning accuracy is extremely important for accurate measurement. It is conjectured the internal dose can be accurately reconstructed when entrance and exit dose could be measured accurately with the diode array. The diode array provides flexibility to accommodate an ion chamber inside for verification purpose.

The diode array is capable of both composite and individual beam dosimetry. Several researchers have shown composite dosimetry tends to smear out the difference between measurement and calculation in individual dosimetry. Excellent agreement has been achieved
with individual dosimetry between the diode array measurement and TPS calculation. The agreement is even better for the prostate conformal arc plan when comparison is conducted on composite dose distribution.

When comparing measurement to calculation, the full 3D calculated dose grid could be interpolated from 2x2x2 mm³ to 1x1x1 mm³ resolution in advance. However, it was found the system became out of memory quickly in this way. In this work, a sub dose grid of 12x12x12 cm³ was extracted on the fly for each diode. Interpolation and comparison were conducted within the sub dose grid. We had no problem with computer memory and average time to conduct the comparison for each IMRT field was only around 10 seconds.

Results in Table 6-1 show the feasibility to use average directional correction factors for each individual diode. This is in agreement with the data in Figure 5-10 (b) where directional correction factors change within 1.8% from diode to diode and the factor that the passing rates are dominated by DTA. However, due to the large variation of directional correction factors with beam incident direction (largest change of 13%), the directional response dependence has to be taken into account.

It was found the couch has significant influence on the agreement between measurement and calculation. The discrepancy is largely due to the couch attenuation. QA with 2D dosimeters would not reveal the problem since radiation beams always pass the couch after the dosimeter. To solve this problem, the couch has to be taken into account by the planning system for planning as well as QA purpose\textsuperscript{112, 113}.

In contrast to per beam analysis in fixed gantry IMRT delivery QA, a composite dose distribution is calculated and compared to measurement for VMAT QA. This is due to the way Pinnacle calculates the dose distribution. It samples a static beam every a few degrees (e.g., 2° or
5° in this work) which results in a large number of control points (36 and 90 control points for 2° and 5° spacing, respectively). The final dose distribution is interpolated from the superposition of the dose distribution calculated from each static beam. It is impractical to perform individual dose comparison for each beam due to (1) the large number of static beams (2) the difficulty in correlating dose calculation for each control point and diode array measurement. As pointed out before, composite dosimetry can help detect errors in the comprehensive system but not identify sources of errors. How to balance the necessity of individual beam dosimetry and its associated computation time and memory space would be a future topic.

The diode array shows potential to catch both minor gantry rotation errors and MLC position errors with results in Figure 6-5 and Figure 6-6. In both cases, tighter criterion shows stronger sensitivity. Gantry rotation error of 2° would be easily captured with average passing rate drop of more than 20% and 30% with 1%/2mm and 1%/3mm, respectively. However, for 1° gantry rotation error, it is easier to make conclusive decision using 1%/1mm, with which the passing rate drops about 10%~20%, while it only drops 5%~7% with 1%/2mm. Similarly, the diode array shows stronger sensitivity to MLC position errors of 2 mm than 1 mm. A tighter criterion should be always used to increase the ability to detect minor errors in MLC position.17

Conclusions

A 4D diode array was characterized and its clinical application for IMRT QA has been demonstrated in this work. The 4D diode array was found to have excellent response linearity and reproducibility. Instant pulse rate and dose rate dependence of the diodes were quantified. Maximum variation was found to be around 1.7% at extreme conditions. The 4D diode array was used to measure 3 head and neck IMRT plans and a prostate conformal arc plan. The measurement was corrected using correction factors obtained in Chapter 5. Algorithms have been developed for the comparison of the measurement and TPS calculation in 3D space.
Excellent agreement was observed between diode array measurement and TPS calculation (99.6% passing rate with 1%/3mm criterion) which shows the diode array has been correctly calibrated and the TPS (Pinnacle 8.0 m) could adequately model the diode array and perform dose calculation. The sensitivity of this new dosimeter to gantry rotation errors and MLC position errors was evaluated. It was found the diode array could detect gantry errors of 2° and MLC position errors of 2 mm. The 4D diode array has been proven to be an effective tool for IMRT QA with planned beam angle and it has great potential for rotational delivery QA.
Table 6-1. Effect of using averaged directional correction factors on passing rate changes for three IMRT beams delivered with planned beam angles. For comparison, passing rates of the measurement without directional correction factors was also shown in the table.

<table>
<thead>
<tr>
<th></th>
<th>Individual correction</th>
<th>Average correction</th>
<th>No correction</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>1%/2mm</td>
<td>1%/3mm</td>
<td>1%/2mm</td>
</tr>
<tr>
<td>G020</td>
<td>91.8%</td>
<td>97.3%</td>
<td>90.4%</td>
</tr>
<tr>
<td>G070</td>
<td>95.6%</td>
<td>100%</td>
<td>93.3%</td>
</tr>
<tr>
<td>G290</td>
<td>92.5%</td>
<td>100%</td>
<td>92.5%</td>
</tr>
<tr>
<td>average</td>
<td>93.3%</td>
<td>99.1%</td>
<td>92.1%</td>
</tr>
</tbody>
</table>
Figure 6-1. (a) Photon beams are created by electron pulses. Radiation pulse rate is defined as $1/dt$, where $dt$ is the gap between continuous electron pulses. Pulse rate could be changed from Elekta console directly. (b) Setup to change dose rate at the position of the diode on the top by varying source to detector distance (SDD) from 70 cm to 130 cm.
Figure 6-2. Linear detector response as a function of delivered monitor units (MU) ranging from 1 to 1000. A beam of 8x8 cm² field size was delivered with center of the diode array aligned with machine isocenter in this study. A linear regression coefficient of 0.9997 is obtained in this study.
Figure 6-3. (a) Relative detector response as a function of pulse rate for diode and a Farmer ion chamber. Pulse rate was changed by changing the repetition rate from Elekta Precise console. The response is normalized at repetition rate of 400 MU/min. (b) Relative diode response as a function of dose per pulse. Dose rate was changed by varying source to detector distance (SDD) from 70 cm to 130 cm. Dose rate was recorded using a Farmer ion chamber and normalized at SDD = 100 cm and estimated to change from 2.4 Gy/min to 7.8 Gy/min approximately. Diode response was normalized at SDD=100 cm.
(a)

(b)
Figure 6-4. Dose profile calculated by Pinnacle (8.0m) and measured by all 4 rings of diodes on the diode array. Dose profiles were unfolded with origin at the top of the cylinder. Diodes passing 1%/2mm criterion are shown as plus signs. Diodes with measurement under and above calculation are shown as stars and diamonds respectively.
(a) TPS (90 CP)  
TPS (36 CP)  
ArcCHECK

(b) TPS (90 CP)  
TPS (36 CP)  
ArcCHECK
Figure 6-5. Dose profile calculated by Pinnacle (8.0m) and measured by all 4 rings of diodes for a conformal arc plan going clockwise from 270° to 90°. The plan was calculated with angular separations of 5° (TPS 36 CP) and 2° (TPS 90 CP) between control points.
Figure 6-6. Average passing rate change of 3 IMRT beams with gantry rotation errors. Nominal beam angles of the three IMRT beams are 20°, 70° and 290°. Positive and negative rotation errors cause gantry over rotate and under rotate, respectively. Error bars show 1 standard deviation.
Figure 6-7. Average passing rate change of three IMRT beams with random and systematic MLC errors of ±1 mm and ±2 mm. Positive and negative MLC errors cause beam aperture (both leaf banks) to expand and contract respectively. For random errors ("R"), a random error was sampled from [0, ±1mm, ±2mm] and applied to each leaf.
CHAPTER 7
CONCLUSIONS

The goal of patient specific IMRT QA is to detect potential sources of errors in the planning and delivery of IMRT treatment plans. The focus of this work is to evaluate and improve the current practice for patient specific IMRT QA. Existing patient specific IMRT QA approaches evaluate the agreement between dose distributions measured in a surrogate phantom with planar dosimeters and calculated in the same phantom by the treatment planning system. Discrepancy beyond a pre-specified tolerance level will be used as an indicator of potential errors in either planning stage or delivery stage. It is expected the current patient specific IMRT QA procedure could catch both gross errors and subtle errors.

The sensitivity of current typical patient-specific IMRT QA procedures to systematic MLC positioning errors on the order of 1 mm and 2 mm is evaluated in Chapter 2. Results show these MLC positioning errors can’t be detected with the widely adopted 3%/3mm criterion. When using 3%/3mm, the average passing rate of 53 IMRT fields with artificially introduced systematic MLC positioning errors of 2 mm stays around 90% for both EBT films and MapCHECK as shown in Figure 2-2 and Figure 2-3, respectively. Figure 2-5 and Figure 2-6 indicate, only with a tighter 2%/2mm criterion, 2 mm systematic MLC positioning errors would cause significant drop in average passing rate. This study shows that a tighter criterion is desirable for IMRT QA to achieve reasonable sensitivity. However, the tighter 2%/2mm criterion would result in low average passing rate even when MLC is free of systematic positioning errors due to uncertainty of other components in the system. As a consequence, large amount of false alarms would be created. This is supported by the survey conducted by Nelms et al\textsuperscript{11} that only few centers are adopting the 2%/2mm criterion. This study shows gross errors would be captured
with the current IMRT QA procedures, however, systematic MLC positioning errors of 2 mm which has significant dosimetric impact would likely be missed.

Study in Chapter 3 reveals large discrepancy between measured and calculated dose distribution of MLC error free IMRT plans is caused by the detector size effect which occurs at the very beginning of IMRT chain, i.e., the TPS commissioning stage. The finite size of the commonly used detectors (2 mm or 3 mm radius) in collecting beam commissioning profiles deteriorates the high gradient nature of IMRT beams. This phenomenon has been recognized for a long time. However, its impact on IMRT implementation has not been documented. A practical and reliable solution is needed to remove or minimize the detector size effect. Chapter 3 studies the clinical impact of detector size effect on IMRT via an analytical fitting/deconvolution approach. Photon beam profiles collected with finite size detectors are deconvolved to remove detector size effect to obtain a set of “true” photon beam profiles free of detector size effect. Beam profiles with and without detector size effect are employed to commission Pinnacle to obtain different beam models. Dose calculations with both beam models are compared with measurement performed with planar diode arrays. It is noticed “true” beam models (commissioned with “true” beam profiles) significantly increases the average passing rate (15% with 2%/2mm). The average passing rate of 53 IMRT beams with the “true” beam models is ~97% with 2%/2mm. The clinical significance of this study is (1) TPS could calculate dose distribution more accurately with the “true” beam models which is more important for IMRT beams characterized by many small high gradient beam segments (2) A tighter criterion (2%/2mm) could be employed due to the increased agreement between TPS dose prediction and dose measurement which leads to higher sensitivity of IMRT QA to subtle MLC positioning errors according to the sensitivity study in Chapter 2.
Current patient specific IMRT QA procedure is being blamed for its low efficiency. Each IMRT beam needs to be delivered and measured usually after clinic hours. Average time spent on IMRT QA with online detectors for one patient is about 30 min from detector setup to final data analysis. With the improved mechanic accuracy, few errors are caught and reported during patient specific IMRT QA. It is argued that physicists’ valuable time should be spent in other activities such as tumor contouring, patient setup and motion management which leads to large dosimetric errors and could not be caught by time consuming patient specific IMRT QA.

Another argument against individual patient IMRT QA is that MLC positioning errors are machine specific instead of patient specific. MLC positioning errors should be caught during dedicated MLC QA. The solution proposed in this work (Chapter 4) in improving the efficiency of current IMRT QA is to combine an independent planar dose calculation with a periodic MLC QA. Instead of delivering and measuring each IMRT beam, dose distribution is calculated with an independent planar dose calculation algorithm. The algorithm is commissioned with photon beam profiles collected from individual machines. Detector size effect is minimized with the deconvolution algorithm proposed in Chapter 3. Accuracy of the algorithm is validated with the 2D diode array measurement (gold standard). TPS dose calculation is compared to planar dose distribution calculated with this independent algorithm. The algorithm utilizes plan parameters from the record and verify system which controls the delivery system, thus errors generated during plan transfer will be caught. Delivery system accuracy will be checked via a periodic MLC QA procedure. Individual measurement for IMRT QA could be replaced by the efficient (10 seconds per beam) independent planar dose calculation algorithm. However, when seeing discrepancy between TPS dose calculation and the independent dose calculation, the plan should be delivered and measured as a gold standard for judgment.
Planned IMRT beams are delivered during IMRT QA but not with planned beam angle due to the limit of current 2D planar dosimeters. The consequence is that errors related to dynamic gantry or dynamic MLC could remain undetected. Current practice of IMRT QA is acceptable for fixed gantry IMRT. However, significant change has to be made for dynamic gantry rotational therapy. In this work, a novel 4D diode array provided by Sun Nuclear Corp. is calibrated in Chapter 5 and its clinical application for IMRT QA is demonstrated in Chapter 6.

The calibration of the 4D diode array is based on the concept of a coherent dosimeter, i.e., isotropic response. Diodes are isotropically arranged in a cylindrical phantom, presenting a consistent detector image to beams eye view at any beam angle. The calibration must eliminate any non-isotropic factors caused by diode response. Based on this basic idea, two sets of calibration factors are obtained: (1) diode sensitivity correction factors, to remove diode response difference under same beam conditions and (2) diode directional response correction factors, to remove diode angular response difference. The calibration factors are validated with water tank scan and found to be very accurate. Same beams are irradiated with different gantry angles and excellent agreement among diode array responses are observed at each gantry angle, which shows the calibration has achieved isotropicity. Instant gantry angle during rotational therapy has to be determined to apply the corresponding angular correction factors. A real time algorithm to derive gantry angle based on diode array signal is developed for this purpose. The algorithm tries to identify beam edges on both sides of the field to determine their intersection as the radiation source. The accuracy of the algorithm is found to be within 1° which is adequate to interpolate the corresponding angular correction factors.

Chapter 6 focuses on the characterization of the 4D diode array and its clinical application for both IMRT and VMAT QA. The diode array is found to have excellent reproducibility and
linearity. Its dependency on pulse repetition rate and diode rate is within ±1.5%. For IMRT QA, 3D dose calculation is performed in the CT scan of the diode array and exported using RTOG format. The same beams are delivered with planned beam angles and measured with the 4D diode array. Calibration factors obtained in Chapter 5 are applied to correct diode sensitivity and angular response dependence. The calculated and measured dose distributions are compared with percent difference and distance to agreement criteria in 3D space. Excellent agreement is found between Pinnacle calculation and diode array measurement, which shows Pinnacle can adequately model the diode array and the diode array has been properly calibrated. The sensitivity of this device to gantry angle errors and MLC positioning errors is also evaluated. Its clinical application for VMAT QA is demonstrated with a prostate conformal arc plan. Diode array measurement is compared to Pinnacle calculated dose distribution in 3D space. Excellent agreement is found between measurement and calculation. This study proves ArcCHCEK is an efficient and valuable tool for both IMRT and VMAT QA.
LIST OF REFERENCES


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BIOGRAPHICAL SKETCH

Guanghua Yan was born in Xiantao, China. He received his bachelor’s degree in computer science in July 1998 from Huazhong University of Science and Technology and his master’s degree in computer science in July 2001 from the same university.

In August 2002, he enrolled in the Department of Computer and Information Science and Engineering at the University of Florida. He joined the Department of Nuclear and Radiological Engineering at the same university for a combined degree program majoring in medical physics in January 2006. He received master’s degree in computer engineering in December 2007, master’s degree in nuclear engineering sciences in May 2008 and the degree of Doctor of Philosophy in nuclear engineering sciences in May 2009. His research in medical physics was conducted under the guidance of Drs. Jonathan Li and Chihray Liu.

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