Commissioning and implementation of an EPID based IMRT system Dosimetry Check for 3D absolute dose measurements and quantitative comparisons to MapCheck

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“Dosimetry Check” for 3D Absolute Dose Measurements and Quantitative
Comparisons to MapCheck

by
Jalpa A. Patel

Submitted to the Graduate Faculty as partial fulfillment of the requirements
for the Masters of Biomedical Science Degree in Medical Physics

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The University Of Toledo

December 2010
An Abstract of

Commissioning and Implementation of an EPID Based IMRT QA System “Dosimetry Check” for 3D Absolute Dose Measurements and Quantitative Comparisons to MapCheck

by

Jalpa A. Patel

As partial fulfillment of the requirements for the Masters of Biomedical Science Degree in Medical Physics

The University of Toledo

July 2010

The software package “Dosimetry Check” by MathResolutions, LLC, provides an absolute 3D volumetric dose measurement for IMRT QA using the existing Electronic Portal Imaging Device (EPID) mounted on most linear accelerators. This package provides a feedback loop using the patient’s treatment planning CT data as the phantom for dose reconstruction. The aim of this work is to study the difference between point, planar and volumetric doses with MapCheck and Dosimetry Check via the use of the EPID and the diode array respectively. Evaluating tools such as point doses at isocenter, 1-D profiles, gamma volume histograms, and dose volume histograms are used for IMRT dose comparison in three types of cases: head and neck, prostate, and lung. Dosimetry Check can be a valuable tool for IMRT QA as it uses patient specific attenuation corrections and the superiority of the EPID as compared to the MapCheck diode array. This helps reduce the uncertainty in dose for less variability in delivery and a more realistic measured vs computed dose verification system as compared to MapCheck.
To my loving parents, brother, and husband, thank for your sacrifices in helping me achieve my dreams. I love you all very much.

To my Professors, thank you for your guidance and support.
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Chapter 1

Introduction

Intensity Modulated Radiation Therapy (IMRT) is performed at hospitals and institutions around the world in treatment of cancer patients. The advantage of IMRT is that it allows delivery of dose to the tumor while allowing larger sparing of normal tissue, which in effect can allow delivery of higher doses to the tumor. Intensity modulation relies upon beamlets within a given field size (usually beam’s eye view of the target plus a margin) to deliver the radiation dose to the target at different gantry angles to provide the necessary desirable dose distribution. These beamlets are shaped by using Multileaf Collimators (MLCs) on board newer linear accelerators. In dynamic IMRT, these metal leaves adjusts the size and shape as they move across the irradiated tissue while the beam is on, blocking out some areas and filtering others to vary the beam intensity and achieving a desirable distribution of radiation dosage in the treated volume.

Since IMRT has become such an integral part of treatment delivery, this has created the demand to produce an adequate IMRT quality assurance (QA). IMRT is usually done through inverse planning, which creates large dose gradients and any delivery errors can have a severe effect on the outcome of the treatment. The difficulty in producing a QA system that accurately predicts the dose lies in the scatter contribution...
generated within the patient; not mentioning when wedges, compensating filters, and MLCs are used. Especially as the MLCs move in and out of the field, the amount of scatter produced is difficult to model, and so when predicting dose distribution in a patient to be treated, the amount of scatter contribution from the machine as well as from the patient to the treated volume is unknown.\(^9\)

Other factors that affect dosimetric uncertainties in IMRT QA can be due to machine errors. Small field sizes and the delivery of small monitor units, where the accuracy in beam output and profile may be lacking, can cause dosimetric differences.\(^1\) The delay and geometric differences in segmental dose using MLCs can also lead to discrepancies between computed and delivered dose.\(^1\)

A software program Dosimetry Check by MathResolutions LLC, Columbia MD provides an absolute 3D volumetric IMRT QA system using an Electronic Portal Imaging Device (EPID).\(^{13}\) A major advantage for this system over the existing IMRT QA packages is that Dosimetry Check does not require purchase of additional detector hardware, which is often inferior to most EPID units in resolution, contrast, and the convenience of being a part of accelerator system. In addition, Dosimetry Check uses patient specific CT data for dose superimposition and direct comparison with treatment planning dosimetry in 3D space. Dosimetry Check provides a method to reconstruct the 3D measured dose on the same patient’s CT as used for treatment planning.\(^1\) The key to this program is that it uses measured IMRT beam output with the EPID which includes all machine variations to reconstruct the absolute dose on a patient’s CT.\(^1\)
Chapter 2

Literature Survey

Traditionally, 3D conformal radiation therapy (3DCRT) is used to deliver a uniform intensity across an irradiated field. Wedges, transmission blocks, and compensators are also used to modify the beam intensity to compensate for missing tissue, and other irregularity in patient anatomy. Now, IMRT is typically performed by MLCs and by a process called “inverse planning”. This requires creation of subfields to paint the radiation dose into the target volume and avoid delivery to critical structures. The process of IMRT and inverse planning is too complicated to be planned by conventional forward planning techniques. With highly complex 3DCRT plans forward planning becomes increasingly difficult. It demands considerable experience on the part of the planner and planning time and yet the full potential of 3DCRT may not be realized even with the most experienced planner. Inverse planning on the other hand, has a distinct difference from conventional forward planning in that the optimization work is performed by computer algorithms that optimize according to the pre-set dose constraints of the targets and organs at risk (OARs). Of the many advantages sought in inverse planning, better plan optimization obtained with aid of complex mathematical algorithms, less dependence on human planning efforts, and reduction in planning time are important.
Currently, inverse planning is mainly employed in IMRT with a highly conformal radiotherapy technique achieved by carefully controlling the intensities of individual beamlets within a radiation beam. It relies upon the user to input criteria and beam directions, and by implementing an adaptive convolution/superposition algorithm achieving optimal fluence distributions for each beam.

Even though IMRT offers a major advantage in sparing the normal tissues and critical structures, 3DCRT has the advantages of a more straightforward dose delivery method, which does not require extra equipment and complicated verification procedures. Ever since the advent of IMRT and inverse planning, medical physicists have relied on comparison between predicted and computed fluence maps on planar axes after radiation delivery of all beamlets in a given gantry position. The Dosimetry Check (DC) software package is the first of its kind in measuring the dose distribution using the patient CT data in three-dimensional space and provides ability to better assess the predicted plan using the many tools provided for quantitative analysis.

### 2.1 IMRT QA

At the University of Toledo Medical Physics division, when using Pinnacle v8.0m treatment planning system (*Philips Radiation Oncology Systems, Milpitas CA*), the software first runs a pencil beam algorithm to produce an initial open density matrix or a fluence map and then uses the adaptive convolution/superposition algorithm for a final dose calculation to produce optimal fluence maps. Next, Direct Machine Parameter Optimization (DMPO) is used which uses the machines parameters, such as MLC leaf
positions and beam segmental weights, to re-create the fluence map by using multiple segments. Due to the complex nature of producing an IMRT plan and near impossible ability to perform a second check dose calculation by hand, thorough verification of plan delivery must be performed adequately enough to detect any variability from the treatment plan.

There has been many ways to perform IMRT QA including the use of MOSFETs, TLDs, radiochromic and radiographic films, diodes and ion chamber. Film dosimetry has been popular due to its good spatial resolution and weak energy dependence. Unfortunately, it also has inconsistencies inherent to film production and processing such as handling, processing, and digitizing each film.

Film dosimetry can be performed by placing a film such as Gafchromic film inside a cubic, cylindrical, or spiral phantom and exposed to IMRT fields. One of the drawbacks is each patient’s treatment fields must be copied over to a phantom and have the dose distribution recalculated, which makes it completely independent from the patient. One difficulty with this is the beams are delivered onto film, and so scatter within the patient is not included and the dose distribution within the patient is not accurately predicted. Once exposed, the film can be analyzed by using RIT software (Radiological Imaging Technology Inc, Colorado Springs CO). RIT specializes in IMRT film analysis tools and can compare predicted to measured doses. A major drawback of film is that it is limited to 2D dose information. Spiral phantoms can be used to yield 3D dose distributions but is spatially limited and only ensures the treatment delivery is correct and yields no information about what the patient actually receives outside the target volume.
Another popular IMRT method is using an ion chamber array such as MatriXX by Scanditronix Wellhofer GmbH, Germany. With a dynamic or step and shoot IMRT techniques, IMRT fields are changing with varying monitor units and field sizes, considerations should be given when using an ion chamber array in evaluating IMRT plans. Ion chambers are energy dependent, the size of collecting volume, charge leakage and material must be chosen appropriately. Again, each patient’s treatment fields must be copied over to a phantom failing to provide the actual dose distribution within the patient. Absolute dosimetry can be performed if absorbed dose-to-water calibration coefficients are available. According to Li et al, MatriXX, which has parallel plate chambers with sensitive volumes of 0.08cm$^2$, shows a less than 0.5% variation and “displays a volume average effect consistent with detector size”, which can be corrected for by “convolving the treatment planning calculation with a Gaussian function” before comparing measurements.

An IMRT QA system more commonly used at the University of Toledo, Radiation Oncology Department is MapCheck by Sun Nuclear Corp, Melbourne FL. This is a popular QA system due to its ease of immediate digital readout. This is a 2D diode array containing 445 n-type solid state detector diodes used to verify planar dose. It has a maximum field size of 22cm x 22cm with an active detector area of 0.64mm$^2$ and a diode spacing of 1cm axial and transverse making it ideal for high gradient dose areas. It uses 2D dose maps for each beam created from the treatment planning system (TPS) as a comparison. In the TPS, each beam is applied perpendicularly to a flat surfaced phantom and these planar dose maps are then transferred and loaded into the MapCheck software.
The analysis tools the program uses is percent dose difference, distance to agreement criteria, and the gamma index\textsuperscript{13}.

According to Li et al, the detector shows no field size or SSD dependence at 6MV and 18MV photon energies, with negligible errors < 1\% for more than 10 MUs\textsuperscript{7}. But MapCheck did exhibit consistent under-response when measuring fewer than 10 MUs\textsuperscript{7}. Unfortunately, this is crucial since IMRT beam segments can usually have monitor units less than 10MUs. Another major limitation to MapCheck is that it is limited to 2D planes on a solid phantom, and so the cumulative effects of the individual beam differences cannot be fully appreciated. Majority of existing IMRT QA tools provide a 2D result, this is the key difference when using Dosimetry Check. This package allows measurement of dose in three-dimensional space reconstructed on the patient’s CT providing cumulative effects of primary attenuation through inhomogeneity in the patient.

COMPASS (IBA Dosimetry GmbH, Germany) is also a 3D absolute dose measuring for IMRT QA system but it requires the purchase of hardware and software. The hardware has 1600 high resolution parallel plate ion chamber detectors with a maximum active area of 40x40cm\textsuperscript{2} at isocenter\textsuperscript{4}. It attaches to the gantry and measures the transmission output while the patient is being treated\textsuperscript{4}. COMPASS software then uses a forward calculation beam model and this measurement to reconstruct the 3D dose on the patient’s planning CT\textsuperscript{4}. There are drawbacks to this system such as the hardware being in the path of the beam while the phantom is being treated and second, it needs to use a source model to predict the dose in the patient. Whilst, Dosimetry Check does not need a source model and solely uses a measured source model to account for all variations in machine delivery\textsuperscript{1}. 
Chapter 3

Dosimetry Check

As previously stated, Dosimetry Check uses the patient’s CT data as the phantom allowing for dose reconstruction on the same CT data set as the treatment plan providing a feedback loop\textsuperscript{11}. The dose calculation technique of Dosimetry Check performs opposite to Pinnacle’s inverse planning; it takes the measured fluence maps, de-convolves a dose kernel to obtain a dose distribution\textsuperscript{6}. The difference is it uses a fast pencil beam algorithm for speed this lacks the accuracy of an adaptive convolution/superposition algorithm but can be acceptable for a second check.

Pencil beam algorithm is a correction based algorithm\textsuperscript{6}. It is semi-empirical and relies on measured data such as percent depth doses and beam profiles\textsuperscript{6}. The algorithm divides the area of the mono-directional beam into small pixels perpendicular to the beam, each pixel now being a separate pencil beam\textsuperscript{1}. The beam is not fully modeled here, but Dosimetry Check uses a Measured Source Model instead\textsuperscript{1}. By measuring the incident fluence through the EPID, it takes into account all variations in the beam such as the collimation system, flattening filter, wedges, interleaf leakage, etc.
The algorithm itself poses discrepancies between the calculated dose by Dosimetry Check and the optimized dose by Pinnacle v8.0m, and the user of Dosimetry Check needs to be aware of these limitations. These inaccuracies due to the use of a simple pencil beam algorithm are especially prominent when dealing with tissue inhomogeneity such as lung, oral and nasal cavities, teeth, and bone where electronic equilibrium breaks down\textsuperscript{14}. The pencil beam algorithm being semi-empirical accounts for initial photon interactions characterized by attenuation changes depending on the incident energy, tissue density and the effective atomic number of the tissue\textsuperscript{14}. Typically, this is taken into account by a simple density scaling by the “radiological –pathlength” thickness for each medium\textsuperscript{14}. However, it does not model secondary electron transport, and hence, overestimates the dose within the low density regions\textsuperscript{5}. This is a concern at higher energy photon beams since the secondary charged particles ranges are longer causing a decrease in lateral spreading of dose in low density tissue\textsuperscript{5}.

A study by du Plessis \textit{et al} in 2001 showed inaccuracies of 10-20\% in lung cases and 20-70\% in head and neck cases when using a pencil beam algorithm as compared to Monte-Carlo results\textsuperscript{14}. Also, high Z materials showed an approximate 10\% discrepancy in dose for a target behind prosthesis such as pelvic IMRT due to significant back scattering from the electron fluence and dose at the interface\textsuperscript{14}.

3.1 Data Acquisition

The major drawback to Dosimetry Check is commissioning of the software with beam data to model machine output. Other IMRT QA systems like MapCheck and RIT
do not need any beam data. Besides the initial commissioning, the convenience to Dosimetry Check is it only needs measurements with the EPID, though film or ion chambers can also be used as input. The EPID is used to measure an incident fluence distribution by accumulating frames individually for each control point in a beam. Dosimetry Check then combines the frames into one image for each gantry angle. Typically a flood field should be taken with each patient’s IMRT beams. This has been defined as a 26x26cm$^2$, 100MU open field which corrects the dose discrepancies in EPID images. The 26x26cm$^2$ is the size of the detector panel on the EPID. One of the major inaccuracies the flood field corrects is any pixel gain variation, since the EPID is very energy dependant and as the spectrum changes off axis, so does the pixel response$^1$. The effect of this correction removes the in-air off axis ratios making the beams flat, which is later restored by the deconvolution process$^1$.

A 10x10cm$^2$, 100MU open field is also taken along with each IMRT QA under the same conditions as the treatment fields. Dosimetry Check normalizes and centers all beam images to the 10x10cm$^2$ open field signal and assumes that the signal value is correct for the given monitor units$^8$. Each pixel on the images is mapped to the open 10x10cm$^2$ field size on the central axis that would produce the same signal. This takes into account the collimator scatter from different field sizes, wedge factors, and other modifiers of the pixel signal. Using this, Dosimetry Check can assign relative monitor units (RMU) that would be needed to produce the same pixel signal on the central axis of a 10x10cm$^2$ field$^1$.

One key calibration to Dosimetry Check stated above is calibrating the EPID images to a range of monitor units. This is done by shooting a 10x10cm$^2$ field size with
different monitor units. This gives a plot of integrated central axis pixel values versus monitor units and provides a means to relate integrated pixel values, which is the composite of EPID image frames, to the output of the linear accelerator\textsuperscript{11}. The resultant images are RMU calibrated and is used to achieve the in air fluence.

Since the algorithm is semi-empirical, beam data for our particular linear accelerator needs to be included. In-air off axis ratios, in-water diagonal profiles and percent-depth dose curves were used to model the output of the machine. A utility \textit{GenerateBeamParameters} is executed. In summary, this computes a poly-energetic pencil kernel from the measured data and a Monte Carlo based point spread function\textsuperscript{1}. It uses the central axis percent depth dose data to fit a spectrum of the beam which in turn forms the pencil kernel\textsuperscript{1}. Along with this, off-axis correction factors are generated which accounts for the changes in beam penetration off axis due to the change in beam energy off axis by using the in-water diagonal profiles\textsuperscript{8}. This is typically a function of radius and depth to produce off-axis angles. Appendix A contains the resultant text file.

After the spectrum is fitted and the pencil beam kernel generated, the beam kernel parameters can then be calculated.

\subsection*{3.2 Deconvolution Kernel}

Returning to the aim of trying to achieve the in air incidence fluence, another factor that must be taken into account is the internal scatter generated within the EPID device\textsuperscript{11}. Here, the EPID is acting as a phantom and so the phantom scatter of the EPID is removed from the images to produce in air fluence images which are only a factor of
the collimator scatter. The equation below illustrates this concept, taken from Mathresolutions.com.

\[ S_c = \frac{S_{p,c} (EPID)}{S_p (EPID)} \]  

(3.1)

The phantom scatter of the EPID is generated and corrected for by the deconvolution process. It is performing a deconvolution by taking the inverse of the point spread function of the device to arrive at in-air fluence in RMU. The point spread function describes the response of an imaging system for a point source. The captured image in actuality will not show up as a point on a single pixel but will be blurred and degraded. Thus taking the inverse corrects for this optical spreading in the deposition of the EPID.

The point spread function (PSF) kernel, given by equation 3.2, consists of two separate kernels: the dose kernel and the glare kernel. The dose kernel accounts for the “dose deposition in the EPID’s scintillator screen”. The scintillator screen of the Elekta EPID is a terbium doped gadolinium oxide layer, also called a gadox scintillator. The glare kernel accounts for the optical spreading from the scintillator screen to the amorphous silicon photodiode layer. The PSF kernel is represented as a composite of the two kernels by the sum of exponentials. This offers a method for deconvolution of EPID images to obtain incident fluence for dose reconstruction.

\[ k(r) = \sum_{i=1}^{n} a_i e^{-b_i r} \]  

(3.2)

Since this is a circularly symmetric function, \( r \) is the radius (cm) from the central axis at a source-axis distance 100cm, and \( a_i \) and \( b_i \) are fitted parameters. The parameters are fitted with three inputs: measured output factors in water, corresponding open field EPID images and the monitor units the images were taken with. The option here was selected
to multiply in-air OCRs. This applies to images where a flood field image is used to
correct for pixel variability which consequently removes the in-air OCRs. Dosimetry
Check fits the parameters using a default of 5 exponentials to minimize the variance
between the computed and the measured dose\textsuperscript{11}. Appendix A contains the program file
manually adjusted to reflect the measured output factors for the SL25 linear accelerator.

The deconvolution is performed in the frequency domain, and so the 2D fast
Fourier transform of the 2D spatial images are taken into the frequency domain and
multiplied by the inverse of the PSF for the EPID\textsuperscript{8}. In general, the deconvolution kernel
corrects for the field size response of the EPID and removes the effects of the EPID being
a phantom. As a side note, the deconvolution is a high pass spatial filter, since the point
spread function of the EPID is a low pass filter\textsuperscript{1}. Any sudden change in image will be
amplified as that comprises of high spatial frequencies and will be imaged as spikes; that
is why the area of the fluence images should be restricted, this also helps save on
computational time\textsuperscript{1}. This will be seen in the results section.

3.3 Dose Kernel

At this point, the images are RMU calibrated and deconvolved. The in-air fluence
images now need to be corrected by the in-air off axis ratios to restore non-flatness which
was previously removed by the flood field image\textsuperscript{11}. The in-air off axis profiles provides a
curve from the central axis out along the radius which is then applied to all pixels in the
EPID. It is measured along the diagonal for the largest field size and is assumed to
correct a symmetrical treatment field. Dosimetry Check is not able to simulate non-
symmetrical field since the kernels and off axis ratios are modeled to be circularly
symmetric. Once these in-air fluence images are achieved, the dose in water can be calculated by using the following equation, taken from Mathresolutions.com:

\[ S_c \times (\text{In-water } S_p) = \text{In water } S_{c,p} \]  

(3.3)

The in-water phantom scatter is computed with the pencil beam kernel and fitted to beam data by program GenerateBeamParameters\(^1\). Appendix A contains the in-water output factor table for various field sizes for SL25, 6MV. All necessary beam parameters and models are defined in order to calculate the dose in the patient’s anatomy.

### 3.4 Issues in Commissioning Dosimetry Check

Most issues encountered were in commissioning Dosimetry Check. The online user manual on Mathresolutions.com does not provide a clear concise method for commissioning Dosimetry Check. All measured data for the software was taken using the Wellhofer Scantronix program, the percent depth dose curves were in ASCII files. Dosimetry Check requires these files to be converted into a proprietary file format. A utility ConvertRFA300Files.exe is provided to convert files into this format through a command prompt.

Other required data includes a CT number to density conversion table, which was not as apparent. A default table is used which caused a scaling of 10-15% dose difference in 1-D beam profiles. A CT scan of a phantom with biological equivalent inserts should have been included to fit a polynomial curve. Instead, the same CT numbers and corresponding densities from the treatment planning system were inputted and a 3\(^{rd}\) order polynomial fit was generated. Once a deconvolution kernel was generated with all the beam data in place, water phantoms 50x50x50cm\(^3\) were used as verification. Different
field sizes were experimented with such as 10x10\text{cm}^2 and 20x20\text{cm}^2 with 25MUs and 100SSD.
Chapter 4

Dosimetry Check vs. MapCheck

Our aim is to study the difference between point, planar and volumetric doses with MapCheck and Dosimetry Check via the use of the EPID and the diode array. Evaluating tools such as point doses at isocenter, 1-D profiles, gamma volume histograms, and dose volume histograms are used for IMRT dose comparison in three types of cases: head and neck, prostate, and lung.

For a similar comparison, EPID DICOM images and absolute dose text files from the MapCheck were read into Dosimetry Check. Dosimetry Check deconvolves these images and reconstructs the dose onto the CT dataset to produce point and volumetric doses. Since MapCheck is only capable of planar dose comparisons, central axis profiles were compared to Dosimetry Check in a phantom for a similar comparison.
Chapter 5

Methods and Materials

5.1 Instrumentation

At the University of Toledo HSC, the linear accelerators used were the Elekta Precise Series SL15 and SL25. The machines consist of nominal energies 6MV and 10MV. The SL25 also has 18MV, but this is typically not used in IMRT due to neutron contamination. Dosimetry Check uses the EPID to obtain fluence images. The iViewGT flat panel imaging system, figure 5.1, is attached to the Elekta linear accelerators and uses a terbium-doped gadolinium oxide layer, also called a gadox scintillator along with an amorphous silicon detector\textsuperscript{12}. The pixel size of the EPID is 0.025x0.025cm\textsuperscript{2}.\textsuperscript{11}.
The treatment planning system used at this institution is Pinnacle v8.0m. Pinnacle uses adaptive convolution/superposition algorithm. The MapCheck diode array, figure 5.2, version 5.00, is used to measure the 2D planar dose maps.
5.2 Commissioning of Dosimetry Check: Data Acquisition

Dosimetry Check uses beam parameters to model the output of a linear accelerator. The process of commissioning Dosimetry Check begins with the calibration of the EPID signals to relative monitor units. A 10x10cm field size is used for exposures with multiple monitor units. At our department, monitor units of 2,5,15,25,50,75,100,200 were used. The linear calibration curve, figure 5.3, provides a means to convert EPID pixel signal to relative monitor units.

![Figure 5.3: Calibration curve for pixel signal value vs. relative monitor units for SL25, 6MV](image)

For the deconvolution kernel, output factors at different field sizes are needed along with EPID images at the different field sizes. This only needs to be done once per energy. The field sizes used were 2x2cm², 3x3cm², 4x4cm², 5x5cm², 6x6cm², 8x8cm², 10x10cm²,
12x12cm$^2$, 14x14cm$^2$, 16x16cm$^2$, 18x18cm$^2$, 20x20cm$^2$, and 25x25cm$^2$ with a source-to-
surface distance (SSD) setup. The IViewGT EPID is limited to a 25.6x25.6cm$^2$ detector
size$^{11}$. The images were taken with a consistent monitor unit of 25. An extra 10x10cm$^2$,
100MU field and a 26x26cm$^2$, 100MU was also taken. The output factors were measured
using a small water phantom using two ion chambers at depth of maximum dose. For the
6MV beam, 1.7cm is the $D_{\text{max}}$ (depth of maximum dose), for the 10MV beam, 2.1cm is
$D_{\text{max}}$. For smaller field sizes, an Extradin A-16 Micropoint chamber with a volume of
0.007cm$^3$ was used from 2x2cm$^2$ to 6x6cm$^2$ field sizes. A Scandatronix CC13 compact
chamber with volume 0.13cm$^3$ was used from field sizes 6x6cm$^2$ to 25x25cm$^2$. Cross
referencing both ion chambers, readings were calculated from both chambers and
averaged. The output factors, figure 5.4 & 5.5, are calculated by dividing the reading at a
particular field size to the reading at a 10x10cm$^2$ square field. These output factors along
with the corresponding field size image was used to calculate the parameters in the
deconvolution kernel.
Figure 5.4: Measured output factors at dmax for various field sizes for SL15, 6MV & 10MV

Figure 5.5: Measured output factors at dmax for various field sizes for SL25, 6MV & 10MV
Using the Wellhofer scanning system (Scanditronix-Wellhofer, Bartlett, TN), we measured in-air and in-water off axis ratios. The in-air off axis ratios were measured at isocenter of the linear accelerator for a 40x40cm$^2$ open field. A plastic build-up cap was used with the Scandatronix CC13 compact chamber. A thickness of 1.3cm for the 6MV and 1.8cm for 10MV was measured for the water equivalent plastic. The Wellhofer tank was rotated 45 degrees such that the largest diagonal length of at least 35cm was achieved. Scans were preformed horizontally at 0.4mm increments to beyond the edge of the penumbra. Only one direction was measured, since the beams are symmetric. The in-air off-axis ratios, figure 5.6 & 5.7, are calculated by dividing the reading along the diagonal length by the reading at the central axis.

![Graph](attachment:image.png)

Figure 5.6: Measured in-air OARs for an open 40x40cm$^2$ field size for SL15, 6MV & 10MV
The in-water diagonal profiles were measured at depth of maximum, 10cm and 20cm for a 40x40cm² open field size at a 100cm SSD. Again, measurements were done along from the central axis to the corner horizontally in order to encompass the penumbra. The Wellhofer tank was rotated 45 degrees such that the largest diagonal length of at least 35cm was achieved and scans were performed at 0.4mm increments. Figures 5.8-5.11 illustrate the scans.
Figure 5.8: Measured in-water OARs for an open 40x40cm² field size for SL15, 6MV at various depths.

Figure 5.9: Measured in-water OARs for an open 40x40cm² field size for SL15, 10MV at various depths.
Figure 5.10: Measured in-water OARs for an open 40x40cm² field size for SL25, 6MV at various depths

Figure 5.11: Measured in-water OARs for an open 40x40cm² field size for SL25, 10MV at various depths
5.3 Using Dosimetry Check and MapCheck

Once plans were optimized on Pinnacle v8.0m, a DicomRT export is used to export the plan to the appropriate computer. The beams were all shot onto the EPID using single frame capturing and manually inputting the number of IMRT segments for each beam. The 10x10cm\(^2\), 100MU centering field and the flood field must also be retrieved. The DicomRT export includes three files, the treatment plan (RP), the Regions of Interest (RS), and the dose grid (RD). Also a DICOM image export is done to export the CT data set. All exported files are then placed in the appropriate folder. In Dosimetry Check, the feature *Auto-Read case* is used and the RP file is selected. This reads in the CT data set including all appropriate Regions of Interests, (ROIs) Points of Interests (POIs), and the dose grid.

Subsequently, Dosimetry Check requires a body contour be defined, typically a body contour produced in the TPS is set in Dosimetry Check. Next, the EPID images are retrieved using the program *GetEPIDImages*. This scans the IViewGT directory and retrieves the latest jpeg images for each patient. Once the appropriate images are selected, the option *WriteDicomFiles* is selected. Here Dosimetry Check integrates the individual jpeg images for each beam and converts them into DICOM files. Next, the *ConvertElektaImages* program is run. Here the EPID images are chosen, corrected with a flood field and centered with the 10x10cm\(^2\), 100MU, and deconvolved resulting in RMU images. Under, *RunDosimetryCheckProgram*, evaluation tools are available to analyze the plan.
For MapCheck v.5.00, the device is setup to the plane of the detectors for a source-to-plane distance of 100cm resulting in a SSD of 98cm and 2cm of equivalent water thickness. The software uses 2D dose maps for each beam created from the TPS as a comparison. In the TPS, each beam is applied perpendicularly to a flat surfaced phantom and these planar dose maps are then transferred and loaded into the MapCheck software. Each beam was measured individually in “Absolute Dose” mode which then is saved in a text file. Again, a 10x10cm, 100MU field is shot onto the MapCheck which is used to calibrate and center the IMRT fields. The images are not corrected with a flood field image because there this minimal diode detector variation and so off-axis ratios are assumed to be part of the image. The program ConvertMapCheckImages reads in these absolute dose text files and converts them to RMU images with the deconvolution kernel. At this point, RunDosimetryCheck utility can be used to calculate the dose to the patient with the MapCheck images.
Chapter 6

Results

6.1 Deconvolution Kernel

Beginning with the deconvolution kernel, below is the resultant file of the deconvolution fit for 6MV, SL 25 linear accelerator:

```plaintext
/* File format version */ 3
/* file type: 104 = convolution kernel */ 104
/* Description: */ <**>
/* machine name: */ <*UToledoSL25*>  
/* for energy */ 6
/* variance for fit = */ 1.280735e-003
/* number of exponentials */ 5
6.907821e+001     2.288060e+001
6.682684e-002     2.274132e+000
4.975527e-003     6.159092e-001
9.900301e-005     5.189464e-002
8.770517e-008     5.847799e-003
/* OCR correction type:  
0: divide out in water OCR at dmax, then multiply in in air OCR after deconvolution  
1: multiply in in air OCR after deconvolution  
2: make no OCR correction. */ 1
/* file written 16-Mar-2010-11:01:19(hr:min:sec) */
/* After Kernel fit.  
Field size response of the deconvolution kernel:  
Field Size     Raw c.a.    After Deconvolution  Ratio  
cm          Signal      c.a. Signal                      
2.0 x  2.0    21.17       23.61                1.1154  
3.0 x  3.0    22.02       23.83                1.0818  
4.0 x  4.0    22.68       24.08                1.0618  
5.0 x  5.0    23.14       24.23                1.0473  
6.0 x  6.0    23.67       24.52                1.0359  
8.0 x  8.0    24.27       24.73                1.0190  
```
In water values.

<table>
<thead>
<tr>
<th>Field Size</th>
<th>Depth</th>
<th>Measured</th>
<th>Computed</th>
</tr>
</thead>
<tbody>
<tr>
<td>10.0 x 10.0</td>
<td>24.86</td>
<td>25.03</td>
<td>1.0067</td>
</tr>
<tr>
<td>12.0 x 12.0</td>
<td>25.33</td>
<td>25.24</td>
<td>0.9968</td>
</tr>
<tr>
<td>14.0 x 14.0</td>
<td>25.73</td>
<td>25.43</td>
<td>0.9884</td>
</tr>
<tr>
<td>16.0 x 16.0</td>
<td>26.10</td>
<td>25.59</td>
<td>0.9806</td>
</tr>
<tr>
<td>18.0 x 18.0</td>
<td>26.37</td>
<td>25.66</td>
<td>0.9728</td>
</tr>
<tr>
<td>20.0 x 20.0</td>
<td>26.72</td>
<td>25.80</td>
<td>0.9655</td>
</tr>
<tr>
<td>25.0 x 25.0</td>
<td>27.35</td>
<td>25.89</td>
<td>0.9465</td>
</tr>
</tbody>
</table>

Field Size      Depth     Measured    Computed
[318x39]2.0 x  2.0      1.70       22.30       22.25  -0.23%
[108x700]3.0 x  3.0      1.70       22.99       23.05  0.28%
[108x689]4.0 x  4.0      1.70       23.54       23.48  -0.25%
[108x678]5.0 x  5.0      1.70       23.77       23.74  -0.09%
[108x666]6.0 x  6.0      1.70       24.14       24.14  0.03%
[108x655]8.0 x  8.0      1.70       24.60       24.55  -0.19%
[108x644]10.0 x 10.0      1.70       25.00       25.01  0.04%
[108x633]12.0 x 12.0      1.70       25.35       25.39  0.15%
[108x622]14.0 x 14.0      1.70       25.67       25.68  0.04%
[108x611]16.0 x 16.0      1.70       25.93       25.97  0.14%
[108x600]18.0 x 18.0      1.70       26.17       26.14  -0.12%
[108x589]20.0 x 20.0      1.70       26.39       26.40  0.03%
[108x578]25.0 x 25.0      1.70       26.75       26.74  -0.04%

Figure 6.1: Text file of the deconvolution kernel fit using varying output factors for SL 25, 6MV.

Four kernels were generated for each modality, this only needs to be done once. Below is the variance for each Deconvolution fit:

Table 6.1: Variance of the deconvolution fit for all modalities

<table>
<thead>
<tr>
<th>Modality</th>
<th>Variance of Deconvolution Fit (5 exponentials)</th>
</tr>
</thead>
<tbody>
<tr>
<td>SL15:6MV</td>
<td>2.0966x10^{-3}</td>
</tr>
<tr>
<td>SL15:10MV</td>
<td>9.8552x10^{-3}</td>
</tr>
<tr>
<td>SL25:6MV</td>
<td>1.2807e10^{-3}</td>
</tr>
<tr>
<td>SL25:10MV</td>
<td>1.6861e10^{-2}</td>
</tr>
</tbody>
</table>
Finally, generating a central axis report file, this shows the calculated percent depth dose versus the measured percent depth dose. The maximum percentage difference between Dosimetry Check model as compared to measured depth dose profiles is as following:

Table 6.2: Percent difference between Dosimetry Check model versus measured percent dose curves

<table>
<thead>
<tr>
<th></th>
<th>SL15</th>
<th>SL25</th>
</tr>
</thead>
<tbody>
<tr>
<td>6MV</td>
<td>1.95%</td>
<td>1.98%</td>
</tr>
<tr>
<td>10MV</td>
<td>2.42%</td>
<td>2.38%</td>
</tr>
</tbody>
</table>

### 6.2 CT number to Density Conversion

A CT number to density conversion fit, figure 6.2, must also be performed since this affects the calculated dose\(^{10}\). The CT numbers with the corresponding density values were retrieved from Pinnacle TPS and were entered into Dosimetry Check for a polynomial fit, below are the values and the fit:

Table 6.3: CT number to density conversion fit

<table>
<thead>
<tr>
<th>CT Number Value</th>
<th>Density (g/cc)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>0.00</td>
</tr>
<tr>
<td>500</td>
<td>0.50</td>
</tr>
<tr>
<td>946</td>
<td>0.95</td>
</tr>
<tr>
<td>1020</td>
<td>1.00</td>
</tr>
<tr>
<td>1116</td>
<td>1.10</td>
</tr>
<tr>
<td>1144</td>
<td>1.20</td>
</tr>
</tbody>
</table>
A 3rd order polynomial fit was done to achieve accuracy especially in the biological range. A lesser order fit will result in a linear slope changing the equivalent depth of computation.

### 6.3 Water Phantom Measurements

Next, solid water phantoms were sampled in Dosimetry Check. The measured data below is for a 50x50x50 cm$^3$ water phantom for a field size of 20x20 cm$^2$ for 25MUs and 100SSD, figures 6.3-6.6. A 25x25 cm$^2$, 100MU flood field image was used to correct for pixel variation and a 10x10 cm$^2$, 100MU open field was used to center and normalize.
Figure 6.3: 6MV SL25 central axis profile at 1.6cm dmax in a 50x50x50cm³ water phantom for field size 20x20cm². Solid line=DC, dotted line=TPS

Figure 6.4: 6MV SL25 Cross plane profile at 1.6cm dmax along central axis in a 50x50x50cm³ water phantom for a field size 20x20cm². Solid line=DC, dotted line=TPS
Figure 6.5: 10MV SL25 central axis profile at 2.1cm dmax in a 50x50x50cm³ water phantom for field size 20x20cm². Solid line=DC, dotted line=TPS

Figure 6.6: 10MV SL25 Cross plane profile at 2.1cm dmax along central axis in a 50x50x50cm³ water phantom for a field size 20x20cm². Solid line=DC, dotted line=TPS
As seen there is some disagreement between Dosimetry Check and the TPS from the surface to $d_{\text{max}}$, more prominent at 10MV. This buildup region along the central axis is where the secondary electrons travel downstream before depositing their energy, hence the fluence of charge particles and the absorbed dose builds up to a maximum depth\(^6\). For 10MV, the secondary electrons have higher energies traveling downstream longer and so have a more prominent buildup region. The data was taken during annual calibrations using a cylindrical Wellhofer chamber CC13 with a volume of 0.13cm\(^3\) which is large causing cavity perturbations near the surface of the water tank. A parallel plate chamber, which has a small electrode spacing, would have allowed for more accurate measurements without any significant wall attenuation\(^6\). Another factor is Dosimetry Check does not model electron contamination which has an effect from the surface to $d_{\text{max}}$. Renner states that the fitted spectrum in Dosimetry Check as compared to Monte Carlo computed kernels is reliable from $d_{\text{max}}$ and deeper\(^1\). But for higher energies such as 10MV, where secondary electrons have larger ranges, electron contamination could have an effect down to 5-6cm of 2-3\(^\%\)\(^1\).

### 6.4 Water Phantom Measurements with inhomogeneity

Figures 6.7-6.8 show a cross plane image of a water phantom for 6MV with an air pocket in the center of the water phantom. Dosimetry Check’s algorithm is compared as to Pinnacle’s adaptive convolution algorithm with scatter homogeneity factor turned on as an option. This was done because Dosimetry Check does not account for scatter
heterogeneity and only accounts for scatter as if it were in a homogeneous medium\textsuperscript{14}. As expected, the pencil beam algorithm overestimates the dose within this region\textsuperscript{14}.

Figure 6.7: 50x50x50cm\textsuperscript{3} water phantom with air pocket to compare algorithms for 6MV, SL25 Magenta is Dosimetry Check, green is Pinnacle’s Adaptive convolution algorithm with scatter homogeneity
Figure 6.8: 50x50x50cm³ water phantom with air pocket to compare algorithms for 10MV, SL25 Magenta is Dosimetry Check, green is Pinnacle’s Adaptive convolution algorithm with scatter homogeneity

Unfortunately, the impact of inhomogeneity corrections using a pencil beam algorithm with IMRT is more complex due to the combination of small fields, the inhomogeneity, and steep fluence gradients\textsuperscript{14}. A study by Jeraj et al. of three cases, lung, head and neck and prostate was performed to characterize the differences in algorithms. The systematic error of using a fast algorithm for iterations brought a 8% error using a
pencil beam dose calculation even though dose calculation in water were accurate within 1%; while for a superposition algorithm the systematic error was within 1%\textsuperscript{14}.

### 6.5 EPID vs. MapCheck Diode Array using Dosimetry Check for point and volumetric comparisons

A total of 6 previous patient’s treatment plans are used for analysis. The neck case shown first is a 6MV, 7 field IMRT neck boost. The second is a 6MV, 5 field IMRT head and neck case. First looking at point doses, table 6.3 shows a comparison of the MapCheck data for point doses at isocenter and the calc point as compared to the TPS, denoted as MC dose, and the EPID based point dose as compared to the TPS denoted as EPID dose.

Table 6.4: Point dose for MapCheck, EPID, and TPS using Dosimetry Check software for two head and neck cases

<table>
<thead>
<tr>
<th>7-field Neck</th>
<th>TPS Dose (cGy)</th>
<th>EPID Dose (cGy)</th>
<th>% Difference from TPS</th>
<th>MC Dose (cGy)</th>
<th>% Difference from TPS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Calc Pt</td>
<td>1030.9</td>
<td>1041.4</td>
<td>1.02</td>
<td>1046.7</td>
<td>1.53</td>
</tr>
<tr>
<td>Isocenter Pt</td>
<td>1029.1</td>
<td>1033.4</td>
<td>0.41</td>
<td>1017.5</td>
<td>1.13</td>
</tr>
<tr>
<td>5 field H&amp;N</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Calc Pt</td>
<td>2020.2</td>
<td>2027.1</td>
<td>0.34</td>
<td>2047.8</td>
<td>1.37</td>
</tr>
<tr>
<td>Isocenter Pt</td>
<td>2034.8</td>
<td>2023.4</td>
<td>1.00</td>
<td>2012.6</td>
<td>1.53</td>
</tr>
</tbody>
</table>
Figure 6.9: Transverse, coronal, sagittal view through the Calc point via EPID and MapCheck array. Green is TPS, Magenta is DC Showing the 100 %,(colorwash), 90% and 75% isodose lines for a seven field neck

Figure 6.10: Gamma volume histogram for the PTV with criteria 3%/3mm of plan dose for a seven field neck
Figure 6.9 shows isodose lines for sagittal, coronal, and axial cuts through the calculation plane for images taken with the EPID compared to the TPS, and the MapCheck compared to the TPS. Figure 6.10 shows a gamma volume histogram for the PTV using the EPID and the MapCheck array. The gamma value of $\leq 1.0$ represents when the dose agrees within 3% or less than 3mm distance to a point where the dose is the same as the TPS\(^8\). Using the EPID, 93.48% of the PTV passes this criterion while only 85.13% of the PTV passes using the MapCheck diode array.

In figure 6.11, the dose volume histogram for both the EPID and the MapCheck array are shown. The dotted line is the treatment planning system and the solid is Dosimetry Check’s calculated dose. As seen, there are variations with different volumes such as the trachea carina, which is mostly air. A 2.7% standard deviation in dose is calculated for the PTV structure by Dosimetry Check using the EPID as compared to a 3.3% standard deviation using the MapCheck device. There is a 1.6% standard deviation in dose from the TPS for the PTV structure. A discussion of these results will be presented in chapter 7.
Figure 6.11: Dose volume histogram via the use of MapCheck diode array and EPID for various structures. Solid line=DC, dotted is the TPS for a seven field neck
Below are the comparisons for the 5 field IMRT head and neck boost.

Figure 6.12: Transverse, coronal, sagittal view through the calc point, green is TPS, magenta is DC for via EPID and MapCheck array. Showing the 100%,(colorwash), 90% and 75% isodose lines for 5 field H&N
Figure 6.12 shows isodose lines for sagittal, coronal, and axial cuts through the calc pt for images taken with the EPID compared to the TPS, and the MapCheck compared to the TPS. For the head and neck boost, figure 6.13 shows 92.55% of the volume passed a gamma value of 3%/3mm using the EPID; while 75.64% of the volume passes the same criteria using the MapCheck array.

Figure 6.14 shows the dose volume histogram for both the EPID and the MapCheck array. Using the MapCheck, the DVH shows the PTV is hotter than the calculated TPS and a larger percentage of the PTV volume is not covered by the prescription dose. A 3.4% standard deviation in dose is calculated for the PTV structure by Dosimetry Check using the EPID as compared to a 4.6% standard deviation using the MapCheck device. There is 1.8% standard deviation in dose from the TPS for the PTV structure.
Figure 6.14: Dose volume histogram via the use of MapCheck diode array and EPID for various structures. Solid line=DC, dotted is the TPS for 5 field H&N
Next, we look at two pelvis IMRT plans. The first plan is a 10MV 7 field IMRT pelvic boost; the second plan is a 10MV 7 field prostate boost. The point doses are as following:

Table 6.5: Point dose for MapCheck, EPID, and TPS using Dosimetry Check software for two pelvic cases

<table>
<thead>
<tr>
<th>Pelvic Boost</th>
<th>TPS Dose (cGy)</th>
<th>EPID Dose (cGy)</th>
<th>% Difference from TPS</th>
<th>MC Dose (cGy)</th>
<th>% Difference from TPS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Calc Pt</td>
<td>1500.3</td>
<td>1473.8</td>
<td>1.76</td>
<td>1472.6</td>
<td>1.85</td>
</tr>
<tr>
<td>Isocenter Pt</td>
<td>1487.7</td>
<td>1471.4</td>
<td>1.08</td>
<td>1457.3</td>
<td>2.02</td>
</tr>
<tr>
<td>Prostate Boost</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Calc Pt</td>
<td>3340.0</td>
<td>3309.4</td>
<td>0.92</td>
<td>3384.4</td>
<td>1.33</td>
</tr>
<tr>
<td>Isocenter Pt</td>
<td>3328.0</td>
<td>3318.9</td>
<td>0.27</td>
<td>3325.3</td>
<td>0.08</td>
</tr>
</tbody>
</table>

Figure 6.15: Transverse, coronal, sagittal view through the calc point, green is TPS, magenta is DC for via EPID and MapCheck array. Showing the 100%,(colorwash), 90% and 75% isodose lines for a seven field Pelvis.
Figure 6.15 shows isodose lines for sagittal, coronal, and axial cuts through the calc pt for images taken with the EPID compared to the TPS, and the MapCheck compared to the TPS. For the pelvic boost, 96.62% of the volume passed a gamma value of 3%/3mm using the EPID; while 90.40% of the volume passes the same criteria using the MapCheck array given in figure 6.16. A 1.8% standard deviation in dose is calculated for the PTV structure by Dosimetry Check using the EPID as compared to a 3.2% standard deviation using the MapCheck device. There is a 1.5% standard deviation in dose from the TPS for the PTV structure.

In figure 6.17, the dose volume histogram for both the EPID and the MapCheck array are shown. There is considerable dose difference for the rectum as depicted by MapCheck and Dosimetry Check. There is a 27.4% standard deviation in dose for the rectum structure by MapCheck as compared to a 20.6% standard deviation using Dosimetry Check; the TPS produced a 19.5% standard deviation in dose for the rectum.
Figure 6.17: Dose volume histogram via the use of MapCheck diode array and EPID for various structures. Solid line=DC, dotted is the TPS for a seven field pelvis.
Below are comparisons for the prostate boosts, the bladder and bowel densities were overridden to water in the TPS and Dosimetry Check since there was contrast present.

![Figure 6.18: Transverse, coronal, sagittal view through the Calc point, green is TPS, magenta is DC for via EPID and MapCheck array. Showing the 100%, (colorwash), 90% and 70% isodose lines for a seven field prostate](image)

![Figure 6.19: Gamma volume histogram for the PTV with criteria 3%/3mm of plan dose for a seven field prostate](image)
For the prostate boost, 96.46% of the volume passed a gamma value of 3%/3mm using the EPID; while 92.75% of the volume passes the same criteria using the MapCheck array, shown in figure 6.19. A 2.1% standard deviation in dose is calculated for the PTV structure by Dosimetry Check using the EPID as compared to a 2.2% standard deviation using the MapCheck device. There is a 1.4% standard deviation in dose from the TPS for the PTV structure. The dose volume histogram in figure 6.20 for MapCheck and Dosimetry are very consistent. This is most likely due to the bladder density overridden to water’s density.
Figure 6.20: Dose volume histogram via the use of MapCheck diode array and EPID for various structures. Solid line=DC, dotted is the TPS for 7 field prostate
Next, we look at a lung IMRT plan. The plan is a 6MV, 5 field plan. The density of the heart was overridden to water in TPS and in Dosimetry Check.

Table 6.6: Point dose for MapCheck, EPID, and TPS using Dosimetry Check software for lung case

<table>
<thead>
<tr>
<th>5-field Lung</th>
<th>TPS Dose (cGy)</th>
<th>EPID Dose (cGy)</th>
<th>% Difference from TPS</th>
<th>MC Dose (cGy)</th>
<th>% Difference from TPS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Isocenter Pt</td>
<td>2755.0</td>
<td>2734.5</td>
<td>0.74</td>
<td>2678.0</td>
<td>2.79</td>
</tr>
</tbody>
</table>

Figure 6.21: Transverse, coronal, sagittal view through the isocenter point, green is TPS, magenta is DC for via EPID and MapCheck array. Showing the 100%,(colorwash), 90% and 70% isodose lines for 5 field lung
For the lung boost, figure 6.22 shows 88.10% of the volume passed a gamma value of 3%/3mm using the EPID; while 75.94% of the volume passes the same criteria using the MapCheck array. A 2.9% standard deviation in dose is calculated for the PTV structure by Dosimetry Check using the EPID as compared to a 6.9% standard deviation using the MapCheck device. There is a 1.4% standard deviation in dose from the TPS for the PTV structure.

Figure 6.23 shows the dose volume histogram. All the structures are similar when compared to MapCheck and Dosimetry Check except for the boost PTV. The boost PTV is hotter in terms of GVH when analyzed with the MapCheck versus the EPID. There were larger hot spots at interface regions of lung and tissue probably due to the way Dosimetry Check software handles lower resolution images from MapCheck by interpolating between measurement points.
Figure 6.23: Dose volume histogram via the use of MapCheck diode array and EPID for various structures. Solid line=DC, dotted is the TPS for 5 field lung
6.6 Planar Dose Comparison between Dosimetry Check vs. MapCheck

Since the MapCheck software is not yet capable of reading Elekta EPID images, 1-D profile comparison, which both softwares are capable off, would give a reasonable assessment. Using the MapCheck device, dose comparison 1-D profiles were looked at 5cm depth along the beam’s central axis. Since the diode array has an approximate 2cm of water equivalent buildup, another 3cm buildup was place above the device resulting in a 5cm depth. For some plans, 3cm depth was analyzed depending on the tumor location. For the 5 IMRT plans, all fields were taken individually, and an option in the MapCheck software to create a composite of the individual beams was chosen. A composite image was also produced using the TPS.

For Dosimetry Check, images were taken with the EPID. For a more accurate comparison, the IMRT beams were also applied to a water phantom 50x50x50cm³ and the dose calculated to a depth in the TPS depending on the tumor’s location within the body. This was then exported to Dosimetry Check and the fluence images applied to it. The dose plane was also looked at using the CT data set which takes into account attenuation from bone, air, and homogeneous scatter. Figures 6.24 & 6.25 below are coronal views for both MapCheck and Dosimetry Check in a phantom for the 7 field neck plan taken at a 3cm depth.
Figure 6.24: Coronal view at 3cm depth in a phantom and CT by DC for a seven field Neck plan

Figure 6.25: Comparison between measured and plan composite using MapCheck at 3cm depth for a seven field neck plan
Figure 6.26: Y-axis comparison through central axis for MapCheck, DC, and Pinnacle at 3cm depth for a seven field neck plan.

Figure 6.27: X-axis comparison through central axis for MapCheck, DC, and Pinnacle at 3cm depth for a seven field neck plan.
From figures 6.26 & 6.27, MapCheck profiles seem to result in a slight under-dose. A discussion of these results will be presented in chapter 7. Next, we look at the 5 field head and neck plan. Figures 6.28 & 6.29 below are coronal views for both MapCheck and Dosimetry Check in a phantom for the 5 field head and neck plan taken at a 5cm depth.

**Figure 6.28:** Coronal view at 5cm depth in a phantom and CT by DC for 5 field head and neck plan

**Figure 6.29:** Comparison between measured and plan composite using MapCheck at 5cm depth for 5 field head and neck plan
Figure 6.30: Y-axis comparison through central axis for MapCheck, DC, and Pinnacle at 5 cm depth for 5 field head and neck plan.

Figure 6.31: X-axis comparison through central axis for MapCheck, DC, and Pinnacle at 5 cm depth for 5 field head and neck plan.
From figures 6.30 & 6.31, MapCheck profiles seem to result in under-dosing. A discussion will be presented in chapter 7.

For the two pelvic IMRT plans, 1-D dose profiles were compared at 8cm depth along the beam’s central axis since the planning treatment volume was deep within the patient. A similar procedure was applied as stated above but 6cm plastic buildup was placed above the MapCheck device resulting in an 8cm depth. Figures 6.32 & 6.33 below are coronal views for both MapCheck and Dosimetry Check in a phantom for the 7 field pelvic boost plan taken at a 8cm depth.

Figure 6.32: Coronal view at 8cm depth in a phantom and CT by DC for a seven field pelvic boost plan
Figure 6.33: Comparison between measured and plan composite using MapCheck at 8cm depth for a seven field pelvic boost plan

Figure 6.34: Y-axis comparison through central axis for MapCheck, DC, and Pinnacle at 8cm depth for a seven field pelvic boost plan
Figures 6.34 & 6.35 both show Dosimetry Check and MapCheck to be comparable to the TPS. That is why 1-D profiles through a plane are not sufficient in determining the goodness of delivery and dose deposition within the patient.

Next, we look at the 7 field prostate boost plan. Figures 6.36 & 6.37 show a coronal view of the isodose overlay at 8cm depth in a water phantom, the patient’s CT, and on the MapCheck software.
Figure 6.36: Coronal view at 8cm depth in a phantom and CT by DC for a seven field prostate boost plan

Figure 6.37: Comparison between measured and plan composite using MapCheck at 8cm depth for a seven field prostate boost plan
Figure 6.38: Y-axis comparison through central axis for MapCheck, DC, and Pinnacle at 8cm depth for a seven field prostate boost plan.

Figure 6.39: X-axis comparison through central axis for MapCheck, DC, and Pinnacle at 8cm depth for a seven field prostate boost plan.
Again, from figures 6.38 & 6.39 both show Dosimetry Check and MapCheck to be comparable to the TPS.

For the lung IMRT plans, measurements were taken at 5cm depth. Figures 6.40 & 6.41 show a coronal view of the isodose overlay at 5cm depth in a water phantom, the patient’s CT, and on the MapCheck software.

Figure 6.40: Coronal view at 5cm depth in a phantom and CT by DC for 5 field lung plan
Figure 6.41: Comparison between measured and plan composite using MapCheck at 5cm depth for 5 field lung plan

Figure 6.42: X-axis comparison through central axis for MapCheck, DC, and Pinnacle at 5cm depth for 5 field lung plan
Figure 6.43: Y-axis comparison through central axis for MapCheck, DC, and Pinnacle at 5cm depth for 5 field lung plan

Again, from figures 6.42 & 6.43 both show Dosimetry Check and MapCheck to be comparable to the TPS. That is why 1-D profiles through a plane are not sufficient in determining the goodness of delivery and dose deposition within the patient. Without 3D dose reconstruction, the cumulative effects of individual beam differences cannot be fully appreciated such as when using MapCheck.
Chapter 7

Discussion

Beginning with point doses, for the same plan, the EPID versus the diode array was compared. As seen in all 5 plans, the EPID produces higher accuracy in point doses as compared to the TPS versus the MapCheck. This is due to the superiority of the EPID as compared to the MapCheck diode array. The pixel size of the EPID is $0.025 \times 0.025 \text{cm}^2$ as compared to MapCheck which has a 1 cm detector resolution.

Since MapCheck is not yet capable of producing 3-D dose evaluations. All plans analyzed in MapCheck passed the distance-to-agreement (DTA) criteria of the dose agreeing within 3%, or less than 3mm distance to a point where the dose is the same as the TPS. The plans all passed above 90% for the 3%/3mm criteria. However, when the plans were analyzed using the gamma volume histogram (GVH), which is typically defined by the gamma function, there is a lower pass rate for plans analyzed with the MapCheck diode array. The gamma function for the GVH is where the acceptance criteria, 3%/3mm forms a sphere around the dose at a particular point in question. If the calculated dose passes through the sphere, the calculation passes the acceptance test.
In all 5 IMRT cases, 90% of the PTV passed the criteria of gamma value less than one using the EPID and 75% using the MapCheck diode array. An exception needs to be made for the lung case where inhomogeneity is high. As stated earlier, the pencil beam algorithm can overestimate the dose within this area. Nevertheless, 88% of the PTV for the lung plan did pass the 3%/3mm criteria.

This lower pass rate using the MapCheck diode array could be due to the Dosimetry Check software handling lower resolution images from the MapCheck device and interpolating between measurement points. However, the fact remains that evaluating points on a single plane on a phantom when using DTA analysis is not as comprehensive or accurate as looking at a GVH, where the dose has be reconstructed onto the patient’s CT and then analyzed volumetrically by gamma analysis.

The planar dose maps and 1-D profiles along the beam’s central axis seem to produce a very similar comparison. A tendency that MapCheck seem to exhibit is the under-dosing for head and neck plans and the lung plans where inhomogeneity is present. Comparisons between MapCheck and Dosimetry Check for pelvic plans seem to be equivalent. This could be the under-response of the MapCheck device when measuring fewer than 10 MUs as demonstrated by Li et al\(^7\). Other possibilities could be field size or SSD dependence. Unfortunately, this is crucial since IMRT beam segments can usually have monitor units less than 10MUs.

Both Dosimetry Check and MapCheck 1-D profiles for all plans were within one standard deviation of the Pinnacle 1-D profile. This is major concern when using MapCheck. When using MapCheck, the assumption is that acceptable measured results on a single plane equate to an approved treatment for target volume and critical
structures, when in actuality machine delivery errors and internal patient scatter can considerably change the dose distribution within the patient.

Using analyzing tools in Dosimetry Check such as gamma volume histogram and dose volume histograms provides a mean to confirm the computed and measured dose in a volumetric geometry. Measurements of dose in three-dimensional space reconstructed on the patient’s CT provide the cumulative effects of all individual beams and their differences. That is why 1-D profiles in MapCheck may not provide the best resource in approving a treatment plan.

As mentioned earlier, spikes are present in most 1-D profiles due to sudden change in image being amplified. The deconvolution is a high pass spatial filter, since the point spread function of the EPID is a low pass filter\(^1\). Any sudden change in image will be amplified as that comprises of high spatial frequencies and will be imaged as spikes; that is why the area of the fluence images should be restricted, this also helps save on computational time\(^1\).

Overall, this concludes that Dosimetry Check can be superior to other available IMRT QA systems. The versatility of Dosimetry Check being able to use the EPID including film, ion chambers, and the MapCheck diode array to analyze plans is exceptional. However, the key advantages of the Dosimetry Check software over the existing systems is the use of the EPID unit, often superior to other IMRT QA tools in resolution, contrast, and the convenience of being a part of accelerator system. Another key advantage for Dosimetry Check is using the planning CT data set which allows for dose superimposition and provides a feedback loop of treatment delivery with direct comparison with treatment planning dosimetry in 3D space.
Chapter 8

Conclusion

The Dosimetry Check program can provide a means for a second check using a fast pencil beam algorithm to reconstruct the dose. A major advantage of Dosimetry Check over existing IMRT QA packages is that it does not require the purchase of additional detector hardware, since most linear accelerators are equipped with EPIDs. A major disadvantage is Dosimetry Checks lacks the ease of MapCheck of initial dose calibrations and instantaneous comparisons. Full beam modeling needs to be completed, which can be tedious, before the software can be used.

Here, the difference between point, planar and volumetric doses with MapCheck and Dosimetry Check via the use of the EPID and the diode array by using evaluating tools such as point doses at isocenter, 1-D profiles, gamma volume histograms, and dose volume histograms for IMRT dose comparison in three types of cases: head and neck, prostate, and lung was studied. The point doses using the EPID have a smaller percentage difference as compared to the TPS for all IMRT plans. The 1-D profiles along the beam’s central axis seem to produce similar comparisons within 1 standard
deviation from the TPS. That is why producing 1-D profiles in MapCheck should not determine the approval of a treatment plan.

In all 5 IMRT cases, 90% of the PTV using the EPID and 75% of the PTV using the MapCheck diode array passed the criteria of gamma value ≥1 for 3mm/3%. The superiority of EPID in pixel resolution and contrast for images as compared to the MapCheck diode array produced a difference in pass rates for the same plan.

Dosimetry Check provides a means to verify the treatment plan delivery within the patient accounting for anatomy, validations show that the system is capable of achieving accuracy to the 3% level. Without 3D dose reconstruction, the cumulative effects of individual beam differences cannot be fully appreciated such as using MapCheck. Dosimetry Check can be a valuable tool for IMRT QA as it uses the patient’s CT data set, thus applying patient specific attenuation corrections and the superiority of the EPID. This can help reduce the uncertainty in dose for less variability in delivery and higher pass rates using a variety of analysis tools.
References


Appendix A

Commissioning of Dosimetry Check

Below is the resultant test file for SL25, 6MV containing off-axis correction factors which account for the changes in beam penetration off axis due to the change in beam energy off axis by using the in-water diagonal profiles. A list of off-axis angles is produced based on the radius and depth.

/* File type, 101 = off axis depth correction table */ 101
/* file format version: */ 1
/* machine directory name: */ <*UToledoSL25*>
/* nominal energy MeV */ 6
/* number of depths: */ 3
/* number of tangents: */ 58

// depth cm             tan = radius/distance
0.0000        0.0050        0.0099        0.0148        0.0198
1.70                1.00000      0.99741      0.99833      0.99921      0.99899
10.00               1.00000      1.00015      1.00032      1.00024      1.00142
20.00               1.00000      1.00190      1.00280      1.00337      1.00458

// depth cm             tan = radius/distance
0.0247        0.0296        0.0345        0.0394        0.0443
1.70                0.99711      0.99903      0.99976      0.99921      0.99899
10.00               1.00111      1.00228      1.00109      1.00001      0.99820
20.00               1.00278      1.00225      1.00194      1.00050      0.99718

// depth cm             tan = radius/distance
0.0493        0.0542        0.0591        0.0640        0.0689
1.70                1.00154      1.00045      1.00271      1.00159      1.00241
10.00               0.99634      0.99673      0.99598      0.99688      0.99751
Below is the program file manually adjusted to reflect the measured output factors for the SL25 linear accelerator, 6MV:

```plaintext
/* file type: 5 = output factors */ 5
/* file format version: */ 1
/* machine directory name */ <UToledoSL25>
/* energy */ 6
/* date of file: */ <1-Mar-2010 11:22:24>

//Normally only square fields.
//        cm   cm        cm     cG/mu
//     field size    SSD    Depth  output factor
2.00    2.00   100.0    1.70    0.8921
3.00    3.00   100.0    1.70    0.9196
4.00    4.00   100.0    1.70    0.9416
5.00    5.00   100.0    1.70    0.9507
6.00    6.00   100.0    1.70    0.9654
8.00    8.00   100.0    1.70    0.9839
10.00   10.00  100.0    1.70    1.0000
12.00   12.00  100.0    1.70    1.0142
14.00   14.00  100.0    1.70    1.0269
16.00   16.00  100.0    1.70    1.0373
18.00   18.00  100.0    1.70    1.0469
20.00   20.00  100.0    1.70    1.0558
25.00   25.00  100.0    1.70    1.0699
```

The in-water phantom scatter is computed with the pencil beam kernel and fitted to beam data by program GenerateBeamParameters. Below is the resultant text file for in-water output factors for SL25, 6MV:

```
Scatter collimator table for 6 MV:
For depth = dmax = 1.70 cm
Field size cm  Sc  Sp  Scp
2.0  2.0   0.938  0.951  0.892
3.0  3.0   0.949  0.969  0.920
4.0  4.0   0.965  0.976  0.942
```