



EPID-based Dosimetry for Remote Auditing of Radiotherapy Clinical Trials

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Abstract

The objective of this research is to implement a novel approach for remote dosimetric auditing of clinical trials. The audit should ensure an accurate dose delivery at different radiotherapy centres with minimum cost.

High variation and complexity of planning and delivery systems may result in discrepancy of dose delivery for the trials. The deliveries are assessed to reduce variability and improve reliability of the trials. The assessment is conducted through rigorous quality assurance (QA) and/or external dosimetric audits. Conventionally, an independent centre performs external audits by site visits or mailing phantoms and dosimeters.

This research presents an innovative approach to remotely audit dose deliveries for clinical trials performed at centres in Australia and New Zealand. Participants are provided with CT data sets of two trial patients and two virtual phantoms. They plan the trials for intensity modulated radiotherapy (IMRT) and/or volumetric modulated arc therapy (VMAT) deliveries using local treatment planning systems (TPSs). Then, they send in-air acquired images from their electronic portal imaging devices (EPIDs) to the auditing site. The EPIDs provide relatively consistent data acquisition system for analysis significantly reducing the audit cost.

A model was developed using images from aS1200 EPIDs for verification of IMRT dose distribution from deliveries of TrueBeam linacs. The model was based on published methods and a clinically established IMRT QA procedure for Varian C-series linacs. Similarly, an Elekta specific model was developed for deliveries of Elekta systems and the results were compared to those from Varian specific model. Minor improvement was observed for the vendor specific models. The QA method was extended for remote auditing of IMRT/VMAT deliveries. The audit instruction provided benchmark planning exercise of two head and neck (HN) and post-prostatectomy (PP) patients and two flat and cylindrical phantoms for participants. The feasibility of the approach including implementation details was demonstrated over six facilities in a pilot study. Then, the audit results from 30 facilities were used to develop a linear model on explanatory variables. It demonstrated significant influence of TPS-linac, calculation grid resolution and IMRT/VMAT type on the audit outcome. The audit outcome demonstrated high gamma pass rates for the trials and provided results comparable to the established more resource-intensive audit methods.

Statement of originality / Declaration by author

I hereby certify that the work embodied in the thesis is my own work, conducted under normal supervision. The thesis contains no material which has been accepted, or is being examined, for the award of any other degree or diploma in any university or other tertiary institution and, to the best of my knowledge and belief, contains no material previously published or written by another person, except where due reference has been made. I give consent to the final version of my thesis being made available worldwide when deposited in the University's Digital Repository, subject to the provisions of the Copyright Act 1968 and any approved embargo.

The author acknowledges that copyright of published works contained within this thesis resides with the copyright holder(s) of those works.

All the enclosed publications in this thesis are the author's original work. The model for EPID to dose conversion for the Varian aS1000 system was developed by P.Greer based on work by B.King and the software used for the VESPA audit results was developed by P.Greer and B.Zwan. The VESPA audit instructions to the centres was developed by P.Greer.

Narges Miri

20/08/2018

Thesis by publication

I hereby certify that this thesis is in the form of a series of papers. I have included as part of the thesis a written declaration from each co-author, endorsed in writing by the Faculty Assistant Dean (Research Training), attesting to my contribution to any jointly authored papers.

By signing below I confirm that Narges Miri contributed analysis and writing to the papers/publications of J2, J3 and J5 in the publication list.

Joerg Lehmann 21/08/2018

By signing below I confirm that Narges Miri contributed measurement, modelling, analysis and writing to the papers/publications of J2, J3, J4 and J5 in the publication list.

Philip Vial, 20/08/2018,

By signing below I confirm that Narges Miri contributed modelling, analysis and writing to the papers/publications of J1 and J3 in the publication list.

Benjamin Zwan, 21/08/2018

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Peter Greer, 21-08-2018

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List of publications included in the thesis

Peer reviewed journal publications:

J1- N. Miri, P. Keller, B. J. Zwan, and P. Greer, "EPID-based dosimetry to verify IMRT planar dose distribution for the aS1200 EPID and FFF beams," Journal of Applied Clinical Medical Physics, vol. 17, no. 6, 2016.

J2- N. Miri, J. Lehmann, K. Legge, P. Vial, and P. B. Greer, "Virtual EPID standard phantom audit (VESPA) for remote IMRT and VMAT credentialing," Physics in Medicine and Biology, vol. 62, no. 11, p. 4293-4299, 01/2017.

J3- N. Miri, J. Lehmann, K. Legge, B. J. Zwan, P. Vial, and P. B. Greer, "Remote dosimetric auditing for intensity modulated radiotherapy: A pilot study," Physics and Imaging in Radiation Oncology, vol. 4, pp. 26-31, 10/ 2017.

J4- N. Miri, P. Vial, and P. B. Greer., " Remote dosimetric auditing of clinical trials: the need for vendor specific models to convert images to dose" Journal of Applied Clinical Medical Physics, vol. 20, no. 1, 11/2018.

J5- N. Miri, K. Colyvas, K. Legge, J. Lehmann, P. Vial, A. Moore, M. Harris, and P. B Greer, " A remote EPID-based dosimetric TPS-planned audit of centers for clinical trials: outcomes and analysis of contributing factors" Radiation Oncology, vol. 13, p. 178, 2018.

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Thesis Structure

This thesis opens by introducing radiotherapy and auditing methods for radiotherapy clinical trials. Then, the body of the thesis will be presented in nine chapters:

Chapter 1) Introduction:

- Chapter 1 introduces an overview of radiotherapy process and quality assurance (QA). It provides a background on the QA of radiotherapy clinical trials and conventional methods for dosimetry auditing. Challenges for current audits open the new approach for the audit.

Chapter 2) Literature review and research design:

- Chapter 2 presents a literature review on conventional dosimetric auditing methods for intensity modulated radiotherapy (IMRT) and volumetric modulated arc therapy (VMAT) deliveries. It then reviews current methods on 2D and 3D dosimetry methods for images from electronic portal imaging devices (EPIDs). Required corrections and calibrations are explained for the images. Then, the chapter outlines the concept of the new approach, virtual EPID standard phantom audit (VESPA), for dosimetric auditing.

Chapter 3) Modelling for Truebeam systems:

- Chapter 3 performs a dosimetry commissioning on aS1200 EPIDs from Truebeam linear accelerators (linacs) compared with aS1000 EPIDs from Varian C-series. Then, a model is developed to convert aS1200 EPID signals to dose inside a virtual flat phantom. The delivered dose is then compared with calculated TPS dose to assess accuracy of the deliveries. This chapter was presented in a journal paper [J1].

Chapter 4) Modelling for Elekta systems:

- Chapter 4 follows on Chapter 3 by presenting a model development for Elekta system deliveries. It evaluates relevant dosimetric differences between Varian and Elekta systems and whether the audit requires a vendor specific model for auditing purpose. This chapter was presented in a journal paper [J4].

Chapter 5) Remote Auditing:

- Chapter 5 introduces a novel approach to remotely audit radiotherapy clinical trials. The approach has a potential to significantly reduce the audit cost. This chapter explains

implementation of the method for auditing IMRT/VMAT deliveries. The material in Chapter 5 was presented in a journal paper [J2].

Chapter 6) A pilot auditing:

- Chapter 6 follows on from chapter 5 by applying the method for six pilot centres. The centres provide pre-treatment IMRT images from their EPIDs while the auditing site converts the images to dose inside virtual phantoms and assesses accuracy of each delivery. The material in Chapter 6 was presented in a journal paper [J3].

Chapter 7) Overall auditing:

- Chapter 7 studies the audit outcome for several remote IMRT/VMAT deliveries. It compares the results with conventional audits and introduces the significance of explanatory variables on the audit outcome. The material in Chapter 7 was presented in a journal paper [J5].

The thesis is concluded in Chapter 8, with a discussion followed by suggestions for future research opportunities in dosimetry auditing of radiotherapy clinical trials in Chapter 9.

Introduction

Background

Depending on the outcome of individualised biological assays and available techniques, an optimal treatment method (e.g. surgery, chemotherapy, hormone-therapy, radiotherapy or any combination of these) is prescribed for cancer treatment [1]. Almost half of all patients benefit from radiotherapy treatments [2]. Radiotherapy is a minimally invasive method which sterilises tumour cells by using ionising radiation [2]. The beams are produced from a generator, mainly linear accelerators (linacs), placed in a shielded treatment room. To increase the therapeutic ratio, the absorbed dose to the normal tissues should be minimised while delivering maximum dose to the tumour cells. An accurate treatment planning system (TPS) and a precise technique is required to calculate the optimised dose and deliver an accurate dose. Accuracy of the calculation and delivery is verified by methods such as conventional quality assurance (QA) procedures. In the context of clinical trials, a dosimetric audit provides a controlled environment to minimise dependency of the trial outcome on stochastic and systematic errors. This aims to reduce the trial cost and enhance the outcome reliability [3]. Conventional methods for auditing can be labour intensive and/or expensive.

Radiotherapy

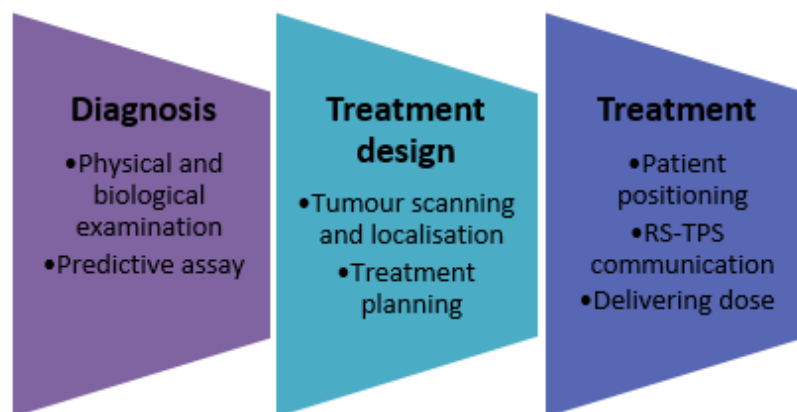


Figure 1-1- Mainstream workflow for cancer treatment: diagnosis to treatment [Radiation Source: RS, Treatment planning system: TPS].

Tumour characterisation and staging is performed using different methods such as biological examinations, biopsy and imaging (e.g. CT-scan, spectroscopy, MRI, PET, ultrasound and X ray) [4]. Appropriateness of radiotherapy treatment is determined patient by patient using predictive assays, e.g. test of oxygen level, proliferation rate and intrinsic cellular radiosensitivity of the tumour and surrounding healthy tissues [5]. Other determinant factors

are tumour type and location, comorbidities and previous medical history (especially previous radiotherapy). Almost half of the patients are referred for radiation oncology treatment.

Medical images are acquired from the tumour site to simulate the treatment by the TPS. When acquiring images, the patient is tattooed to set a relatively fixed mark for patient positioning. The simulation helps doctors plan an accurate geometry for the treatment [6]. Pencil beam, convolution-superposition and Monte Carlo methods are three major model-based dose calculation systems used for dose prediction in inverse planning [7]. Currently, several TPSs provided by different companies are being used clinically, such as Pinnacle (by Philips Healthcare), Monaco (by Elekta) and Eclipse (by Varian) [7, 8].

In order to increase therapeutic ratio, a high tumour control probability (TCP) must be achieved with a minimal risk of normal tissue complications (normal tissue complication probability, NTCP). This requires an accurate irradiation of the planning target volume (PTV). Minimising the irradiation to marginal volume of the PTV is achieved through different localisation strategies, e.g. patient marking when scanning, laser alignment, adjustable treatment couch and different imaging modalities [34]. Figure 1-1 demonstrates the mainstream workflow for cancer treatment.

External beam radiotherapy (EBRT) is mainly performed using linacs. In 1980, linacs were equipped with multileaf collimators (MLCs) to shape fields instead of using shielding blocks, and three-dimensional conformal radiotherapy (3D CRT) was introduced as a standard treatment method [9]. Soon after, intensity modulated radiotherapy (IMRT) was presented as a more precise method by introducing MLC motion while irradiating the beam [10]. IMRT allows high control of the dose distributions and simultaneous irradiation with different doses to different target volumes [11, 12]. In 1995, intensity modulated arc therapy (IMAT) was introduced as an alternative to IMRT. IMAT delivers planar/non-coplanar doses while varying MLC apertures and rotating the gantry [13]. The relatively low treatment time of IMAT can improve clinical throughput. Otto introduced an improved planning method for IMAT and termed the technique volumetric modulated arc therapy (VMAT) which quickly gained widespread acceptance [14]. Figure 1-2 demonstrates typical 3D CRT, IMRT and VMAT treatments.

To simulate treatments, a TPS is used to calculate dose [15]. Assessment of the calculation is normally performed using dose volume histograms (DVH) showing the dose distribution inside the tumour and peripheral normal tissues [15]. For less complicated treatments, e.g. 3D CRT, a forward planning method is normally used to design the treatments. An initial plan is made by

a radiation therapist (RT) based on the department protocol and the RT decides on the beam number, delivery angles, attenuations and configuration of the MLCs. Once the initial plan is made, the TPS specifies the radiation required to deliver the prescribed dose. For more complex tumour shapes and for tumours close to critical organs where IMRT/VMAT treatment method is common, inverse planning is recommended. Initially, an oncologist defines the patient's organs at risk (OAR) and PTV, then an RT performs the TPS optimisation. Therefore, decision in inverse planning is based more on an automated process rather than a trial-and-error determination. Additional pre-treatment dose verifications are required for these treatments [16]. Machine and patient specific quality assurance (QA) measurements are taken by local physicists to ensure the accuracy and stability of IMRT/VMAT deliveries.

An interface software known as record and verify system (R&V) records the information flow to minimise incidents and human errors when data is entered manually during treatments [17]. The R&V provides a basic QA performing the communication between the TPS and linac. It includes an electronic form of the patient, recording fraction number, imaging information and applied shifts.

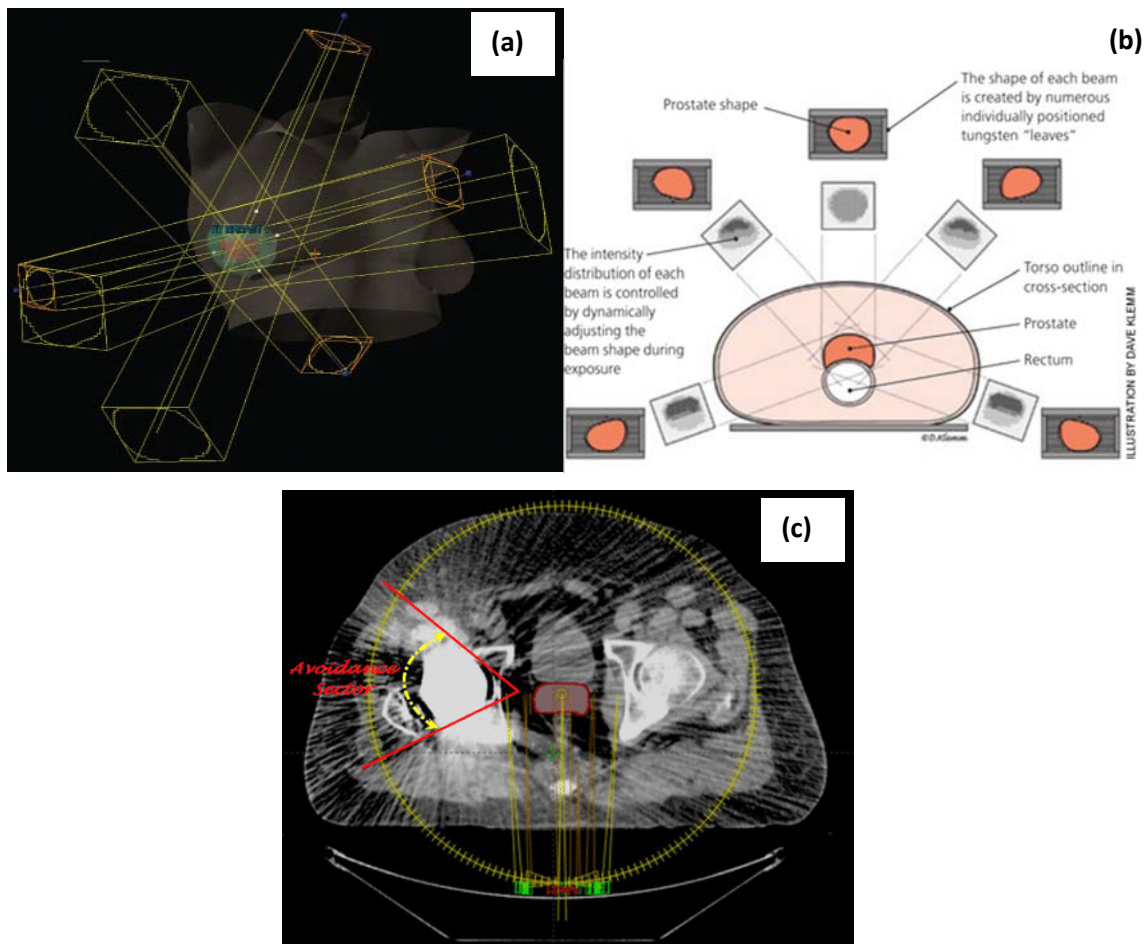


Figure 1-2- A schematic of a) three-dimensional conformal radiotherapy (3D CRT) [18], b) intensity modulated radiotherapy (IMRT) [19] and c) volumetric modulated arc therapy (VMAT) setup for prostate treatment [20].

Quality assurance in Radiotherapy

QA tests ensure the accuracy of radiation therapy treatments. The European Society for Therapeutic Radiology and Oncology (ESTRO) adapted ISO 9000 QA standards of industry as QA standards for radiotherapy [21]. The American Association of Physicists in Medicine (AAPM) has also provided recommendations for local QA tests at Task Group (TG) 142 and TG 43 [22, 23]. Most centres follow relatively *ad-hoc* methods for their QA, while local QA tests could be insufficient to predict all failing results [24].

The recommended QA tests are for machines and patients. Machine QA is a performance-oriented test undertaken frequently on the machines. It checks for the performance of the delivery system and its accuracy to the baselines acquired at the time of the acceptance and commissioning [22]. Patient QA checks are for the TPS calculation, correctness of the data transfer to the linear accelerator and accuracy of the delivered plan. Patient-specific QA tests verify the intended dose distribution and the intensity modulated beams/arcs for each specific patient. Patient-specific QA is basically designed for verification of IMRT deliveries by

checking the relevant plan prior to the patient delivery. Repeating the measurement, exportation, setup and evaluation are suggested actions if the QA result fails. The error could be from the delivery, the device/phantom setup, the ‘wrong plan’ being exported or delivered, exclusion of the couch, an unready and/or non-calibrated detector, using the wrong calibration curve, an outdated calibration detector, too steep a dose gradient for the detector resolution or TPS dose calculation error. De Wagter summarised IMRT verification in four levels of dosimetric QA within a conceptual pyramid, Figure 1-3a [25]. The pyramid combines both periodic machine QAs, levels 1&2, and clinical QAs, levels 3&4. Figure 1-3b presents methodology and tools for the corresponding level. For a new clinical IMRT, though many people start from level 3, Vergote et al recommend to start from level 4, and if the 3D verification shows an unacceptable discrepancy with treatment planning, level 3 is performed [25].

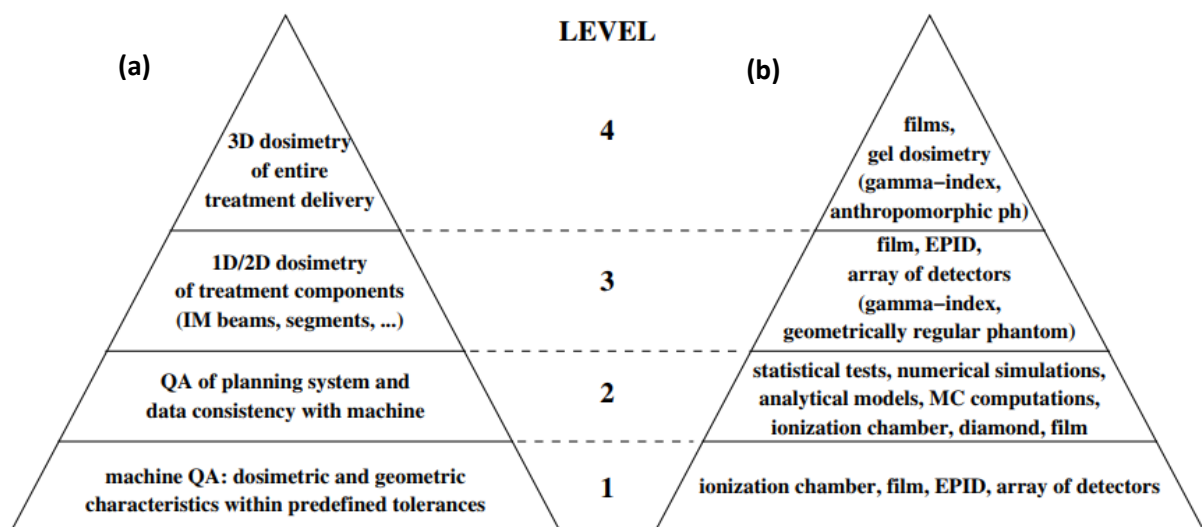


Figure 1-3- Conceptual pyramid correlating the different levels of IMRT verification. Levels 1 and 2 can be part of the machines/tools periodic procedure used for IMRT planning and delivery [25].

Accurate characterisation of the measurement system is required to find the eventual errors within the system. The characterisation should be performed by a dosimeter with an appropriate sensitivity and limited errors. Dosimetry of test plans is used to verify conformal plans. Depending on the beam complexity level, different dosimeters are employed to measure dose. Point dosimeters (e.g. ionisation chambers, diodes) are sufficient for less complicated deliveries such as open field deliveries and 3D CRT. They measure absolute dose at each point by energy averaging over their tip volume. Point by point dose verification is impractical for complex

treatments of IMRT/VMAT. A 2D/3D dosimetry method is more appropriate for these dose verifications [26]. The planned dosimetry method may be followed by an independent in-vivo dosimetry method. In-vivo verification monitors real-time dose of the treatment to control accuracy of the delivery [27, 28]. Thermoluminescent dosimeters (TLDs) and optically stimulated luminescence dosimeters (OSLDs) are proper candidates for in-vivo dosimetry [29]. The metal–oxide–semiconductor field-effect transistor (MOSFET) is another dosimeter which could be embedded in the body cavities to measure dose in vivo [30].

Dosimetry tools and methods

For pre-treatment patient-specific QA of IMRT/VMAT, using 2D dosimeters is prevalent. The dosimeters include arrays of diodes/ionisation chambers, radiographic/radiochromic films or computed radiography [31]. One of the high resolution 2D dosimeters is the electronic portal imaging device (EPID), which includes sensitive arrays converting light to electronic data at a computer used for visualisation and archiving [32-34]. Some studies also place a phantom on the beam path to introduce an attenuating medium. It could be a physical phantom with a simple structure and material or an anthropomorphic phantom mimicking a human organ. Figure 1-4 demonstrates ion chambers embedded in a physical phantom including required instruments to read the measurements.

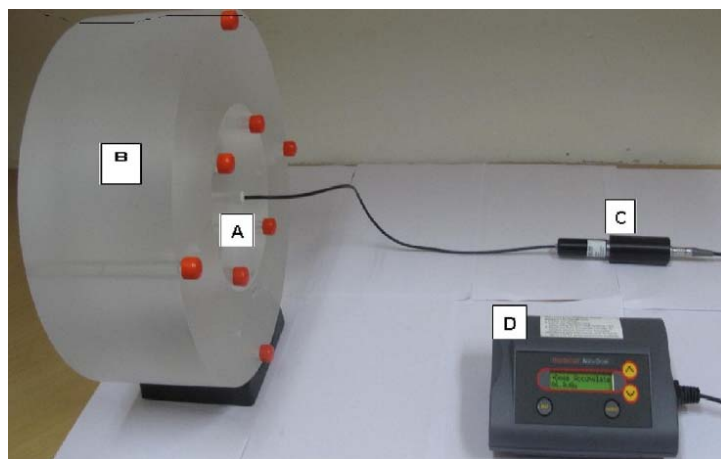


Figure 1-4- a) ionisation chambers, b) acrylic phantom, c) digitiser and d) dosimeter [35].

To view and compare 3D/2D dose distributions, dose is often measured and compared in sagittal, coronal and transverse planes [36]. A mathematical expression is a straightforward method to compare the planar doses. The most common expression is gamma analysis, which

takes into account both dose and spatial differences point by point. The gamma expression is defined as:

$$\gamma(r_e) = \min\{\Gamma\} \forall \{\vec{r}_e\} \quad (1)$$

Where $\vec{r}_e$ is the vector position of the evaluating point, and Γ is defined as:

$$\Gamma = \sqrt{\frac{|\vec{r}_e - \vec{r}_r|^2}{\Delta d^2} + \frac{[\vec{D}_e - \vec{D}_r]^2}{\Delta D^2}} \quad (2)$$

where $\vec{r}_r$, $\vec{D}_e$ and $\vec{D}_r$ are vector positions of the reference points, evaluating dose and reference doses respectively, and Δd^2 and ΔD^2 are the distance to agreement (DTA) and dose difference (DD) criteria respectively, which are mainly decided based on trial and error [37]. However, the most common criteria (accepted threshold) is 3% DD and 3 mm DTA. At high gradient regions, the gamma value is more influenced by defined criteria of the DTA whilst in shallow gradient regions, it is more determined by the DD criteria [38, 39]. Therefore, to determine the accuracy of the dose delivery, there are inherent limitations in defining a reference dose with maximum accuracy and measuring an absolute dose at the equivalent point for comparison. These limitations result in determining different gamma criteria for comparison. To pass IMRT QA, TG218 recommends passing rates of >95% at 3% DD and 2 mm DTA with 10% dose threshold as a tolerance limit. If the pass rate at this criteria decreases to 90%, an action limit is reached and a solution is required. Recommended ‘action levels’ for local QAs are 1) for any daily measurement of $2\% < DD < 4\%$, treatment may continue but the senior physicist must be notified and, 2) for any daily measurement of $4\% < DD$, the treatment has to stop immediately and the senior physicist should resolve the issue [40]. At the multi-centre level, AAPM TG 21 and TG 51 have recommended that the reference dose (machine output) among different centres should not vary more than 2% and the combined uncertainties for treatments should be less than 5% [41]. Eventually, a DVH analysis may be performed to evaluate the clinical relevance of the gamma results, especially when the gamma passing rate fails the tolerance limits.

Use of EPID for dosimetry

EPIDs were originally designed for patient position verification. It was later realised that EPID images contain dose information [31, 42, 43]. They replaced traditional film dosimetry due to the films intrinsic limitations and long post-processing times [44]. EPIDs are useful tools for pre-treatment and/or in-vivo dosimetry as they are readily available on linacs [45-47]. EPID

pre-treatment dosimetry is performed by either simulating the response of the pixel values using empirical techniques/Monte Carlo calculations or converting grayscale signal of the images to dose inside the patient/phantom using models [33, 47-52]. EPID dosimetry has also provided the possibility of 3D dose calculations using mathematical models while conventional 3D dosimetry mainly relies on relatively less robust methods such as embedding point/array dosimeters inside phantoms or utilising commercial gels [26].

Currently, the two main vendors of linacs are Varian and Elekta. They demonstrate some considerable differences in their EPID structures for dosimetry, mainly the detector size and resolution. Varian aS1000 EPIDs have a $40 \times 30 \text{ cm}^2$ active area with a 1024×768 image resolution (resulting in a 0.039 cm pixel resolution), and Elekta iViewGT EPIDs have a $41 \times 41 \text{ cm}^2$ active area with a 1024×1024 image resolution (resulting in a 0.040 cm pixel resolution) [53]. Unlike the aS1000, the iViewGT EPIDs are positioned at a source to detector distance (SDD) of 160 cm.

Varian has also introduced a new type of linac, known as the Truebeam system. It has a new design for the EPID, aS1200, containing additional backscatter shielding layers. The aS1200 EPIDs are attached to the gantry base through a robotic arm, consisting of an array of photodiodes made of amorphous silicon with a phosphor layer on top converting photon energy into electrons. The EPIDs have also been adapted for use in flattening filter free (FFF) beams, without saturation at any source to detector distance. These new detectors have a large active area of $43 \times 43 \text{ cm}^2$ with 1280×1280 pixel arrays, small pixel size of 0.034 cm, and advanced acquisition electronics.

Radiotherapy clinical trials

Quality assurance in radiotherapy clinical trials

Radiotherapy clinical trials are research/experiments undertaken in a controlled radiotherapy environment to assess safety and efficacy of a biomedical/behavioural intervention [54]. Depending on the study stage, four phases are defined for the clinical trials: **Phase I**, which is an establishment stage, determines initial safety and side effects of the trial in a small group [55]. **Phase II** on the other hand tests the trial efficacy by extension of the phase I study to a larger group [56]. **Phase III** studies efficacy, safety and effectiveness of the trial in a very large group and compares them with current standard/conventional methods. It also monitors adverse effects of the intervention within a widespread population [57]. After marketing the method,

long-term and adverse effect of the method in the real world are surveyed and monitored by a **Phase IV** study [56].

Failure in delivering accurate dose may result in severe consequences in treatments and mislead the trial outcome [58, 59]. To ensure the validity of trial results, introduced intervention (e.g. imaging, TPS and delivery accuracy) requires a precise definition and high accuracy. Studies show significant variability of centres to deliver accurate dose in agreement with their TPS and insufficiency of local QAs to predict all failing results [60-63]. ESTRO recommends an additional external audit for independent verification [64]. The audit groups perform independent QAs within corresponding centres and study consistency of the machines performance [65, 66]. They verify the adequacy of local QAs, image guided radiotherapy (IGRT), accuracy of source calibration and TPS calculation [67-69]. An optimal approach provides a real-time, inexpensive and informative audit.

Different organisations have been funded to monitor and assess the accuracy of the trials at a multi-centre level. The International Atomic Energy Agency (IAEA) and World Health Organisation (WHO) are worldwide auditing networks supporting simple beam calibration [70]. Many multi-centre auditing networks are supported by the Imaging and Radiation Oncology Core (IROC), formerly known as the Radiological Physics Centre (RPC), in North America [71]. Significant attempts for credentialing European centres are performed by the Radiation Therapy Oncology Group (RTOG), ESTRO and the European Organization for Research and Treatment of Cancer-Radiation Oncology Group (EORTC-ROG) [72]. Audits are also performed at the national level by multiple regional organisations such as the South East Central Regional Audit Group in UK, and the Trans-Tasman Radiation Oncology Group (TROG) and the Australian Clinical Dosimetry Services (ACDS) in Australia and New Zealand. Figure 1-5 demonstrates dosimetric auditing networks at the international level by IAEA.

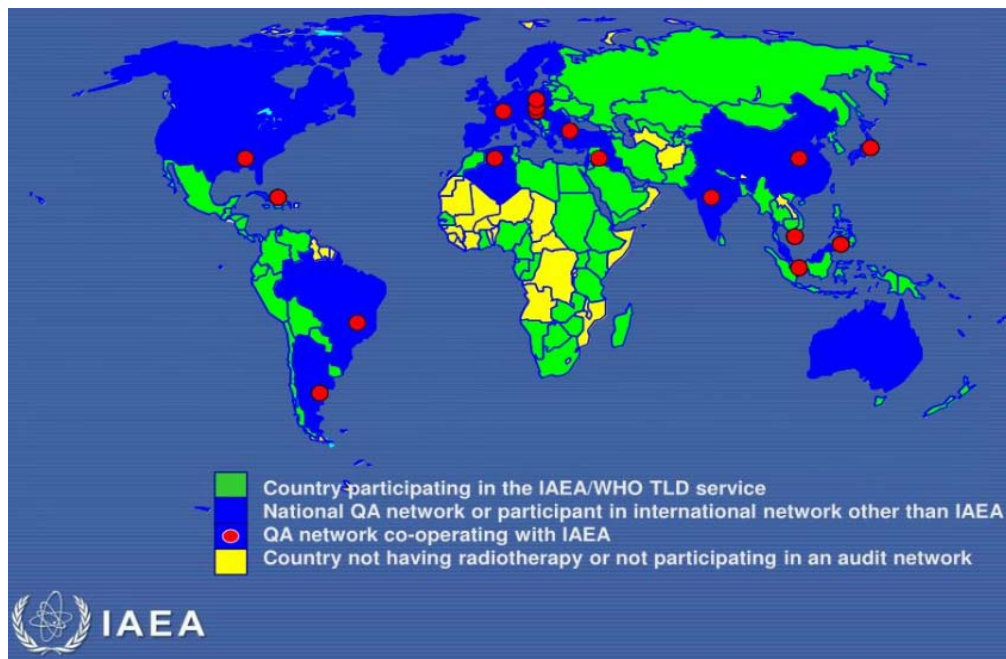


Figure 1-5- Schematic of radiotherapy dosimetric audit networks provided by IAEA/WHO. [International Atomic Energy Agency: IAEA, World Health Organization: WHO] [73].

The benefit of external dosimetric audits for clinical trials have been well demonstrated [74-76]. The enhancements are summarised below.

- 1- Improving efficiency: An external audit improves overall clinical throughput by providing feedback and training. It provides adequate resources for technological implementation by sending feedback to the centres. The feedback originates from studies over a wide range of centres assessing availability, feasibility and safety of new technologies/techniques, machines, TPSs and imaging systems in different centres and the centres' ability to use them. Even if some centres could not participate in the trial, the audit results could provide them an appropriate procedure for review and adjustment with their available techniques. External audits could also improve the overall efficiency of the treatments by staff training. Studies by RTOG showed the QA impact on improving the staffs understanding and compliance with the protocols, and provided them with guidelines [77]. The EORTC demonstrated that participant centres in a “dummy run” were highly successful in following “dummy run” performance and delivery of the protocol compliance with radiotherapy [78].
- 2- Improving reliability: The audit outcome results in data integrity and reliability of the trial. It clarifies and addresses common flaws and ensures that irrelevant factors are omitted so the trial's outcome is reproducible.

- 3- Reducing cost: Implementation of clinical trials could be very expensive and more expensive at higher phases of the study. Auditing minimises the risk of costing on unanswered trials and reduces deviation rate of the trials at different centres. The reduction lowers numbers of required patients for the trial assessment, resulting in lowering the trial cost and quickening the availability of the results [74, 79, 80].

There are different approaches on how to perform, evaluate and describe the QA in clinical trials. The methods should also be updatable with the introduction of new advanced technologies, as they require co-operation and/or consistency between the trial groups for a comprehensive analysis.

Dosimetric auditing methods

Conventionally, an independent centre performs the audit by mailing tools [81-85] or by site visit(s) [86-88]. In mailing methods, phantoms and dosimeters are mailed to the participants and local physicists perform measurements according to provided instructions. The method is easier to schedule for the host centre, though it may lack consistency in the procedure. While successfully established, the mailing audit approach is limited by the resources and costs involved in transporting equipment to and from each centre. As the measurement is the responsibility of the local physicists, phantom and dosimeter set-up errors can result in measurements out of tolerance and therefore, the need for repetition. In site-visit audits, on the other hand, experts from the auditing site travel to each centre and perform the measurements themselves. This method reduces set-up errors and increases the consistency of measurements. On-site audits have the opportunity to immediately study the issues and exchange knowledge in a wider discussion opportunity. However, they can be expensive, time-consuming and logistically difficult to perform [89]. To reduce the audit cost, the majority of regional groups in the UK perform the audit by a “round-robin” method, in which all participants play the role of visiting and hosting centres following a pre-defined plan [90, 91]. This method however provides less consistent results from each measurement. An alternative virtual method was proposed to perform the audit remotely. It reduced the audit cost and increased efficiency but was not able to analyse all centres’ data due to diverse planning and delivery systems [92].

Different levels of external audits may be used to verify absolute dosimetry of radiation sources, accuracy of the planned dose and/or accuracy of the delivered dose distribution among participant centres. Depending on the complexity level, dosimetry audits are conventionally classified into three categories: ‘Reference audit’, ‘TPS planned audit’ and ‘end-to-end audit’.

‘Reference audit’ is performed in a reference condition while the other two audits are undertaken in a non-reference condition.

Reference audit: ‘Reference audit’ is a standard external dosimetry at a reference point to verify the output calibration of the linac/radiation source. In this audit, absolute dose is measured in a single point and compared with a reference dose at the auditing site [72, 93, 94]. Some non-reference services on the beam axis/off-axis points were offered to extend the audit scope to more complicated measurements [70]. The services include a horizontal bar allowing TLD off-axis positioning and more reference depths, e.g. 10 cm. The IROC has proposed two sets of three TLD capsules embedded at two depths in a Perspex phantom to characterise electron beams (Figure 1-6) [71].

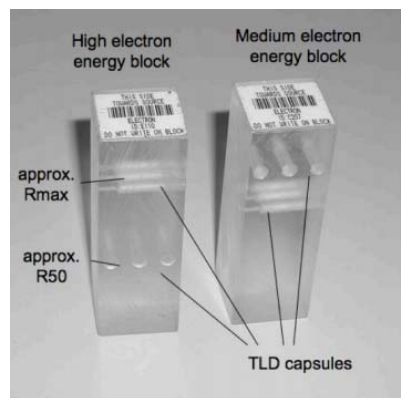


Figure 1-6- RPC Perspex holder to position TLDs for measurement of electron beam dose. TLDs are placed at approximately maximum dose depth ($d_{\max}$) and 50% of $d_{\max}$ [71]. [Thermoluminescent dosimeters: TLD]

TPS planned audit: ‘TPS planned audit’ measures dose in a non-reference condition and it mainly assesses the TPS dose calculation [62]. It measures the 2D dose using a film or detector array in a simple physical phantom which may include inhomogeneity [95, 96]. This audit is based on the original dosimetry design for IMRT QA. ‘TPS planned audit’ is an easy tool to characterise beams by capturing a large amount of data in a single exposure of the 2D arrays.

End-to-end audit: ‘End-to-end’ audit is the most comprehensive audit to verify the whole treatment chain, from diagnostic imaging to the planning and delivery system [97]. It involves dose measurements inside an anthropomorphic/semi-anatomic phantom embedded with dosimeters, TLDs, ion chambers or radiochromic films. The current audits mainly work within ICRU50 recommended criteria, which is (-5, +7) % of the prescribed dose with a spatial

accuracy from a few to less than a millimetre depending on the treatment site. Schematic of a head and neck phantom is shown in Figure 1-7.

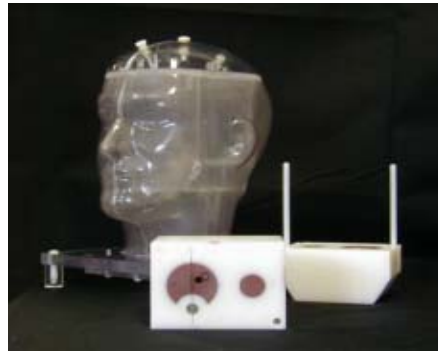


Figure 1-7- The IROC phantom and inserted dosimeters [98]. [Imaging and Radiation Oncology Core: IROC].

The three categories of dosimetry audits and their general features are summarised in Table 1-1.

Table 1-1- The levels of external audits and features of each level. [Thermoluminescent dosimeters: TLDs, optically stimulated luminescence dosimeters: OSLDs, treatment planning system: TPS].

Audit level	Condition	Detector type	Verification	Phantom	Mode
Reference	Reference	TLD, OSLD	Point dose	Homogeneous	Postal/on-site
TPS planned	Non-reference	Detector arrays	TPS calculation	Physical inhomogeneous/homogeneous	Postal/on-site
End-to-end	Non-reference	Ionization chamber, film	Whole treatment chain	Anthropomorphic	Postal/on-site

Challenges for dosimetric auditing

Regardless of the method and level, a dosimetric auditing may involve any of the challenges summarised below.

1) Cost and coverage: On-site audits provide a high coverage with accurate, consistent and reliable data and in case of non-optimality, an immediate support is undertaken saving time for optimal delivery. Nonetheless, cost of infrastructure, training and employing the experts makes this audit quite expensive. The mailing method on the other hand is less expensive, though the number of participants is limited by their data capabilities and ability to provide uniform data. In this method, a large portion of centres could not be analysed centrally due to high variability of the techniques in different centres [99, 100]. This method also requires special packaging for

device transportation and backup systems for any breakage or loss, making the audit more expensive.

2) Random/systematic errors: Another important factor in dosimetric auditing is to reduce random and systematic errors. A site-visit audit will have less errors as it is performed by one person/group. The mailing method however is more likely subject to human errors in the phantom/chamber positioning and/or local physicists misunderstanding of the protocol or the protocol ambiguities [101]. In a study by Ibbott et al, many centres failed to meet the audit defined standards and 70% of participants showed discrepancy in their dosimetry parameters with an ionisation chamber in a water phantom [98].

3) Action levels: Collection and interpretation of the auditing data should be quantitative, reflective of the real outcome, dependent on achievable outcomes, clear and understandable. Conventionally, a dosimetry audit compares a measured dose with a reference dose. In the case of planar doses, the comparison is mainly performed by a gamma function with a defined criteria including uncertainty. For an acceptable range, the criteria needs to be clinically meaningful and the uncertainty predictable as much as possible. However, the gamma results are not always reflective of real outcomes and they do not demonstrate a clear correlation with clinical meaning. The uncertainty in the acceptable range is also reflected in the action level. If the results are not within the acceptance range, conventionally a second audit is required to identify the error source and take the required action. The action level however is not always straightforward, and it may vary from critical reviewing of relevant physical and clinical parameters in the treatment workflow to measurement of physical parameters of the machine, imaging and/or collimation systems. The action may be followed by demand for manufacturing, logistics and maintenance.

EPID-based dosimetric audit

This research presents an innovative approach for remote dosimetric auditing of clinical trials. The novel concept uses the 'TPS planned audit' model and pre-treatment images from electronic portal imaging devices, EPIDs. The EPIDs are available on most linear accelerators. They have been making their way into machine specific QA [102-105] and patient specific QA [106-112]. The EPIDs provide a consistent system for data acquisition, while their measured data is easily transferred through the Cloud. This research uses EPIDs for a standardised measurement and analysis process combining the cost and efficiency benefit of remote audits. The approach is termed the Virtual Epid Standard Phantom Audit (VESPA), based on a model converting the

images to dose onto a virtual phantom. The VESPA audit has been used to analyse data from centres in Australia and New Zealand for the Trans-Tasman Radiation Oncology Group (TROG).

Before this thesis study, a patient-specific QA method existed at Calvary Mater Newcastle Hospital which used pre-treatment images from aS1000 EPIDs for local Varian Clinacs deliveries. The image signals were converting to dose using an in-house developed model. Though benchmarked, the method was developed for limited local QA measurements. To use the digitalised method for remote auditing of the VESPA, a reliable extension of the measurements, modelling and their feasibility are required. Different structures of the EPIDs, presence/absence of the EPID arm and different architectures for the linacs should be considered in the method. Furthermore, measurement outcome from remote facilities can model statistical significance of explanatory variables for the audit and it can present a comparative study of the VESPA outcome with conventional auditing methods.

Thesis Aims

A remote audit could significantly reduce the audit cost. In a well-designed audit, all participants have a consistent detection system capable of providing unified data while minimising the number of errors. Using planar dosimetry of the delivered dose also enables capturing a large amount of data in a single exposure, which provides an easy tool for beam characterisation. This thesis introduces and applies images from EPIDs to remotely audit radiotherapy facilities. The approach involves a consistent and automated system for data acquisition and analysis.

The primary contributions of the project would be as follows:

- a) Is a vendor specific dose conversion model required for each EPID system?
- b) How could the virtual method be used for remote auditing of radiotherapy clinical trials?
- c) How feasible is the remote approach?
- d) What variables have contributed the most to the auditing results and how does the auditing outcome compare with other methods?

The following aims will enable the achievement of the contributions described above:

- 1- Study dosimetric characteristics for aS1200 EPIDs and develop a model converting pre-treatment images from the EPIDs to planar dose inside a virtual flat phantom, detailed in Chapter 3.
- 2- Evaluate dosimetric differences of Varian and Elekta systems and assess the audit need for a vendor specific model, detailed in Chapter 4.
- 3- Design a remote audit for a defined IMRT/VMAT clinical trial, detailed in Chapter 5.
- 4- Conduct a pilot study to assess the method feasibility, as detailed in Chapter 6.
- 5- Perform the audit for several IMRT and VMAT deliveries at centres in Australia and New Zealand. Study the contributions of involving variables at deliveries and measurements and compare the results with the more resource intensive audits. The details are explained in Chapter 7.

Literature review and research design

Literature review

Dosimetric Auditing methods

Auditing of complex planning and deliveries, e.g. 3D CRT, IMRT and VMAT, can be performed by 'TPS planned' or 'end-to-end' methods to reduce the risk of non-reliability and non-integrity of different groups against relevant written standards. Below are some regional/international studies from both audits.

TPS planned audit: The ACDS conducted several studies to remotely check planning of treatment systems by sending a phantom CT to the facilities. The phantom had a 'slab' geometry with selected points and planes for determining absorbed dose to water. In 2013, after auditing simple deliveries, e.g. reference fields, asymmetric fields, and wedged fields, J. Lye et al audited 3D planning for 6 MV photons with a static gantry [113]. They acquired passing criteria of $< 3.3\%$ for 3D CRT and, $\gamma < 1$ at $3\%/3\text{ mm}$ for $> 97.5\%$ of points for IMRT and VMAT. The pass rates between 90% and 97.5% required an action and $< 90\%$ was considered out of tolerance. For 24 audits, 63% of the facilities passed and 33% were out of tolerance. The remainders were not assessable. They found a systematic issue with modelling asymmetric 60 degree wedges which caused (5-8) % overdose deliveries, resulting in a large portion of the facilities failing. J. Lye et al used the TG119 'C' shaped target plus additional diagnostic tests. They measured the plans using a PTW Octavius 1500 array in a solid water and lung slab phantom. The array was positioned stationary in a single plan and flipped to measure the posterior beams. Average dose planned by Eclipse resulted in the lowest variance in the central axis of the low dose region [113].

Studies by Clark et al have targeted IMRT, VMAT and Tomotherapy [114, 115]. They conducted the audits to verify TPS modelling and/or treatment delivery. For rotational treatments, Hussein et al developed a methodology for using a commercial detector array for the dosimetry and suggested using the detector over other conventional dosimetry systems such as film, ion chambers and alanine [95]. They used benchmark CT data sets and planning instructions to produce local treatment plans. The first plan, known as 3DTPS, was a generic one to compare all involved VMAT and Tomotherapy techniques. It was on a virtual phantom with pre-delineated volumes. The second plan was a selection from three clinical sites of prostate and pelvic nodes (PPN), head and neck (HN) or breast. Both plans were part of the credentialing programme to participate in the trial when the auditing team visited the sites. During each site-visit, the plan was transferred to CT data sets of the audit QA phantoms or 2D/3D detectors, then the dose distribution was calculated and compared to perform measurements [114, 115]. They measured point dose differences and global gamma index in

regions corresponding to PTVs and OARs. They determined mean gamma pass rates at 3%/3 mm, 3%/2 mm and 2%/2 mm criteria for all points in the 3DTPS and clinical plans. They also calculated the percentage of planes achieving at least 95% of gamma < 1 at the three criterion. Out of 43 delivery and planning combinations of the facilities, 34 achieved all measured planes with more than 95% of pixels passing gamma < 1 at 3%/3 mm, while this number rose to 42 for clinical trial plans. A statistical significant difference was observed between the TPS systems designed for the manufacturer's own treatment delivery system and those designed independent from the delivery system. Clark et al achieved more accurate delivery when using the former design [116]. However, this statistical result could be biased due to insufficient combinations of TPS-linacs. These measurements had a significant involvement of the audit team and site visits. The involvement reduced any lack of understanding of the protocol as well as human error, though it significantly increased the cost of the audit. Another limitation of this audit is that no clear assessment for heterogeneous tissue was available as the used phantom was homogeneous. Finally, the 2D verifications in two vertical sagittal and coronal planes did not necessarily represent the results for 3D measurements.

The EORTC, in conjunction with IROC, conducted a remote audit termed Virtual Phantom Project (VPP). The study used the IROC's anthropomorphic head phantom for IMRT credentialing. Retrospectively, participating institutions sent CT data sets of their institutional phantom (INST_{PH}), measured 2D matrices and planned dose distributions to an EORTC uploader. The institutions used the same treatment plan of the IMRT credentialing by IROC. They used a Delta4, 2D array, portal devices or film as measurement tools. The EORTC team compared the measured and calculated data of each centre using Radiological Imaging Technology software 113 (RIT, V, 5, 2, Colorado Springs, USA). Meanwhile, the IROC analysis performed the comparison using in house software. The comparisons used global gamma index evaluation at 3%/3 mm, 5%/5 mm and 7%/4 mm, and the normalisation point was considered as the maximum measured dose for the EORTC analysis and prescription dose for the IROC analysis. Among the facilities, 33% of the institutions could not be analysed centrally due to the variation of employed techniques and dosimeters. At 5%/5 mm criteria (90% pixel passing), the IROC and EORTC analysis showed respectively 92% for 11 centres and 100% for 12 centres. The corresponding pass rates were 17% for 2 centres and 75% for 9 centres at 3%/3mm. Results of the gamma indexes from the IROC and EORTC methods were compared using the Wilcoxon signed ranks test [99]. They showed $p = 0.29$ and $p = 0.01$ differences at respectively 5%/5 mm and 3%/3 mm.[99, 100]. The significant difference between the results of two methods at more stringent criteria, e.g. 3%/3 mm, suggested that further investigation was required to allow IMRT credentialing for the trial using the EORTC

method. Another issue was that the IROC method assessed the whole QA process, from CT calibration, data transfer, dose calculation and dose delivery accuracy to an anthropomorphic phantom, while the EORTC mainly verified the TPS planned dose to a homogeneous phantom. Moreover, a specific gamma evaluation for EORTC was required to unify gamma analysis of different planar tools and an equal sensitivity was required for error detection of different commercial QA systems. Measurement precision of the institutions in respect to their phantom and measurement procedure was not known, as it was dependent on the equipment and staff expertise. This might result in false negative/positive credentialing results. Finally, the data transfer of a high percentage of the institutions were incompatible. If all mentioned issues were addressed, the EORTC method could provide a reliable, inexpensive method for the auditing of clinical trials. Using the VPP method and the electronic data transfer, there was no need for any site visits.

End-to-end audit: In addition to auditing the TPS calculations of the institutions and their QA procedures, several end-to-end audits have been performed to assess the complete treatment chains of the institutions [117, 118]. Molineu et al initiated the institutions credentialing for complex clinical trials, including IMRT and VMAT, with a questionnaire assessing their understanding of the protocols and their capabilities. They mailed a head and neck (H&N) phantom to institutions and assessed the institution's irradiation of the phantom [83]. The phantom was the IROC's phantom, an anthropomorphic phantom designed in 2000 in collaboration with medical physicists from RTOG. It included two PTV structures and one OAR with embedded TLD and film dosimeters in the PTV. The phantom was treated like an actual patient by institutions, with the whole treatment chain from imaging to dose delivery being applied to the phantom. The TPS should cover at least 95% of the primary and secondary PTVs with respectively 6.6 Gy and 5.4 Gy, and the dose to OAR should be less than 4.5 Gy. Then, the phantom was returned to the RPC for analysis. Pass/fail criteria were 7%/4 mm, i.e. 7% for the TLD in PTVs and 4 mm in the high dose gradient region between the PTV and OAR, and the results were reported for all-inclusive variables for planning and deliveries. The 7%/4 mm criteria was based on the study results on the first 10 institutions so that 90% of them met the criteria. Further analysis however did not follow this percentage. In this study, the phantom was irradiated 1139 times by 763 institutions within ten years. 929 of the irradiations passed the criteria (81.6%), 156 failed only at the TLD (13.7%), 21 failed at the film (1.8%) and 33 failed at both the TLD and film (2.9%) [83]. Out of the 210 failures, 30 were due to gross setup error which may be less likely to happen in a site visit audit. At $\pm 5\%$ criteria for TLD regions, 31% of the irradiations failed. Highest pass rates were reported (90-93)% for Varian-Eclipse

and TomoTherapy-HiArt. At 5%/4 mm, the pass rates dropped to (54-79)%. A highly precise gamma index for analysis of the film data could increase the pass rates significantly.

Clark et al conducted an audit in the UK to verify the plan delivery for IMRT and 3DCRT of head and neck cancer and to ascertain suitable tolerances for the trials [114]. The centres underwent rigorous quality assurance before joining the trial, then were visited for a dosimetry audit. The visit consisted of treatment planning system tests, fluence verification films, combined field films and dose point measurements. For 6 centres, the differences between measured mean dose point with TPS dose for the PTV were -0.6% (1.8% to -2.4%) and 0.7% (2.0% to -0.9%), for the IMRT and CRT arms respectively. For individual fields, 94% of the IMRT fluence films passed gamma criterion at 3%/3mm and for combined fields, 75% of the films passed gamma criterion at 4%/3mm. This audit suggested 3%/3 mm criteria on individual fields and 4%/3 mm for combined fields for multi-centre head and neck IMRT trials.

A similar study to the IROC study was conducted by TROG to audit the dose accuracy of institutions' prostate IMRT treatments. They mailed an anthropomorphic pelvic phantom to 19 centres and an expert attended in all centres to carry out the assessment on site and resolve any possible discrepancy. At isocentre within the phantom, all centres delivered dose within $\pm 3\%$ of the planned dose. They used 5%/3 mm gamma criteria for film dosimetry of multiplanar dose analysis. At the coronal plane through the isocentre, the pass rates were more than 90% [119].

In a relatively similar study, the TROG in collaboration with RMIT University performed another audit on 12 institutions to assess the accuracy of their adaptive 3D CRT bladder treatments. They studied feasibility of on line adaptive radiotherapy on reducing small bowel irradiation in single institution trials. They developed a questionnaire for the facilities and created an adaptive plan based on the TPS and cone beam CTs. Experts from the auditing site also travelled to the centres to assess quality, dose and image guidance procedure of each institution. A Perspex phantom (Modus QUASAR) was used, mimicking different sizes of bladder, and the phantom dose was measured using TLDs. All participating institutions were able to generate a correct target volume in the planning exercise and positioned the bladder part of the phantom with 3 mm accuracy. All imaged doses were less than 5 cGy [120]. Figure 2-1 demonstrates the pelvic and bladder phantom used at the two TROG studies.

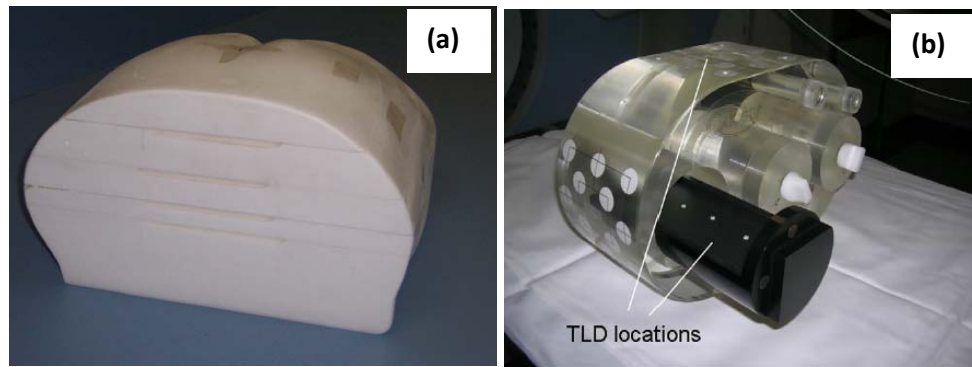


Figure 2-1- Phantoms used for TROG studies including a) anthropomorphic pelvic phantom consisting of embedded radiochromic films [\[119\]](#) and, b) bladder phantom including embedded TLDs [\[120\]](#). [TROG: Trans-Tasman Radiation Oncology Group, Thermoluminescent dosimeters: TLDs] .

Table 2-1- Summary of auditing methods for complex radiotherapy treatments including condition, results, pros and cons.

Audit	Ref	Mode	Detector-Phantom	Facility No	criteria	γ Results	Conclusion	Pros	Cons
TPS planned	[113]	Remote	Octavius array-solid water	24	3%/3mm	->97.5 %: 63% no -<90 %: 33% no -Unassessable:4% no	-Lowest variance along central axis of dose: planned by Eclipse	-Inexpensive -High pass rates	-Homogeneous phantom -Requires imitating lung movement -Large no out of tolerance
	[115]	On-site	A developed detector-Octavius II	43	3%/3mm 3%/2mm 2%/2mm	-34 no >95% for trials & 3DTPS at 3%/3mm -42 no >95% for trials at 3%/3mm - Significant difference between TPS designed for the manufacturer's own delivery system (T1) and independent designed TPS (T2)	High accuracy of plan delivery was achievable for VMAT/Tomotherapy	- Reduced lack of understanding of the protocol or human error	-Significant involvement of the audit team -High cost -Homogeneous phantom -Indirect measurement of 3D -Insufficient combination for TPS-linac.
	[99]	Remote	Delta4, 2D array, portal devices or film- institutional phantom	18	3%/3mm 5%/5mm 7%/4mm	-3%/3mm: EORTC 75% for 9 no (IROC 17% for 2 no)	-Significant difference of EORTC&IROC at 3%/3mm	-Virtual phantom -Inexpensive	-Inconsistency with IROC result -Only TPS planned assessment

						-5%/5mm: EORTC 100% for 12 no (IROC 92% for 11 no)			-Homogeneous phantom -Incompatible gamma analysis and data transfer for different detectors. -1/3 rd of centres non-analysable
End-to-end	[83]	Remote	TLD&Films- IROC's HN phantom	1139	7%/4mm	929 passed (The TPS >95% of PTV _{primary} & PTV _{secondary} with respectively 6.6 Gy and 5.4 Gy and Dose _{OAR} < 4.5 Gy.)	Highest pass rates: (90-93) % for Varian- Eclipse and TomoTherapy-HiArt. At 5%/4 mm, the pass rates (54-79)%	-large number of data - Informative for both parties -Inexpensive	Long waiting time ~3% setup error -Loose gamma criteria -18% facilities failed -Requires more precise gamma for analysis by the film
	[114]	On-site	Film/ Ion chamber- rectilinear/CIRS 002HN	6	3%/3mm 4%/3mm	-Individual fields: 94% of fluence films passed 3%/3mm -Combined films: 75% of combined films passed 4%/3mm -IMRT: D _{PTV} -D _{measured} = -0.6% -CRT:	-Proposed criteria: 3%/3mm for planar & 4%/3mm for combined analysis	-Low human error	-High cost -Homogeneous phantom -Indirect measurement of 3D

						$D_{PTV}-D_{measured}= 0.7\%$			
	[119]	Remote& site visit	Film-pelvic phantom	19	5%/3mm	Pass rates>90%		- Includes low failure	-Expensive
	[120]	Site visit	TLD-Modus QUASAR	12	3mm	All participants generated a correct target volume in the planning exercise & positioned the bladder with 3 mm accuracy and all imaged doses<5 cGy		- Standardised -Consistent -Comprehensive -Assessing adaptive radiotherapy (movement)	-Time-consuming -Expensive -No staff assessment

EPID

EPID structure

Electronic portal imaging devices (EPIDs) are a supplementary part of modern linacs. Every modern linac is equipped with this two dimensional megavoltage imaging device attached to the gantry base through a supporting arm. This project utilises EPIDs as a unified detection system for auditing the deliveries of facilities. Therefore, the structure of EPIDs is explained in this section. EPIDs can provide real-time data during/before treatments and acquired images from EPIDs are easy to process and archive [121]. Three types of commercial EPIDs are camera based, liquid filled ionisation chamber (LFIC) and amorphous silicon (a-Si) EPIDs.

Camera based EPIDs, also known as video based or fluorescent based EPIDs, initially were described by Sven Benner et al, then became commercialised in the late 1980s [122]. High energy photons of x-rays interact with a metal sheet detector, (1-1.5) mm, and produce high energy electrons which interact with the phosphor screen in turn, producing several optical photons. The light is reflected 45° and is captured by the camera. The camera is connected to a computer system converting the video signals to digital format frames for further analysis [123]. The image quality is degraded and its contrast decreases by the metal absorbed scattered electrons. The electrons are low energy scattered electrons from the gantry head, patient and couch. These EPIDs suffer from poor image qualities due to capturing a small portion of the emitted light by camera. Light scattering inside the phosphor screen and scattering and reflection from the mirror also reduces the image contrast [124].

The liquid filled ionisation chamber (LFIC) EPID was developed by Meertens and Herks in 1985 and commercialised in 1990. It contains 256×256 ionisation chambers filled with an organic liquid fluid and covered by electrode plates. A plate of plastoferrite layer converts the incident photons into electrons. A sequential high voltage is applied to the electrodes and a generated signal is measured for each electrode. The image acquisition is (0.6-2) s. Though these EPIDs are compact and light, their signals are dose rate dependent [125]. At the start of radiation, formation of ion pairs increases over ~ 0.3 s, then it saturates at equilibrium. The LFIC EPIDs also require higher doses to generate images compared with other EPIDs. They require a gap time between image acquisitions to avoid recombination [126].

Amorphous silicon (a-Si) EPIDs, also known as active matrix flat panel imagers, are the most advanced EPIDs in the field of megavoltage imaging, invented in 1987 [127]. The quality of images from a-Si EPIDs is superior to the previous types of EPIDs. It compacts the detector layer in the vicinity of the scintillator screen [122]. The main components of an a-Si EPID are

the x-ray convertor, fluorescent screen, diode array, electronic data acquisition system and computer system.

X-ray convertor: The x-ray convertor is a thin, $\sim 1\text{mm}$, metallic plate to which the incident x-ray photons interact and release Compton electrons. The metal is usually made of copper or steel which produces high energy electrons and provides a dose build-up medium for the incident beam. The x-ray convertor also absorbs scattered electrons/photons from the gantry head, patient and patient support tools to increase the quality of the image [128].

Fluorescent screen: The Compton electrons are absorbed in a phosphor scintillator screen and optical photons are released. The phosphor is mainly doped with a rare earth material and it has a thickness of $\sim 0.4\text{ mm}$ in both Varian (aS500 and aS1000) and Elekta iViewGT EPIDs [129].

Photodiode array: The photodiode array is a large 2D array of amorphous silicon on a glass substrate. The array is pixelated and consists of a number of electric circuits. Each pixel is a capacitor which stores charges that are produced due to interactions with light photons (electron-hole pairs). The capacitor is coupled with a thin film transistor (TFT) to control the signal readout. The TFTs are controlled by a control line row-by-row and they show conductivity when readout. The control is performed by applying reverse bias voltages to change the voltage of the control lines. Different EPID designs present different numbers of pixels. For example, Varian aS500 and aS1000 EPIDs have arrays of respectively 384×512 and 768×1024 pixels [47, 122].

It is worth note that the array and TFTs are made highly resistant to radiation damage, $>10^4\text{Gy}$ per year. Figure 2-2 represents a schematic of the internal structure of an a-Si EPID including photodiode arrays.

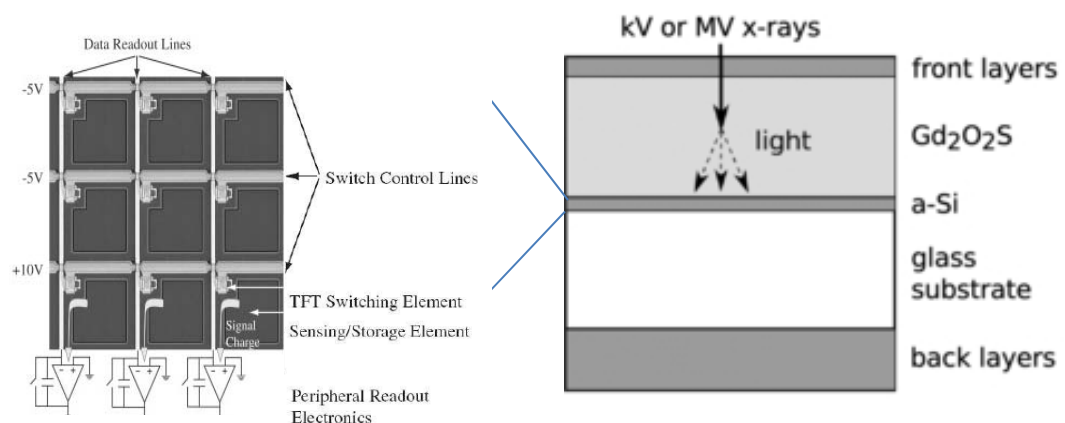


Figure 2-2- Schematic of an a-Si EPID and 3×3 pixel arrangement from the flat panel a-Si array [130, 131].

Electronic data acquisition and computer system: The array electronics are connected to an electronic data acquisition system which receives the signals pixel-by-pixel and forms the image. The electronic acquisition system can be operated by integrated or continuous (cine) mode. The former takes a single image by averaging over several frames following an irradiation while the latter takes consecutive frames during the radiation delivery. For Varian EPIDs, the acquisition system has options for “low dose” or “high quality” imaging modes, where images are acquired by averaging 2 and 4 frames [127].

The image is processed by a computer system connected to the electronic acquisition system. The computer system applies a gain correction for pixels to reduce pixel-by-pixel variations.

2.2. EPID acquisition modes

A single scan of all pixel rows makes a frame. Depending on the dosimetric application, acquisition modes of the images are selected as either integrated or cine. For integrated acquisition, all image frames during an acquisition are summed and a single ‘integrated’ image is recorded and returned to the user. The integrated image is pixel averaged over all constituting frames [132]. Then, multiplication of the frame-averaged image by the number of frames obtains the integrated pixel values [133]. Another acquisition mode is cine or “movie” mode, where images are recorded at a fixed time interval, thus creating a sequence or movie of images captured during an irradiation. In cine mode, the individual frames can be saved for post-delivery analysis. Cine mode was not primarily designed for dosimetry, since clinical implementation of VMAT deliveries was introduced around 2009 when integrated acquisition modes were already established for static delivery dosimetry. Unlike integrated mode, there are a few cine options available on commercial linacs that vary depending on the vendor and type of linac. For the Varian Clinac-Series IAS3 EPID, the clinical acquisition software returns a series of frame-averaged images rather than a series of individual image frames. The user controls the frame rate of up to 10 Hz for recording each image. A higher frame rate can be provided at the cost of reducing the resolution from 1024×768 to half the value of 512×384 . The limitation of this system is that two partial frames at the start and at the end of the acquisition are discarded. This effect reduces the signal to MU ratio in clinical images, with the reduction effect becoming significant at small MUs. The signal reduction should particularly be corrected when calibrating signal to dose. Another limitation is buffer overflow which can occur for too many frames. It can be reduced by operating the EPID in half resolution mode

with a small source to detector distance for the EPID, though the increased signal may result in image saturation. The cine acquisition mode for the Varian aS1200 and Elekta iView EPIDs provide normalised data for storage, so the dosimetric information is lost. However, for the former, an ‘image processing service’ is provided that saves individual frames in DICOM format. For Elekta iView EPIDs, Perkin-Elmer service software (XIS) can be used to obtain frames of deliveries in a single image file excluding the information on gantry angles. The angle information is particularly important for sag correction and dose verification of VMAT deliveries. An image acquisition software with iView EPIDs has been developed by Mans et al. to acquire image frames at 2.5 Hz [134].

2.3. Dosimetry properties of EPIDs

In addition to the initial purpose of EPIDs design for patient positioning, they are a proper candidate for quality assurance measurements and dosimetry purposes. Dosimetric characteristics of the EPID are required to be determined, similar to any other detector, to attribute an accurate dose. Main technical factors in using the images for dosimetry purpose include ghosting and lag [135], linearity, reproducibility [135, 136], dead-time [137], build-up factor [138], optical glare [139], sag, arm backscatter, frame dose distribution and delivery angle inclusion.

Ghosting and lag effect: The electron-hole pairs within photodiodes of a-Si EPIDs impacts the pixel sensitivity (ghosting) and the memory (lag) [140]. Both effects are due to charge trapped in defect energy levels. Charge trapping is mainly produced by three mechanisms: 1) recombination of trapped charge with a charge from a subsequent image, 2) generation of new traps by x-ray exposure and, 3) modification of the electric field distribution within the photodiodes. The three mechanisms usually result in a reduced sensitivity [140]. Less important sources of image lag are incomplete charge transfer and phosphor after-glow [141]. Several studies have demonstrated image lag by exposing the EPID with a small field, immediately followed by another exposure with a larger field. Greer and Popescu [133] and Van Esch et al [132] measured a small effect of image lag (less than 1%). Winkler measured image lag as a function of time and ratio of MUs between the first and second image [142]. They demonstrated larger image lag effects of up to 9% with minimum time between irradiations and maximum ratio of MUs. McDermott et al measured image lag by monitoring image signal as a function of time after exposure [143]. Image lag has been compared for indirect and direct detection EPIDs and found to be approximately equal [141].

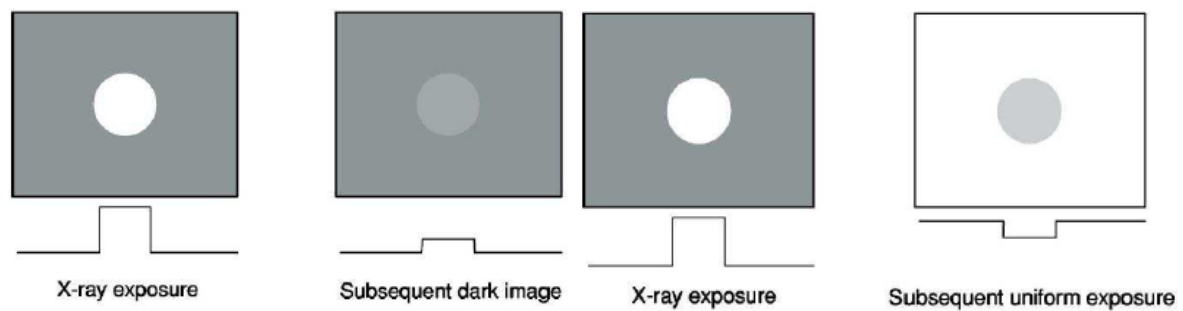


Figure 2-3- Diagrams illustrating lag and ghosting concept. a) Lag: Images acquired immediately after an x-ray exposure show an increase in pixel values in areas of previously exposed, b) Ghosting: reduction in pixel sensitivity due to previous exposure to radiation when exposed to subsequent irradiation. [140]

Linearity: Linear response versus dose/dose rate has been established for the integrated mode of aS500 and aS1000 EPIDs [132, 144, 145]. The nonlinearity, however, has been introduced as a general characteristic of a-Si EPIDs [133, 146]. The nonlinearity effect was specially observed in small monitor units (MUs) and some particular imaging modes. The majority of nonlinearities in EPID response at low MUs originates from incomplete signal capturing by frames at the beginning/end of the image acquisition and/or image lag and ghosting effects [140, 144]. The nonlinearity due to the incomplete frames can be significantly removed by manufacturer acquisition improvement, as IAS2 was upgraded to IAS3[147, 148]. McCurdy and Greer observed a small but almost constant amount of missing signal in cine acquisition mode which had only a significant dosimetric effect for low MU irradiations ($\sim <100$ MU) [149]. These MUs were much less than typical VMAT deliveries. The nonlinearity did not have a large effect on dose measurements when typical treatment doses were integrated. However, if small MU images of each segment were acquired, e.g. for calibrating signal to dose, a nonlinearity correction was required. In a study by McDermott et al., the EPIDs showed under response for the first 5 MU by about (3-5) % compared to the response for 1000 MU. They attributed the under response to trapped charge in the photodiodes [146]. The under-response in dose/dose rate has been determined by lowering the dose per pulse using the linac pulse-repetition frequency [142, 143] and by moving the detector further from the source [143]. McDermott et al. and Winkler et al. also studied the dosimetric performances of Elekta iViewGT EPID systems operated in integrated mode [142]. McDermott et al. observed some nonlinearities correctable $\sim 1\%$ by using a 5 mm copper buildup plate and a time-dependent ghosting correction factor [143]. Winkler et al. observed about 7% nonlinearity versus dose rate, attributing the variation to a dose-per-frame effect. These variations were reproducible and could be corrected with a custom calibration to recover the linearity per EPID and to improve the dosimetric accuracy [142].

Reproducibility: The short and long term reproducibility of a-Si EPIDs has been demonstrated for all EPID vendor models [150] [142]. Varian EPID systems have shown long term reproducibility of $< 1\%$ for all pixels over a three year period [151] and short term reproducibility of $< 0.5\%$ using central axis measurements. The Elekta EPID system demonstrated $< 0.5\%$ for all pixels over nearly two years [152]. Significant variations were observed in dose response of different EPIDs of the same model and in the dose response of individual EPIDs over time, particularly in the first weeks of operation [153]. They suggested characterisation of an EPID specific dose response and an EPID QA program to maintain suitable accuracy in EPID dosimetry. An inter-comparison of 11 Varian EPIDs also found differences in response between EPIDs. Another study showed that the dosimetric differences with more advanced model EPIDs had decreased [148].

Dead time: The limitations in image acquisition electronics may cause dose response errors [133]. The Varian imaging system showed a dead time after each 64th frame when the image data was transferred to the CPU. This was due to the dynamic range of the 14 bit A/D converter. During time of this transfer no signal was lost, but a “reset” frame was applied following the transfer. During the “reset” no signal was recorded. Though the “reset” frame has been removed, the frame integration time, which is twice as long, can result in saturation of EPID signal at sufficiently high dose rates. To avoid this issue, lower dose rates or larger source to surface distance (SSD) was recommended [132]. Reducing the gain of the EPID for dosimetry acquisition modes at higher dose would also reduce the dead time. Saturation problems have not been reported on EPIDs of any vendors other than Varian. The reported signal loss with the Varian EPID during continuous acquisition was attributed to missing signals at the end of the irradiation [149]. The dead time for open field irradiations is easily corrected as it produces a small uniform signal loss across the image. For IMRT beams, particularly sliding window delivery, the dead time correction is important and effective on local signals since the delivered dose is spatially and temporally dependent, and dose may be delivered to a particular spatial location for only a small time period.

Buildup factor: The copper layer present in the EPID introduces an inherent buildup. Any buildup placed onto the EPID reduces the low energy photons reaching the EPID phosphor screen and ensures electronic equilibrium. The buildup is more important for transit than non-transit dosimetry as in transit dosimetry, an additional low energy is scattered from the object/patient. However, the buildup in non-transit dosimetry should be accounted for in high energy irradiations (18MV) as it can attenuate head scattered radiation and influence the EPID field size response [154]. Figure 2-4 demonstrates the effect of buildup layer thickness and

material on the EPID field size factors at two energies of 6 MV and 18 MV. At the lower energy, addition of a copper layer introduced a small effect on attaining equilibrium. Similar effect was observed on equilibrium when using solid water slabs. Slabs of water equivalent material [133, 155] or thin metal sheets [142, 143, 156] were used to introduce a buildup factor in different studies. The idea to determine sufficient buildup for transit dosimetry is to modify the air gap between the phantom and EPID until a consistent EPID response is achieved.

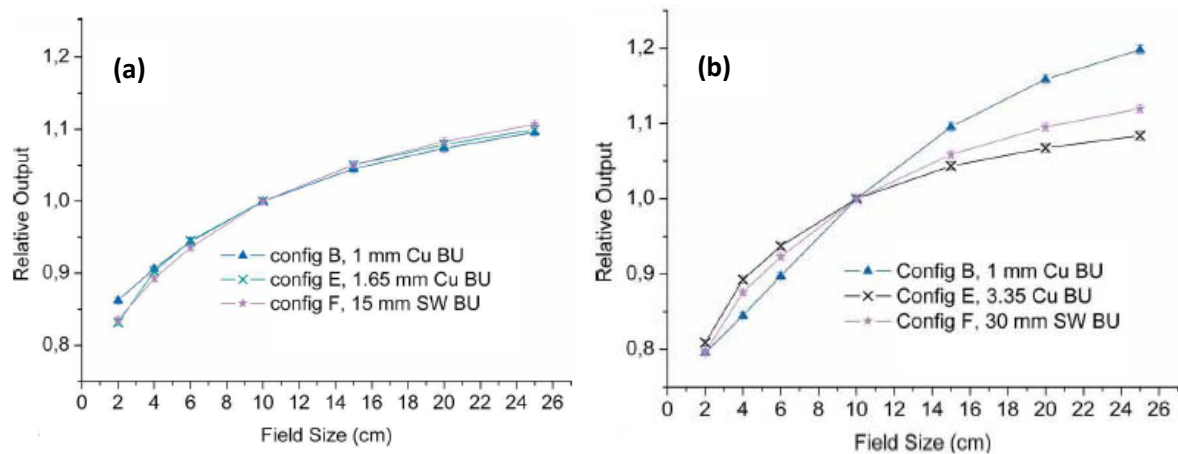


Figure 2-4- Effect of buildup layer thickness and material on EPID field size factor at a) 6MV and, b) 18MV irradiation [154].

Optical glare: Deposition of radiation energy onto the EPID releases optical photons from the phosphor layer. The photons are detected by photodiodes then converted to electric charge. Phosphor is a translucent material which causes photon scattering and submillimeter diffusions over time. This phenomena is called optical glare, which causes blurring of the deposited dose pattern. This effect is more severe in camera based EPIDs wherein the optical photons experience multiple scattering between the screen and mirror [157]. The scattering in the phosphor layer of a-Si EPIDs is quite small as the layer is coupled to the photodiodes and there is no gap between them. Munro and Bouius found negligible glare in their experimental study [158]. McCurdy et al applied their experimentally determined glare kernels to their portal dose prediction model and found improvement in out of field areas [159]. Kirkby et al also used a glare kernel to improve the accuracy of their Monte Carlo model for fluence prediction. They discovered the necessity of using a 1 cm water slab downstream for accurate modelling, which was equivalent to the effect of the glare correction [160]. Gustafsson et al observed different field size factors and penumbras in absence of the phosphor layer, though their EPID response could be attributed to energy dependency response rather than optical scattering [154].

Sag: EPIDs are extended outwards into the treatment beam from the main gantry of the linac. This causes sag (mechanical flexion) of the imager from an ideal central axis alignment due to gravitational force. The sag introduces a small shift in location of the EPID image as a function of gantry angle. For Varian E-arm systems, a shift of ± 1 mm in-plane and ± 0.5 mm cross-plane has been reported [161]. The reported shift for Varian R-arm systems was significantly larger than the E-arm systems, ~ 10 mm [162]. Mans et al. and Rowshanfarzad et al. reported ± 2 mm for Elekta EPIDs [134] [163] while Poludniowski et al. reported ± 4 mm in the in-plane for Elekta EPIDs [164]. Submillimeter accuracy was achieved for the newer model of Varian EPID, aS1200 EPIDs, using a pre-measured sag calibration function. Most studies measured combined EPID sag and gantry wobble of the linac, though different approaches have been suggested on separation of both components [165, 166]. One method involved positioning a ball bearing at/close to the linac isocentre, then taking images at discrete angles/cine images. In this method, the developed algorithms for marker detections could not measure sag along the beam axis [167, 168]. Bakhtiari et al used jaw-defined square fields to irradiate different centre positions on the EPID at different gantry angles. Their method however was under the impact of jaw sag. [169, 170]. Another work studied the modification of leaves' position due to the gantry rotation [171]. The direction and magnitude of sag have been shown to be consistent and reproducible with gantry angle, therefore the sag can be corrected once the relationship is mapped [52, 103].

Arm backscatter: EPIDs are attached to the linac gantry through a robotic support arm. The design and movement of the arm is different among manufacturers. The “R-arm”, consisting of two bars attached by an axle to hold another bar, is used in Varian aS500 EPID systems. Very soon, the “R-arm” was replaced with the “E-arm”, in which only two bars firmly position the EPID for vertical and horizontal movements. However, the EPID cassette has indentations to incorporate the motion wheels and rails and the imager cabling. This non-uniform structure and presence of metallic parts contribute to additional signal, up to 6% of maximum dose, to the image from increased backscattered photons [147, 172], and have an impact on the dosimetric application of EPIDs [173]. The backscatter signal contribution is known to be asymmetrical and field size/field location dependent. In-plane motion is more complicated as it involves more junctions than cross-plane motion. The backscatter effect can be modelled using a simple backscatter kernel convolved with a portion of the incident beam impinging on the arm support components. King and Greer measured this effect using a binary mask representing the arm shape to the beam shape [174]. They optimised the estimated backscatter kernel and provided an iterative correction technique to estimate and remove the backscatter from the measured image. Other methods involved placing an additional backscatter material, e.g. 8.9 mm water

[172], or 2 mm lead sheet [175], beneath the EPID cassette and upstream the arm component. These methods resulted in a uniform and symmetric backscatter response which could be easily corrected. The aS1200 EPIDs have reduced the backscattered signal to $<0.5\%$ using a shielding material, and the Elekta iViewGT EPIDs have represented insignificant backscattered signals.

Frame dose distribution: Images are read out sequentially, e.g. row-by-row. Each row has a shifted time interval over which the dose is integrated between two intervals. If the dose pulse rate varies over comparable time intervals, different image rows can integrate different doses, resulting in large signal differences within a single frame. This problem is observed in cine mode and is due to interplay of image readout scanning and discrete beam pulsing/accelerator dose rate [176, 177]. This artefact does not appear in integrated mode as if it occurs in a frame, it is markedly reduced after averaging over frames. The EPID dosimetry systems, and also current TPSs, do not include these variations. Some algorithms have been suggested to predict and remove this artefact [176]. Frame rate modification is another suggested method to remove this artefact [177]

Gantry angle inclusion: Assigning an accurate gantry angle to each EPID image is critical for sag correction and for dosimetry purposes of VMAT deliveries. As gantry speed may vary during VMAT delivery, acquired images at fixed intervals do not necessarily present a linear correlation with gantry angle. Different methods have been proposed for this purpose. The Varian C series records the gantry angle in the header of cine image but with a 3^0 error [178-180]. The Elekta system has used an iCom connection to the linac to assign gantry angle to each frame with a measured lag of ~ 0.4 s [134]. Using a separate inclinometer placed on the linac, trajectory log files, and retracting MLC leaves and jaw positions are other suggested methods for accurate determination of the delivery angle for each image [180].

2.4. Inter-vendor a-Si EPIDs

Commercial a-Si EPIDs are available from three vendors: Elekta iViewGT (Elekta, Crawley, United Kingdom), Varian aS500/1000 (Varian Medical Systems, Palo Alto, California), and Siemens OptiVue 500/1000 (Siemens Medical Solution, Concord, California). The general principle of operation and dose calibration are similar for each vendor's EPIDs. The EPIDs from all vendors have shown non-constant signal-to-dose ratios [142, 143]. But, ghosting and lag effects have demonstrated dependency on the design and exposure time of the panel [141, 181-183]. In practice, both effects combine to impact the dose per frame readout by the detector [143]. For EPIDs from all vendors, less defect has been observed when using longer irradiation since the deficiency is integrated over all frames of the image. They all have shown nonlinearity at small MUs [146] and short and long term reproducibility [150] [142]. Signal per MU for

short irradiations were up to 10% lower than for long irradiation times. All the EPIDs demonstrated a relative under-response at low MU irradiations, (~ 4 -10) % lower than 1000 MU irradiations. They also showed a constant signal-to-MU within ± 1.5 MUs at radiations more than 200 MUs. The ratio showed a decrease of 4% for the Elekta iView EPIDs, and 5% for the Varian and Siemens panels (MC). All the comparisons presumed similar beam characteristics for various linacs and investigated the detector response variabilities. The signal loss during cine acquisition was reported with the Varian EPIDs [149]. All the EPIDs demonstrated a small optical glare due to their relatively similar structure [158]. Backscatter artefacts were observed in Varian EPIDs, but only due to components from their support arm lying underneath the EPID active area. The gantry angle for images from aS500/aS1000 has been recorded with a 3° error while the Elekta iView EPIDs have measured ~ 0.4 s lag [134, 178-180].

The pixel pitches of the EPIDs are in the range of $0.4 \times 0.4 \text{ mm}^2$, providing higher spatial resolution than most other available dosimeters. This higher resolution is significantly larger than most ionisation chambers, and also superior to patient computed tomography (CT) data sets, where voxel sizes are typically about $1 \times 1 \times 2 \text{ mm}^3$. Pitches of the EPID pixels for aS500 and aS1000 EPIDs are $0.39 \times 0.39 \text{ mm}^2$, and for iViewGT and Siemens EPIDs are $0.40 \times 0.40 \text{ mm}^2$. Active areas of the aS500 and aS1000 EPIDs are $40 \times 30 \text{ cm}^2$ and for aS1200 EPIDs are $40 \times 40 \text{ cm}^2$. The active area for iViewGT and Siemens EPIDs are $41 \times 41 \text{ cm}^2$. The corresponding pixel numbers are respectively 1024×768 , 1024×1024 , and 1024×1024 [146].

The Varian EPIDs, aS500, aS1000 and aS1200, provide images with Digital Imaging and Communications in Medicine (DICOM) format, while images from Elekta iView EPIDs should be exported with '.his' format in order to access the delivery information. The information is recorded in a separate '.log' file header along with the corresponding '.his' file.

2.5. EPID to dose conversion method

According to Van Elmpt et al, non-transmission dosimetry (or 'non-transit dosimetry') refers to dosimetry when the treatment beams have not passed through an attenuating medium to reach the EPID. If the beams do pass through an attenuating medium, the method is described as 'transmission dosimetry' (or 'transit dosimetry'). Discussion on transit dosimetry is out of the scope of this thesis. Van Elmpt et al also categorised EPID calibration methods for dose calculation into: 1) simulation of EPID grey-scale values and, 2) conversion of grey scale values to dose in water. The first method uses empirical calculations, such as Monte Carlo, to simulate the detector response. The second method is a semi-empirical method calibrating the grey scale pixel values to dose by inter-comparison with a calibrated dosimeter. Both methods can be used in pre-treatment or during treatment approaches. The second method is used and explained in

the following section as the thesis uses in-air acquired images from EPIDs for auditing purposes. For more details of other EPID dosimetry methods, the reader is referred to van Elmpst et al [47].

2D verification in phantom or planning CT

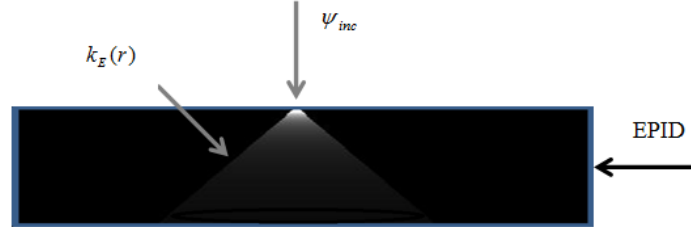


Figure 2-5- Incident fluence ψ_{inc} onto an EPID and its kernel $k_E(r)$ spread within the EPID.

Several approaches have been developed for the conversion of the grey scale pixel values of the images to dose planes within a water/solid water phantom. Available commercial software include epiQA, EPIDose, Dosimetry Check, and Adaptive. The software attempts to perform pre-treatment dosimetry using EPID acquired images to verify the delivery of the correct dose/fluence. They use models mainly derived from earlier works undertaken with camera based and LFIC EPIDs. Deconvolution-convolution is a common method for planar dose estimation using images from EPIDs. Kirkby et al initially estimated the incident energy fluence, then calculated dose at particular depths. Schematic behaviour of incident fluence onto the EPID is shown in Figure 2-5. Monte Carlo simulations were used to determine the scatter kernel of an EPID for subsequent deconvolution of images to derive the incident energy fluence [160]. The fluence was then compared to diamond detector profiles in air. The scatter kernel of the EPID included two kernels of deposition and optical scatter. The latter improved the model accuracy compared with the measured fluence. Similarly, Warkentin et al used a Monte Carlo generated EPID response kernel to derive primary fluence by deconvolution of the EPID image [155]. The fluence was then convolved with a Monte Carlo calculated dose in water kernel to calculate the dose at particular depths. The calculated dose was then compared to 2D dose distribution measurements in a solid water phantom. More recently, the kernel based method was used by King et al, summarised in the following steps [184].

-EPID deconvolution to fluence: the incident fluence $\psi_p(r)$ onto the EPID was defined as

$$\psi_p(r) = [C_A(r)D_E(r)] \otimes^{-1} k_E(r), \quad (3)$$

where $C_A(r)$ is the profile correction matrix removing asymmetries in the image around the beam axis [185, 186]. The $C_A(r)$ was calculated as a ratio of the predicted fluence for an open field to computed fluence from an open field EPID image. The predicted fluence was considered the fluence from the Pinnacle TPS based on Monte Carlo simulations [160]. $D_E(r)$ is the dose of the midplane images, and r is the distance from the beam centre. $k_E(r)$ is the deconvolution kernel, determined empirically as [185]

$$k_E(r) = e^{-(a_1 r)} + a_2 e^{-(a_3 r)} + a_4 e^{-(a_5 r)}. \quad (4)$$

To identify optimum values of a_i , $i = (1-5)$, some measured fluence profiles were used.

-Fluence to dose: the dose profile in a water phantom $D_W(r)$ was defined as

$$D_W(r) = D_{CAL} \cdot [T(r)\psi_p(r)A(r)] \otimes k_W(r) \quad (5)$$

Where D_{CAL} is a scaling value converting EPID grayscale values at each point to absolute dose [184]. $T(r)$ and $A(r)$ are radially symmetric terms and attenuation factors, defined respectively by $T(r) = 1 + b_1 r$ and $A(r) = e^{-(b_2 r)^2}$. The former corresponds to an increase in a deposited dose due to lower off-axis energy spectrum, and the latter corresponds to a longer path length due to the beam off-axis. The $k_W(r)$ is the deposition kernel of a radially symmetric dose in the phantom, defined as

$$k_W(r) = b_3 e^{-(b_4 r)^2} + b_5 e^{-(b_6 r)^2} + \frac{b_7}{r} e^{-(b_8 r)^2} \quad (6)$$

Extensive validation of the model for 2D planar dose was conducted with comparison to MapCHECK 2D planar dose distributions (Sun Nuclear Corporation, Melbourne, FL).

Nicolini et al developed a different method for calibrating the grey scale pixel values to dose plane in water. They used a series of field-size dependent empirical correction factors. The field sizes were found from individual segments of each IMRT beam, using the MLC delivery file, and they were used for the signal to dose calibration. The doses of all segments were added up to create the planar dose for the IMRT field [145].

3D dose verification using images from EPIDs

Non-transit images from EPIDs have been used to calculate dose within phantoms/patient models. Steciw used the Warkentin method to determine the fluence by deconvolution of an EPID scatter kernel [187]. Then, he used the fluence as an input into a commercial TPS for 3D

dose calculation and compared the dose and DVH with the planned dose [188]. An available commercial system performs a similar procedure; it obtains the incident fluence, which in turn is used in a dose calculation system to estimate the dose distribution within a patient CT model for comparison with the planned dose [189].

Several other 3D dose reconstruction models have been reported using non-transit EPID images [190-192]. Ansbacher derived planar dose at 10 cm depth of a virtual flat phantom (VFP), then extended it to 3D dose in a virtual cylindrical phantom (VCP) using exponential percentage depth dose (PDD) modelling and a buildup factor. [190]. However, to calculate 3D dose inside the VCP, a new coordinate system was required to include EPID-gantry rotation. Ansbacher converted the room coordinates to an EPID coordinate system, including a plane perpendicular to the beam rotating with gantry angle. Then, each pixel of the TPS dose matrix point (axial (z) slice) was projected onto the EPID midplane dose matrix. The orthogonal distance between the axial TPS dose matrix point and the EPID dose plane was considered as s , and the distance orthogonal to s in the axial plane from the cylinder axis to the projection of s onto the EPID was considered as v (see Figure 2-6). As this figure illustrates, s and v are changed by the gantry angle of θ

$$\begin{aligned} s &= Ym \cos \theta + Xm \sin \theta \\ v &= Xm \cos \theta + Ym \sin \theta \end{aligned} \quad (7)$$

Considering the beam divergence, the TPS needed to be projected in a plane perpendicular to the axial plane, the coronal plane.

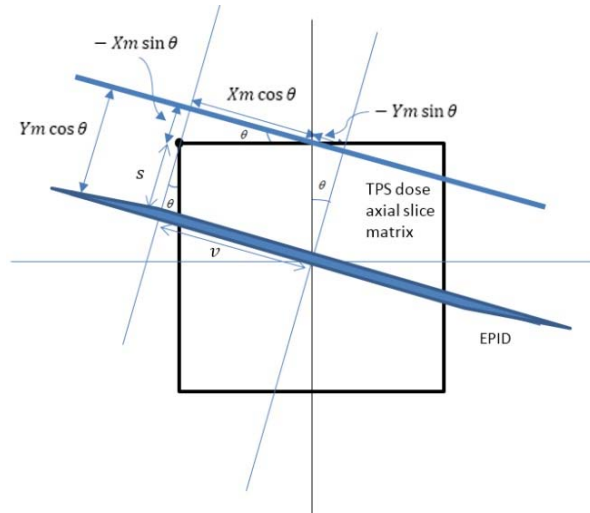


Figure 2-6- a) Illustration of variables for the new coordinate system versus gantry angle ($z = 0$).

The EPID dose (x,y) was then interpolated onto the coordinates (X, Y) to give the dose values corresponding to the projection of the TPS axial dose points. Considering n as the number of axial TPS dose slices, the software calculated the dose of the axial planes n times. This axial dose was then adjusted for percentage depth dose (PDD) and buildup factor. The method determined an approximate equivalent square field size for the IMRT field, and then determined an attenuation factor from this field size. The exponential attenuation was used to model percentage depth dose plus an exponential buildup term to model the buildup region [52]. The dose was calculated for each individual image at each gantry angle, then added to the total dose matrix to give the combined 3D dose distribution. This dose was stored in the coordinates as the TPS dose matrix so that the dose distributions could be easily compared quantitatively. To date, limited measurements have been performed to validate the 3D model, while their accuracy relies on comparisons with the Eclipse planning system calculations.

Research design: VESPA

Participating centres in the audit programme provided in-air acquired images from their EPIDs. The images are acquired from pre-treatment delivery of two benchmarking plans. In the auditing site, the images are converted to dose planes in a virtual flat water phantom in a process referred to as Absolute Portal Dosimetry (APD). These planar doses are used for field-by-field analysis compared with corresponding TPS doses. For 3D analysis, a correction is initially applied to convert the flat phantom to a cylindrical phantom, then the planar dose is converted to 3D dose using the Ansbacher method.

Images from EPIDs and corrections

A list of centres/facilities from Australia and New Zealand that are currently treating patients with IMRT/VMAT deliveries are prepared for the VESPA study. They provide images from their EPIDs, including aS500, aS1000, aS1200 and iViewGT EPIDs. Depending on the vendor, the EPID position is either at 5 cm below isocentre or at isocentre, with the EPID centred in lateral and longitudinal directions. The images from Varian EPIDs have DICOM format and those from Elekta iView EPIDs should be exported one at a time in '.his' format with the associated log file. The log file is generated with the pixel scaling information required to create integrated images. For consistency, the '.his' format images are converted to DICOM format at the auditing site. Integrated modes were used for IMRT analysis, and cine modes for VMAT analysis. For the integrated mode, any normalization is removed from the image, and the relative intensities of each segment is maintained. The integrated images are acquired in Clinical Mode/QA patient. In this thesis, the IMRT study is performed on both Varian and Elekta linacs, while the VMAT study only includes deliveries from Varian linacs.

Standard EPID image correction

All acquired images are conventionally flood-field, dark-field and pixel-defect map corrected.

Coordinate system

Two EPID images of a 10×10 cm² field with 90° and 270° collimator angles at gantry zero (gantry pointing vertically down) are acquired to determine the sub-pixel central axis (CAX) location on the EPID. An EPID coordinate system is then referenced to the radiation isocentre. The field edges (50% dose points) of each image are determined using linear interpolation between the pixels. The average mid-point of the two images gives the CAX location independent of jaws positioning.

EPID sag correction

The current study acquires EPID images of a $10 \times 10 \text{ cm}^2$ field at either 45° or 90° gantry angle intervals. Following the Rowshanfarzad et al. method, the field mid-point location is determined on each image and compared to the mid-point at gantry zero to calculate sag relative to gantry zero (where the CAX position is known) [103]. The difference versus gantry angle shows best fit with a first order Fourier series, $Sag(\theta) = a_0 + a_1 \cos(\theta) + b_1 \sin(\theta)$. This model is then used to correct the coordinate system for each acquired image depending on its gantry angle. This method corrects sag for the combined gantry wobble and EPID sag.

The conversion model calibration images

The model currently requires a $10 \times 10 \text{ cm}^2$ and $40 \times 30 \text{ cm}^2$ image [193], acquired with 20 MU for Elekta systems and 100 MU for Varian systems, for calibration purposes. The difference in MU for the two systems is related to the methods employed for IMRT image acquisition on these systems. The $10 \times 10 \text{ cm}^2$ image determines the current EPID response and corrects for drift in the central axis linear accelerator output and EPID response. The converted dose value in a region of interest (ROI) at central axis is compared to the corresponding TPS value for the calibration factor determination. All acquired images are also divided by the $40 \times 30 \text{ cm}^2$ image to account for the linear accelerator EPID off-axis response drift since the flood-field calibration. The latter division is not an essential correction.

Backscatter Correction

As previously mentioned, the support arm of aS500/aS1000 EPIDs produce some backscattered radiation in the images, and a correction is required for the backscatter artefact. The “backscatter-free” image equivalent to acquiring the image without the support arm is acquired using the King and Greer method [194]. The method uses a backscatter kernel which is convolved with the fluence to estimate the backscatter image. The fluence is derived from the image itself. The model iterates, generating estimates of the backscatter-free image and comparing the sum of this estimate and the backscatter image to the measured image until a certain agreement with the measured image is obtained. To remove backscatter from an image, the flood-field correction image is first “removed” from the image to obtain the raw image. The backscatter is then removed from the raw image and flood-field image. The corrected flood-field is then reapplied to the image to yield a backscatter-corrected (BSC) image. No backscatter correction was applied to the images from aS1200 and iViewGT EPIDs due to their negligible backscatter effects.

2D dose planes in virtual flat phantom (VFP)

The core process of the modelling is conversion of the grey scale image to dose plane in a virtual flat phantom (VFP). The developed model by King et al is used to convert the EPID signals to dose in the VFP [193]. The model can calculate dose at any depth that appropriate field-size factors and beam profiles have been measured and kernel parameters determined. Although kernels exist to calculate dose at 1.5, 5, 10, 20, and 30 cm depths, currently the VESPA uses the model made for dose reconstruction at 10 cm depth of the VFP. King et al developed the model parameters for deliveries from a prototype backscatter shielded EPID [193]. For this work, model parameters derived from images acquired with a Varian aS1000 EPID were used. Before model derivation all acquired images were backscatter corrected as described above. The model was tested using IMRT images of a prostate and head and neck patient. The method was validated by comparison of EPID images converted to dose compared to measured MapCheck (Sun Nuclear Melbourne, FL, USA) dose planes. EPID images of 36 sliding window IMRT fields were acquired on the aS1000 EPID. These were backscatter corrected and converted to dose at 5 cm depth in water. Gamma comparison using an in-house implementation of the Low method was made between those dose distributions and Mapcheck measurements at 5 cm depth in solid water with 2% of maximum dose, 2 mm criteria and a threshold of 10% of maximum global dose. The resulting pass rates for 14 prostate fields were (mean $\pm$ 1 SD) $99.4 \pm 1.0\%$ and 22 head and neck fields were $99.3 \pm 1.3\%$. Corresponding mean gammas were 0.31 ± 0.3 and 0.33 ± 0.5 respectively. This model was developed by P. Greer and was used for auditing of centres with Varian aS500 and aS1000 EPIDs and for Elekta iView EPIDs. Chapter 3 presents a model developed by the author for deliveries from the new Varian TrueBeam system, using images acquired with an aS1200 EPID for both flattening filter (FF) and flattening filter free (FFF) beams. Chapter 4 further investigates the need for vendor specific conversion models for image-based auditing. Profiles and field size factors for Varian and Elekta EPID systems are compared, along with the performance of the existing Varian model and a new Elekta model for a series of audit IMRT fields measured on Elekta systems.

3D dose distribution in virtual cylindrical phantom (VCP)

Cylindrical phantom contour correction:

Right before conversion of the planar dose in the VFP to planar dose in the VCP, an off-axis correction matrix is used. The matrix is normalised to 1 at the centre of the phantom and is derived as the normalised ratio of the Eclipse calculated coronal dose plane at the midplane of a 10 cm radius phantom for a $25 \times 25 \text{ cm}^2$ field at gantry zero to the dose plane derived with the

backscatter corrected model from the EPID image of the same field size. To encompass the 30 cm length of the EPID, the correction is extrapolated. Figure 2-7 shows an image and profile of the correction contour.

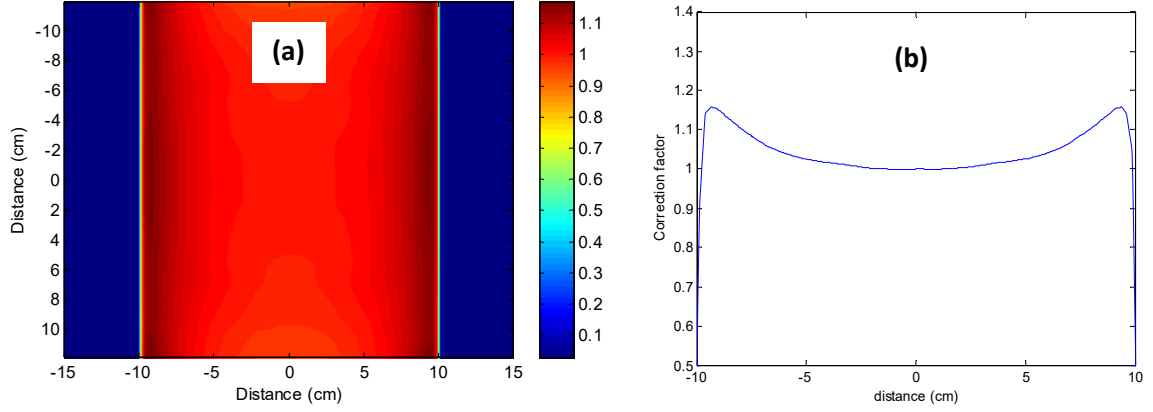


Figure 2-7- a) Cylindrical phantom contour correction image, to convert dose at 10 cm depth in the flat phantom to dose at the 10 cm depth in a cylindrical 10 cm radius phantom. b) Crossplane profile through the central axis of the cylindrical phantom correction image

3D dose reconstruction in VCP

The planar dose is converted to 3D dose in the virtual cylindrical phantom (VCP) using the method introduced by Ansbacher. In this method, each dose plane is projected along the beam raylines accounting for gantry angle using exponential attenuation. The attenuation factor is set to a single value factor for a 12.5 cm square field, which is equivalent to 0.0034 according to Table 1 from Ansbacher [190]. This is repeated for each beam, and the doses from each beam sum to give the 3D dose distribution. The employed buildup region model parameters are also found in Ansbacher [190]. Summary of the conversion of the grey scale pixel values of the image to 2D and then 3D dose is demonstrated in Figure 2-8.

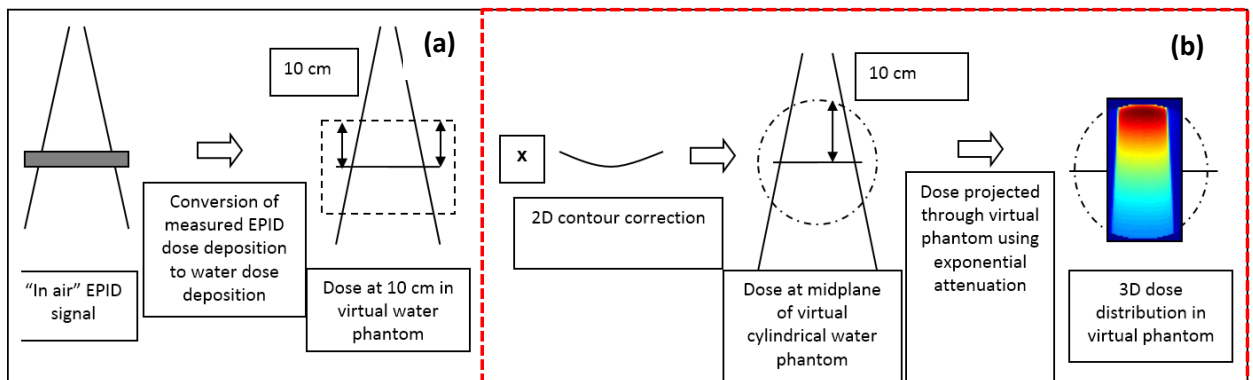


Figure 2-8- Steps for calculation of a) 2D dose in a virtual flat phantom (VFP) and, b) 3D dose in a virtual cylindrical phantom (VCP).

A graphical user interface (GUI) software for analysis

The 2D and 3D models include labour intensive codes requiring input of parameter files that list all files to be loaded for the optimisation and for testing of the model results. Therefore, a graphical user interface (GUI) software was designed at the Calvary Mater Newcastle Hospital (CMNH) for easy evaluation of deliveries by P.Greer and B.Zwan. The GUI requires the user to select images from the EPID, pre-treatment and calibration images, DCM and plan dose from the TPS, and machine parameter and model parameter files. The software does not require parameter adjustment by the facility. However, an individual machine specific file, using information provided by each facility, is used to refine the model and adapt it to each machine/delivery type. The machine specific file uses calibration images from each facility to determine the central axis coordinate on the EPID and correct sag and backscatter artefacts as described above. The GUI then loads this data and calls the EPID model code. The 2D dose planes are displayed through the reference point location, either sagittal, coronal or axial planes, and the Gamma results for the displayed plane is reported. This gamma map can be displayed separately, and a comprehensive comparison is performed between the reconstructed dose and the TPS dose. An in-house developed gamma algorithm is used for the dose comparison. All doses above 10% of the maximum dose are assessed with a search region of 6 mm radius. The gamma function uses a global dose difference (DD) criteria defined by percentage of maximum dose of each measured image. For individual fields, 2D gamma analysis is employed, while for combined dose distributions, 3D gamma analysis was used. Similar to 2D, the 3D dose assessment result is demonstrated on the GUI page. Figure 2-9 demonstrates a typical comparison for a post-prostatectomy patient using the GUI software. The left side dose planes are 2D converted dose from the images, and the right sides are the corresponding TPS calculated dose.

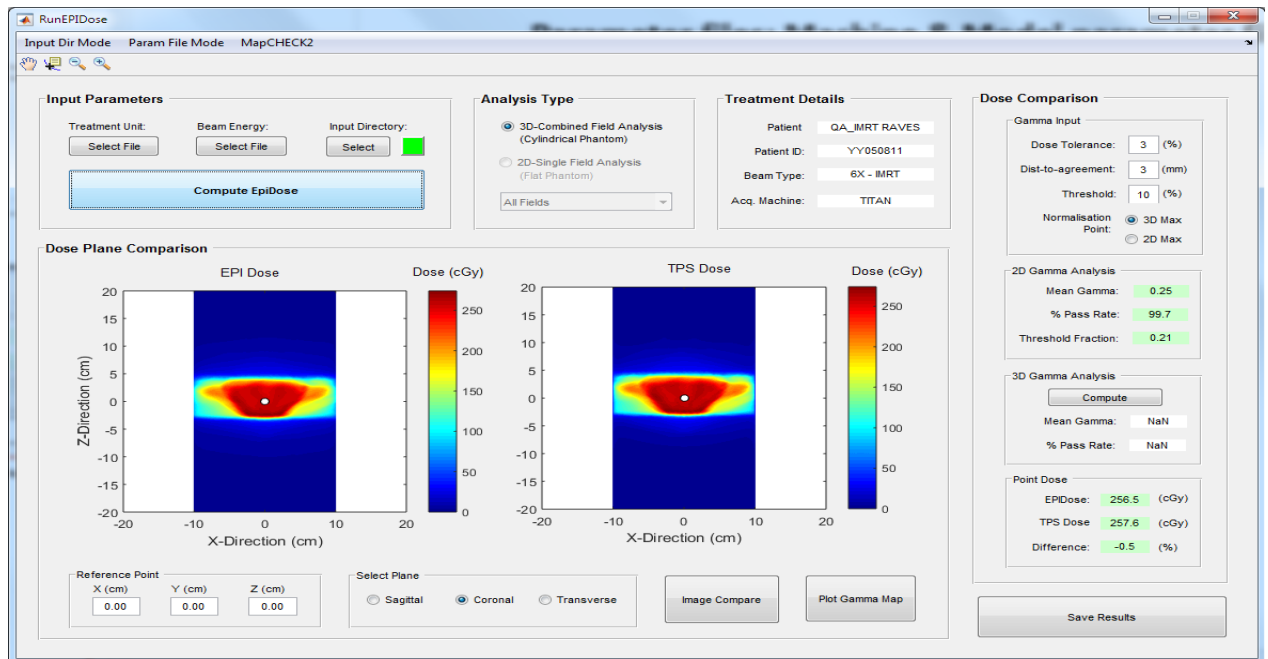


Figure 2-9- Snapshot of the graphical user interface (GUI) software for assessment of a pre-treatment delivery for a post-prostatectomy plan. Left and right images show respectively delivery and treatment planning system (TPS) dose for the plan.

Instruction for participating centres in the audit

The EPID to dose conversion model is used to remotely audit IMRT/VMAT deliveries of clinical trials, and the virtual phantom concept introduces a web-based method to exchange data between participating and auditing centres. The participating centres are provided with comprehensive audit instructions developed by P.Greer, including a separate EPID guide to assist with correct calibration and operation. The Trans-Tasman Radiation Oncology Group (TROG) supplies IMRT head and neck (HN) and post-prostatectomy (PP) trial benchmarking plan instructions and CT data sets. Prescriptions, PTV and OAR constraints for both cases are shown in Table 2-2. The CT datasets of two standard virtual water-equivalent QA phantoms are also provided; a VFP and a VCP. The VFP is 41 cm in length (superior-inferior direction) and 43 cm×35 cm in cross-section. The VCP is 40 cm in length and 20 cm diameter in cross-section. Summary of the instructions for participating centres is demonstrated in Figure 2-10. The full audit instructions are provided in Appendix.

Table 2-2- A summary of planning constraints for the two benchmarking plans: head and neck (HN) and post-prostatectomy (PP) plans.

PTV					
PP: Total Dose:V100% = 64 Gy			HN: Total Dose:V100% = 70 Gy		
Criteria	Dose range (Gy)	Value (Gy)	Criteria	Dose range (Gy)	Minor & Major violation (Gy)
D98%	>95%XV100%	>60.8	D95%	PTV70>=66.5	65.1<D<66.5 & D<65.1
				PTV67>=63.65	60.3<D<63.65 & D<60.3
				PTV63>=59.85	58.6<D<59.85 & D<58.6
				PTV54>=51.30	45.9<D<51.3 & D<45.9
Mean dose (Dmean)	- 1%<Dmean<2%	63.4<Dmean<65.3	Median Dose for PTV70 (Dmedian)	68.8<Dmedian<71.4 (±2% of 70 Gy)	
Maximum D2%	<107% $\times$ V100%	<68.48	Maximum D2% for PTV70	<77.0	77<D<80.5 & D<80.5
Normal tissues					
Rectum: V60Gy & V40Gy	<40% & <60%	<24 & <24	(D1%) for Spinal cord & PRV Spinal cord	-	<45 & <50
Femoral heads: V35Gy, V45Gy and 60 Gy	<100%, <60% and <30%	<35, <27 & <18	(D1%) for Brachial plexus	-	<66

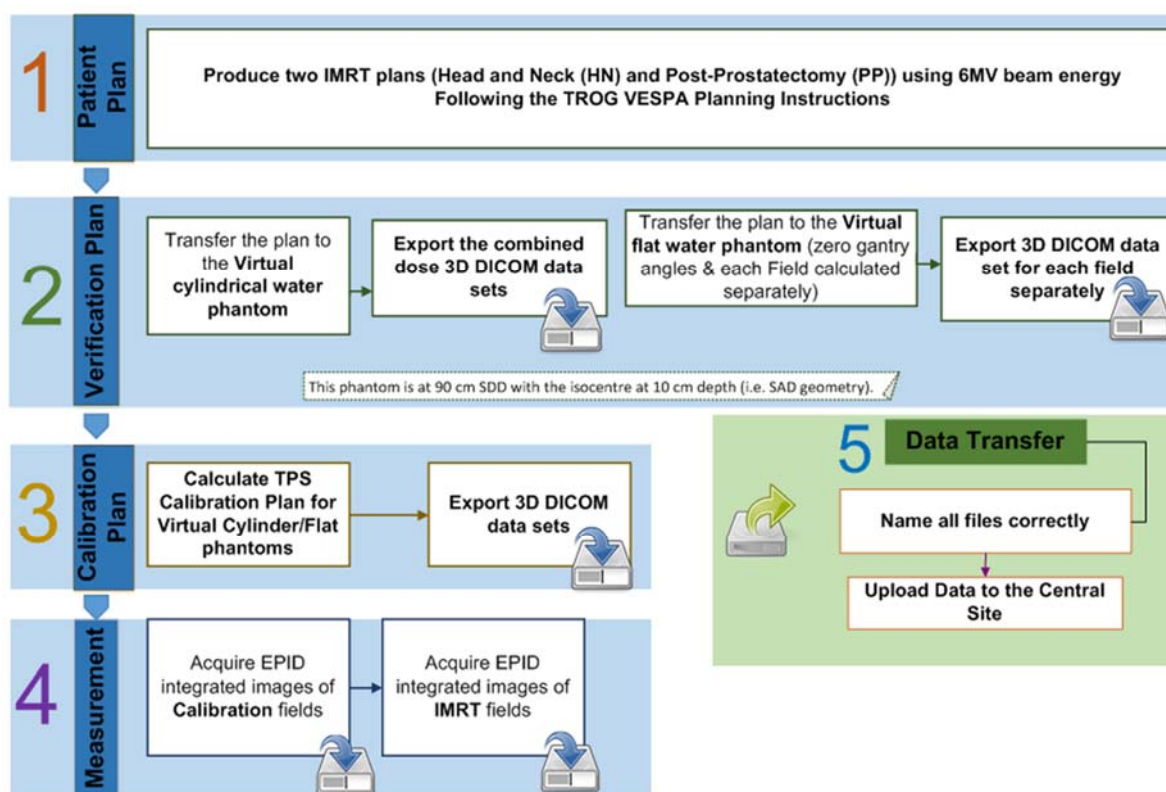


Figure 2-10- An overview of the VESPA instructions for participating centres.

Scope of the audit in this thesis

In this thesis, audit methods and results are described for Varian linear accelerators for IMRT using integrated images of each field, and VMAT using cine image acquisitions. For Elekta systems only IMRT audits were performed using integrated images for each field. As described above, the Elekta system at the time of this thesis did not have a clinical cine mode that could be used for the audit. The XIS software was investigated separately but due to the lack of gantry angle information for the cine images it was not used.

EPID-based dosimetry to verify IMRT planar dose distribution for the aS1200 EPID and FFF beams

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Abstract

We proposed to perform a basic dosimetry commissioning on a new imager system, the Varian aS1200 electronic portal imaging device (EPID) and TrueBeam 2.0 linear accelerator for flattened (FF) and flattening filter-free (FFF) beams, then to develop an image-based quality assurance (QA) model for verification of the system delivery accuracy for intensity-modulated radiation therapy (IMRT) treatments. For dosimetry testing, linearity of dose response with MU, imager lag, and effectiveness of backscatter shielding were investigated. Then, an image-based model was developed to convert images to planar dose onto a virtual water phantom. The model parameters were identified using energy fluence of the Acuros treatment planning system (TPS) and, reference dose profiles and output factors measured at depths of 5, 10, 15, and 20 cm in water phantom for square fields. To validate the model, its calculated dose was compared to measured dose from MapCHECK 2 diode arrays for 36 IMRT fields at 10 cm depth delivered with 6X, 6XFFF, 10X, and 10XFFF energies. An in-house gamma function was used to compare planar doses pixel-by-pixel. Finally, the method was applied to the same IMRT fields to verify their pretreatment delivery dose compared with Eclipse TPS dose. For the EPID commissioning, dose linearity was within 0.4% above 5 MU and ~1% above 2 MU, measured lag was smaller than the previous EPIDs, and profile symmetry was improved. The model was validated with mean gamma pass rates (standard deviation) of 99.0% (0.4%), 99.5% (0.6%), 99.3% (0.4%), and 98.0% (0.8%) at 3%/3 mm for respectively 6X, 6XFFF, 10X, and 10XFFF beams. Using the same comparison criteria, the beam deliveries were verified with mean pass rates of 100% (0.0%), 99.6% (0.3%), 99.9% (0.1%), and 98.7% (1.4%). Improvements were observed in dosimetric response of the aS1200 imager compared to previous EPID models, and the model was successfully developed for the new system and delivery energies of 6 and 10 MV, FF, and FFF modes.

Key Words: EPID, dosimetry, IMRT, IMRT treatment plan verification, FFF beams

I. INTRODUCTION

Accuracy of dose delivery for IMRT treatments should be determined by an accurate quality assurance (QA) procedure [1]. Recently, there has been a lot of interest in using flattening filter-free (FFF) beams which give the benefit of reduced headscatter and hence dose outside the field [2]. These beams also deliver the dose faster than flattened beams, which could be beneficial for hypofractionated treatments and reducing intrafractional organ motion [3]. Therefore, they require accurate and efficient quality assurance procedures including patient-specific quality assurance.

Linear accelerators (linacs) are equipped with EPIDs originally designed for patient positioning, [4] but because EPIDs have high sensitivity, spatial resolution, and immediate digital format, they have also been utilized to determine dose for routine QA of linacs or dose verification of treatments [5, 6, 7, 8]. The Varian aS1200 EPID detector (Varian Medical Systems, Palo Alto, CA) was released recently and has a large area ($40 \times 40 \text{ cm}^2$), small pixel size (0.0336 cm), and advanced acquisition electronics, and is potentially an improved design for dosimetry [9]. It contains additional backscatter shielding layers to reduce backscatter artifacts from the robotic support arm. It has been adapted by Varian for use in FFF beams without saturation at any source-to-detector distance [10, 11].

EPID-based dosimetry is performed by either (a) simulating the pixel values or (b) converting the pixel values to dose in phantom using a conversion model [12, 13]. The former is based on modeling the detector (EPID) response through Monte Carlo calculation [14, 15] or empirical techniques. The most commonly used empirical model is based on pencil-beam convolution of a simple fluence model with an EPID dose kernel, and Varian Medical Systems has commercialized this method. [16] For the latter image conversion methods, several mathematical models have been developed to estimate dose to water from EPID images [17, 18, 19, 20, 21, 22]. To date, very limited investigations of models to calculate dose in water from EPID images have been reported for high dose-rate FFF beams, higher energies, and for the new Varian aS1200 EPID design. Podesta et al. [23] reported development of a model for time-resolved assessment of VMAT for FFF beams for the aS1000 imager; however, they reported time-dependent gamma evaluations rather than integrated image comparisons.

Recently, an EPID to dose conversion model was developed and validated for 6 MV flattening filter energy (6X), using square field images defined by the multileaf collimator [22]. The model converts images to incident fluence then calculates dose in water using depth-dependent scatter kernels. They recorded nontransmission images with a prototype backscatter shielded aS1000 EPID and C-series Varian linac. Gamma comparisons were made to MapCHECK 1 (Sun

Nuclear Corporation, Melbourne, FL) measurements for 28 IMRT fields. More recently Keller et al. [24] reported on a Varian implementation of this model for a selection of 6X and 6XFFF fields for the TrueBeam and aS1200 imager comparing converted images to MatriXX (IBA Dosimetry, Schwarzenbruck, Germany) dose measurements.

In this paper, dosimetric testing of the new aS1200 EPID with a Varian TrueBeam linac is performed to verify dose linearity response of the imager, imager lag, and effectiveness/improvement of its backscatter shielding over previous EPID designs. Then, to verify pretreatment dose deliveries, an in-house image-to-dose conversion model is investigated for the EPID using beams at 6 and 10 MV energies and FF and FFF modes. The model parameters are identified using images acquired with open jaw-defined fields, fluence from the Acuros (Varian), and measured doses in water phantom. To validate the model performance, the modeled dose is compared to planar dose for 36 IMRT fields measured by a MapCHECK 2 detector. Finally, the model was used to verify delivery accuracy in comparison with TPS dose from Eclipse AAA (V.11). This work should allow for efficient and comprehensive verification of conventional and FFF IMRT deliveries for 6 and 10 MV energies.

II. MATERIALS AND METHODS

A. Experimental measurements

An aS1200 EPID with a Varian TrueBeam linac (V.2.0) was used to acquire images. The EPID is attached to the gantry through a robotic arm [25]. The active area of the EPID for dosimetry mode is $40 \times 40 \text{ cm}^2$ with 1190×1190 pixel arrays and pixel pitch of 0.336 mm.

To perform the imager dosimetric testing, dose linearity response, lag, and symmetry of the EPID were studied. To verify linearity of the EPID dose response versus delivered dose, $10 \times 10 \text{ cm}^2$ images were acquired at incremental MU irradiations from 2–600 MU, and the central integrated pixel values (IPVs) per MU were plotted against MU. The images were acquired using 6X, 6XFFF, 10X, and 10XFFF beam energies with dose rates of 600, 1400, 600, and 2400 MU/min, respectively. Furthermore, the imager lag or charge carry-over from frame to frame was examined using frames captured by a frame-grabber system. The frame-grabber is a graphic card housed in a separate PC, and connected to the TrueBeam XI node via a unidirectional cable link. The EPID signal was found in a region of interest (ROI) of size $0.33 \times 0.33 \text{ cm}^2$ at the center of each image frame. Finally, to verify the effectiveness of the aS1200 backscatter shielding layers, cross-plane and in-plane profiles were compared through the central axis for different size square field images, 2×2 , 3×3 , 4×4 , 6×6 , 8×8 , 10×10 , 15×15 , 20×20 , and $25 \times 25 \text{ cm}^2$.

Following EPID testing, the model was adapted for pretreatment dose verification for the aS1200 for the higher 10X energy and the FFF modes. To identify the model parameters, for each energy, a set of jaw-defined square field size (FS) images was acquired, 3×3, 4×4, 6×6, 10×10, 15×15, 20×20, and 25×25 cm² at zero gantry angle and 100 cm source-to-detector distance (SDD). The Acuros TPS fluence for a 25×25 cm² beam was used to identify the parameters of the fluence model. Measured central axis dose and dose profiles of the fields in water phantom were used to identify parameters of the dose model. Dose profiles were measured by an IBA PFD-3G diode detector and central axis dose was measured by two detectors: a microDiamond (SCD) detector, type 60019 (PTW-Freiburg GmbH, Freiburg, Germany) with 3.5 mm radius and 45.5 mm length, for 3×3 cm² field size and, 0.13 cm³ Scanditronix CC13 ion chamber (IBA Dosimetry, Schwarzenbruck, Germany) for the other fields. The measurements were performed in a Scanditronix Wellhofer water tank at depths of 5, 10, 15, and 20 cm, with 100 cm source-to-surface distance (SSD).

After parameter identification, to validate the model, the modeling results were compared with measurement results. For validation, integrated EPID images of nine head and neck IMRT fields were acquired at 6X, 6XFFF, 10X, and 10XFFF energies and 100 cm SDD at gantry zero. Delivered dose of each field was recalculated for the same fluence but modified dose rate and energies. This was done to better enable comparison between results for the four energies. These were used to model the dose at 10 cm depth in water. For the same fields, doses were measured with a MapCHECK 2 array (Model 1177, Sun Nuclear Corporation) at 10 cm depth in solid water and 100 cm to the detector plane. An in-house gamma function was used to compare planar doses pixel-by-pixel. The function uses a global dose difference (DD) criteria defined by the percentage of maximum dose of each 2D image plane. All doses above 10% of the maximum dose are assessed with a search region of 6 mm radius (26). The employed (DD) / (Distance-to-Agreement) mm were 3%/3 mm, 2%/2 mm, and 1%/1 mm. All doses are absolute dose as the model converts EPID grayscale images to absolute dose in Gy (i.e., no normalization is performed). The model was then used to verify pretreatment IMRT deliveries by comparison to Eclipse dose planes for the same fields at 10 cm depth using both 3%/3 mm and 2%/2 mm criteria. The IMRT fields were calculated separately on a virtual water phantom with 90 cm SSD and the isocenter at 10 cm depth. Doses were calculated with at 1.5 mm grid size and the three-dimensional DICOM dose file exported. The TPS dose plane at 10 cm depth was then extracted for comparison to the EPID modelled dose.

B. Modeling

The method in King et al. [22] was developed to convert EPID images to 2D dose inside a virtual water phantom. This method uses two steps:

1. Incident fluence modelling

$$\psi_p(r) = [C_A(r) D_E(r)] \otimes^{-1} k_E(r)$$

Where $C_A(r)$ is a profile correction matrix, $D_E(r)$ is EPID image signal matrix, and is $k_E(r) = e^{-(a_1 r)} + a_2 e^{-(a_3 r)} + a_4 e^{-(a_5 r)}$ the EPID dose deposition kernel.

2. Fluence to dose in water phantom modelling

$$D_W(r) = D_{CAL} \cdot [T(r) \psi_p(r) A(r)] \otimes k_W(r)$$

where D_{CAL} is a calibration factor, $T(r) = 1 + b_1 r$ and $A(r) = e^{-(b_2 r)^2}$ are, respectively, terma and attenuation factors, and $k_W(r) = b_3 e^{-(b_4 r)^2} + b_5 e^{-(b_6 r)^2} + \frac{b_7}{r} e^{-(b_8 r)^2}$ is the dose deposition in water kernel.

In summary, for the modeling $C_A(r)$, $a_i (i=1-5)$, and $a_j (j=1-5)$, require identification. This was done following the procedure outlined in King et al. [22].

III. RESULTS

A. EPID dosimetry commissioning

Figure 1 demonstrates the EPID dose response linearity. The IPV per MU at central axis was determined for each energy and normalized to the value at 600 MU. Then, the EPID lag was quantified by calculating frame-by-frame EPID signal at the central axis. Figure 2 demonstrates the EPID signal versus frame number for the four beam energies. Finally, to examine backscatter shielding effectiveness, cross-plane and in-plane profiles were plotted for different square field size images with 6X energy. Figure 3 shows the profiles in both planes.

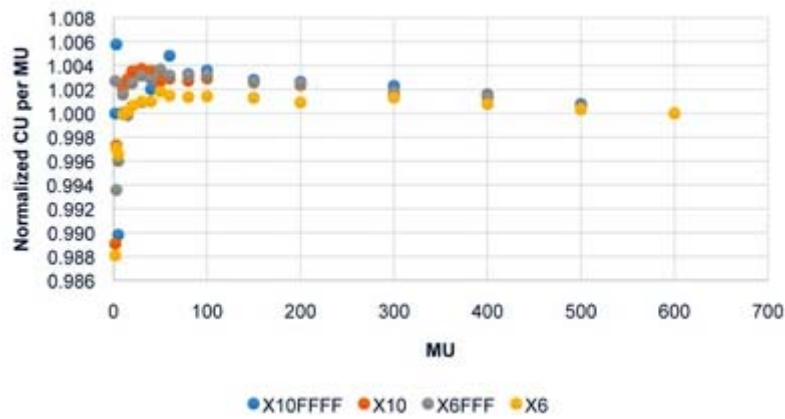


Figure 3-1- EPID dose response: IPV per MU versus MU (normalized to 600 MU values).

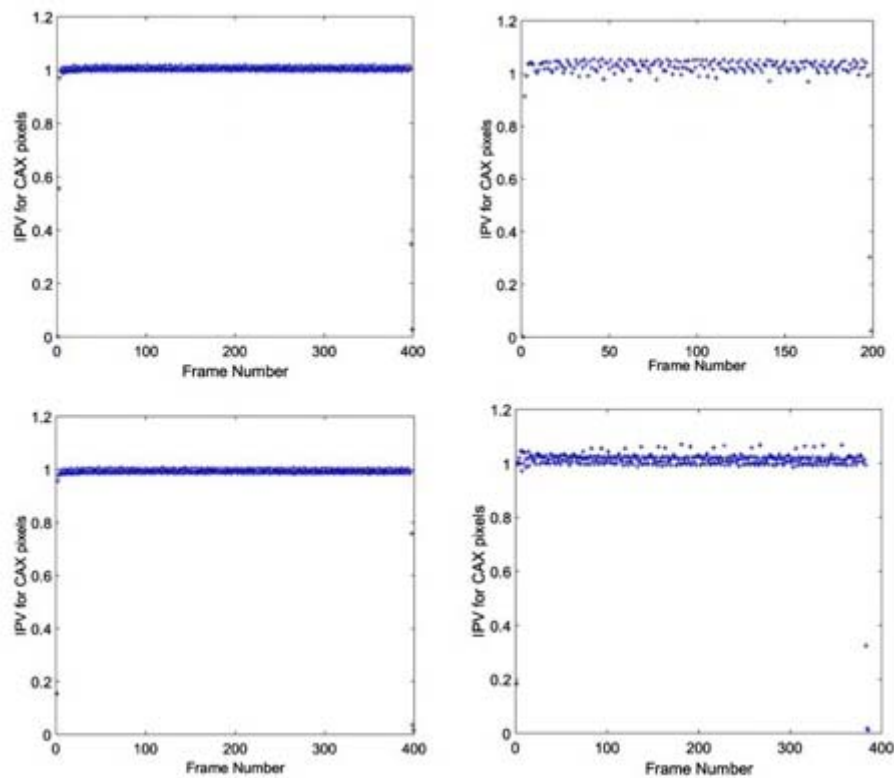


Figure 3-2- The imager lag for different beam energies. EPID signal in each frame was determined at the central axis, and normalized to the value at frame number 200.

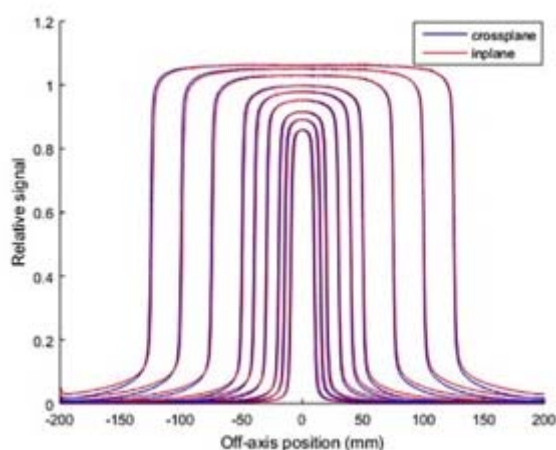


Figure 3-3- In-plane/cross-plane profiles to examine backscatter shielding effectiveness (6X).

B. EPID dose modeling

B.1 Fluence profile

For each beam energy, the EPID kernel parameters were identified using the 25×25 cm² fluence profile from the Acuros and the rest field sizes were used for cross-validation of the fluence model. The parameters have been summarized in the Appendix, Table A1. Figure 4 demonstrates the agreement between the modeled and the TPS fluence for the field sizes used.

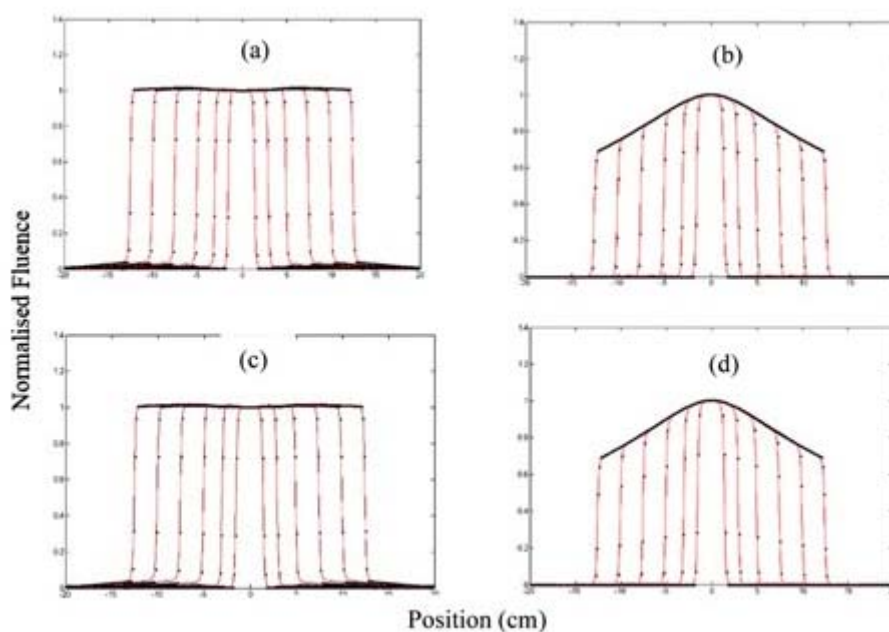


Figure 3-4- Cross-plane fluence profile versus field size for different beam energies: (a) 6X, (b) 6XFFF, (c) 10X, and (d) 10XFFF. Model: solid red lines, TPS: black dot lines.

B.2 Dose profile

The parameters of dose calculation in water were identified using measured central axis dose and dose profiles of 3×3 , 10×10 , 15×15 , and 20×20 cm² fields at depths of 5, 10, 15, and 20 cm in the water tank and the rest field sizes were used for cross-validation. The identified parameters are shown in the Appendix, Table A2. For the four beam energies, Figure 5 illustrates the comparison of the modeled (solid red line) cross-plane profile from the EPID images and the measured (black dot points) cross-plane profiles at 10 cm depth in water tank. The figure includes both training and cross-validation results. All dose profiles were normalized to the central axis dose of the 10×10 cm² image. Figure 6 demonstrates comparison of the modeled and measured central axis dose for all beam energies at the four different depths in water. All doses have been normalized to the 10×10 cm² field dose.

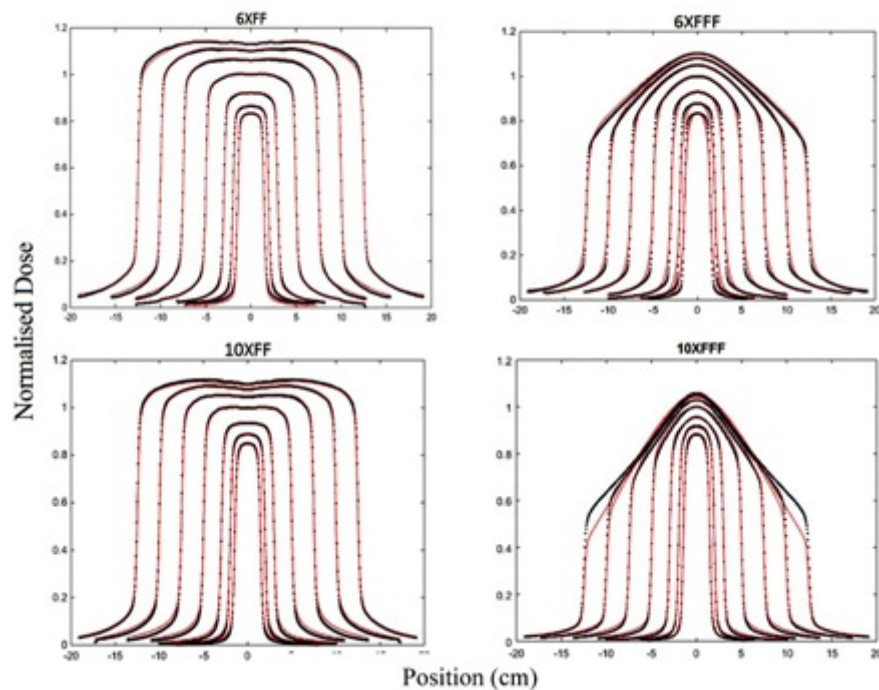


Figure 3-5- Comparison of modeled and measured cross-plane dose profiles at 10 cm depth in water. Model: solid red lines, measurements: black dot lines.

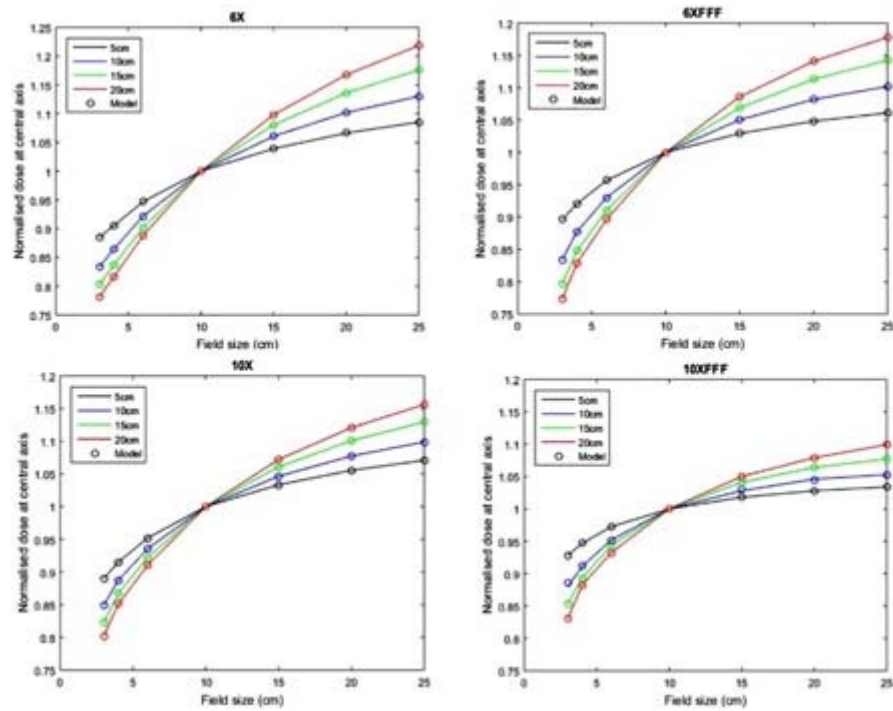


Figure 3-6- Normalized central axis dose in water versus field size at different depths. Model: circles, measurement: lines.

B.3 Model validation

To validate the model performance, the dose for nine IMRT head and neck fields were modeled from EPID images and compared to the measured doses with MapCHECK 2. The validation results have been summarized in Table 1.

Table 3-1- Model validation using MapCHECK 2 measurements

<i>Fields</i>	<i>10X</i>			<i>10XFFF</i>		
	<i>3%/3mm</i>	<i>2%/2mm</i>	<i>1%/1mm</i>	<i>3%/3mm</i>	<i>2%/2mm</i>	<i>1%/1mm</i>
1	99.7	96.2	75.7	99.2	96.0	74.3
2	99.6	93.2	67.9	97.0	85.6	57.3
3	99.3	93.6	71.9	97.8	90.9	63.0
4	98.9	93.9	73.0	98.4	89.0	63.4
5	99.6	95.0	73.0	98.0	89.6	64.4
6	99.5	94.8	71.5	98.3	98.1	92.9
7	99.0	95.4	72.8	98.9	93.8	79.4
8	98.7	93.8	70.3	97.4	89.9	57.8
9	99.8	95.3	77.4	96.7	88.2	58.4
Mean (SD)	99.3 (0.39)	94.6 (0.99)	72.6 (2.8)	98.0 (0.84)	91.2 (4.0)	67.9 (12.0)

B.4 Model performance

Finally, the modeled dose was compared to the TPS dose for the same fields. The comparison results have been summarized in Table 2. Figure 7 shows an example of the model performance compared with the TPS dose.

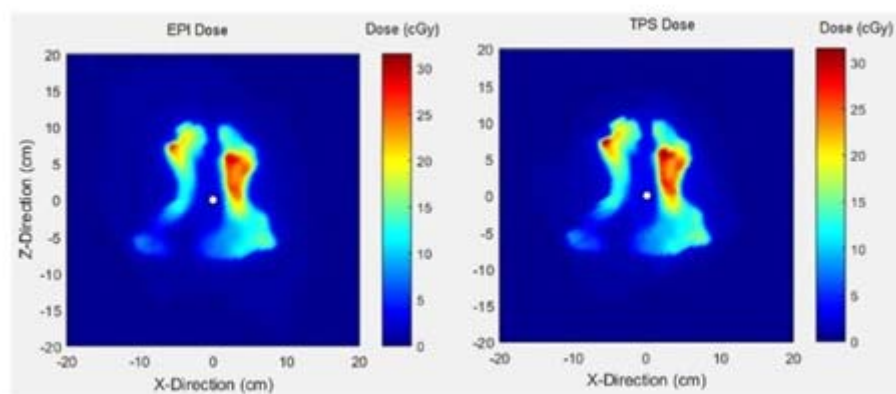


Figure 3-7- Dose matrix for a head and neck field of a 6XFFF beam with the modeled dose (left-side) and TPS dose (right-side) at 10 cm depth in water.

Table 3-2- Pretreatment verification using the model compared to TPS dose at 10 cm depth

<i>Fields</i>	<i>6X</i>			<i>6XFFF</i>		
	<i>3%/3mm</i>	<i>2%/2mm</i>	<i>1%/1mm</i>	<i>3%/3mm</i>	<i>2%/2mm</i>	<i>1%/1mm</i>
1	100.0	99.8	81.6	100.0	98.6	74.2
2	100.0	96.4	66.0	99.5	90.6	50.9
3	99.9	96.7	81.8	99.8	97.1	71.2
4	100.0	99.9	88.0	99.1	97.3	68.9
5	100.0	99.9	88.7	99.4	90.3	57.9
6	100.0	99.9	88.7	99.9	98.0	72.7
7	100.0	99.3	89.0	100.0	99.2	78.7
8	100.0	99.8	84.0	99.3	94.4	67.3
9	100.0	100.0	93.9	99.8	93.9	61.2
Mean (SD)	100.0 (0.0)	99.1 (1.4)	84.6 (8.0)	99.6 (0.33)	95.5 (3.4)	67.0 (8.8)

<i>Fields</i>	<i>10X</i>			<i>10XFFF</i>		
	<i>3%/3mm</i>	<i>2%/2mm</i>	<i>1%/1mm</i>	<i>3%/3mm</i>	<i>2%/2mm</i>	<i>1%/1mm</i>
1	100.0	99.8	78.3	99.9	99.2	83.1
2	99.8	98.5	71.7	98.2	88.1	58.5
3	100.0	97.0	79.7	98.2	94.1	68.8
4	100.0	99.7	84.5	97.9	91.9	68.5
5	100.0	99.1	85.6	98.6	92.6	68.4
6	100.0	99.6	78.9	99.9	99.7	98.1
7	100.0	99.2	81.7	99.9	99.0	80.5
8	99.8	98.6	83.0	98.9	94.4	65.3
9	99.9	99.6	88.1	96.7	89.0	61.7
Mean (SD)	99.9 (0.09)	99.0 (0.89)	81.3 (4.8)	98.7 (1.1)	94.2 (4.3)	72.5 (12.5)

IV. DISCUSSION

Initially, this paper outlines major dosimetry tests performed to commission the new aS1200 EPID system on the TrueBeam accelerator. The linearity of the EPID dose response was within 0.4% above 5 MU and ~1% above 2 MU. This linearity of response is a considerable improvement over previous reports for both Varian IAS3 and other vendor EPID systems which show under-response of 3%–5% for small MU [27, 28, 29]. Moreover, the measured lag for the EPID was found to be extremely small compared with previous reports, which had shown lag effects of several percent with signal increasing with increasing MU due to charge carry-over. No increase in image signal with MU was apparent for the aS1200 EPID model. Furthermore, the symmetry of the profiles for the EPID was considerably improved over the aS1000 imager, indicating the effectiveness of the backscatter shielding in the new system [29, 30, 31]. This was previously investigated with a prototype shielded panel [22]. Studies on aS1000 imagers have demonstrated around 8% additional nonuniform backscatter to the panel introduces dosimetry artifacts [30, 32-34]. Combined with the active repositioning of the detector specified to within 0.5 mm for all gantry angles, these results suggest that the aS1200 has excellent properties for dosimetry and is clearly superior to previous models.

Secondly, to verify delivery dose, a kernel-based model was employed to determine delivered dose to a virtual flat water phantom. The model input is images acquired with EPIDs and its output is dose onto the virtual phantom. Jaw-defined fields were used to identify the model parameters for aS1200 imager; however, in King et al [22] MLC-defined fields were used. While MLC-defined fields should accurately account for the phantom scatter, they do not incorporate the variation in dose due to headscatter, which then may require a separate

correction factor. To identify the model parameters, TPS fluence and measured dose from square field irradiations inside a rectangular water phantom were utilized. As Figures 5 and 6 illustrate, the modeled dose profiles closely follow the measured profiles for square field irradiations. Disagreement between the modeled and measured results was slightly larger for 6XFFF profiles and large field sizes of 10XFFF profiles. This could be because the model was originally developed to model flattening filter beams. The reduced performance of the model for FFF beams is likely due to the more complex structure of FFF beam profiles with field size. This structure also makes kernel parameter identification more difficult. Adaptations to the model to improve this could include an improved off-axis model for FFF deliveries. Another possibility would be to investigate whether the 6X and 10X kernels can accurately model the FFF beams allowing parameter identification to concentrate on modelling the beam profiles for these beams. The model comparison to measured MapCHECK data gave average gamma values over 99% for three energies and 98% for 10XFFF. These were assessed only at 3%/3 mm criteria as the MapCHECK is a low-resolution dosimeter with detector spacing of 7.07 mm.

To ultimately validate the model for clinical fields, modeled dose was compared with measured dose. Table 1 shows the validation results at three gamma criteria. According to this table, for all four energies, the modeled dose had more than 97% agreement with measured dose at 3%/3 mm criteria. Using tighter criteria, the lowest mean pass rates were 91.2% and 67.7% respectively for 2%/2 mm and 1%/1 mm criteria. This relatively poor accuracy for the more stringent criteria could come from MLC interleaf leakage alignment with diode detectors in MapCHECK, detector limitation in measurement, and/or human errors. Altogether, the validation results show a slight improvement over similar studies comparing their model with MapCHECK measurements [35]. Finally, the model was used to verify pretreatment deliveries of the same clinical fields in comparison with corresponding TPS prescribed dose. According to Table 2, more than 99% and 94% pixel similarity was observed at respectively 3%/3 mm and 2%/2 mm. However, one may observe the higher pass rates when comparing to TPS than the MapCHECK measurements, similar to other studies [35, 36]. This is possibly due to smaller number of detectors in MapCHECK compared to the EPID and measurement uncertainties.

V. CONCLUSIONS

Images from electronic portal imaging device (EPID) provide an efficient tool to verify pretreatment delivery dose for radiation therapy. In this paper, a model was derived to estimate the dose inside a virtual flat water phantom for the aS1200 EPID and flattened and FFF beams at 6 and 10 MV. The model parameters were identified using measured dose in water phantom for open field beams. Then, the model performance for IMRT planar fields was validated in

comparison with MapCHECK measurements at 10 cm depth in solid water. The model later verified delivery dose of 36 IMRT fields.

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APPENDICES

Appendix A. Identified parameters for kernels.

Table 3-3- Identified parameters of the EPID kernel for different beam energies

<i>Profile</i>	<i>a₁</i>	<i>a₂</i>	<i>a₃</i>	<i>a₄</i>	<i>a₅</i>
6X	34.99996	0.00025	1.28729	0.20810×10^{-6}	0.00010
6XFFF	34.99910	0.00040	1.80128	0.10621×10^{-4}	0.32360
10X	34.99999	0.00024	1.12503	0.36956×10^{-6}	0.00010
10XFFF	34.99910	0.00042	1.91661	0.13382×10^{-4}	0.31100

Table 3-4- Identified parameters of the dose kernel for different beam energies

	b_1	b_2	b_3	b_4	b_5	b_6	b_7	b_8
<i>6X</i>								
5 cm	0.00518	0.00037	0.99515	5.39435	0.00339	1.22809	0.00146	0.06500
10 cm	0.00553	0.00057	0.99281	5.61374	0.00323	1.07309	0.00387	0.07789
15 cm	0.00555	0.00077	0.98969	5.51726	0.00249	1.02052	0.00782	0.09486
20 cm	0.00479	0.00081	0.98745	5.79484	0.00189	1.10226	0.01066	0.10108
<i>6FFF</i>								
5 cm	0.00297	4.233×10^{-10}	0.99456	7.80186	1.18×10^{-10}	0.41998	0.00544	0.18896z
10 cm	0.00381	1.048×10^{-10}	0.96347	14.9991	0.03170	3.31156	0.00483	0.16257
15 cm	0.00310	2.341×10^{-9}	0.95928	14.9991	0.03424	3.23009	0.00648	0.14223
0 cm	0.00206	2.287×10^{-9}	0.99287	9.91172	5.92×10^{-10}	0.18822	0.00707	0.18876
<i>10X</i>								
5 cm	0.00312	0.00013	0.99451	8.01828	0.00536	1.56025	0.00012	1.13×10^{-6}
10 cm	0.00285	0.00029	0.99470	8.71554	0.00479	1.39946	0.00051	0.02454
15 cm	0.00196	0.00031	0.99497	10.5116	0.00396	1.44461	0.00107	0.06937
20 cm	0.00210	0.00049	0.99658	14.9910	0.00245	1.53943	0.00097	0.09055
<i>10XFFF</i>								
5 cm	2.880×10^{-9}	0.00328	0.98490	6.96354	0.00019	0.28823	0.01502	0.99999
10 cm	4.869×10^{-9}	0.003377	0.990033	6.414833	1.41×10^{-11}	0.471982	0.009967	0.171306
15 cm	4.42×10^{-10}	0.003379	0.987812	14.99998	7.853×10^{-5}	0.248740	0.012109	0.999985
20 cm	1.015×10^{-8}	0.003579	0.989572	7.275969	1.08×10^{-11}	0.768778	0.010428	0.137472

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Remote dosimetric auditing of clinical trials: the need for vendor specific models to convert images to dose

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Abstract

Introduction: A previous pilot study has demonstrated the feasibility of a novel image-based approach for remote dosimetric auditing of clinical trials. The approach uses a model to convert in-air acquired IMRT images to delivered dose inside a virtual phantom. The model was developed using images from an electronic portal imaging device (EPID) on a Varian linear accelerator. It was tuned using beam profiles and field size factors (FSFs) of a series of square fields measured in water tank. The current work investigates the need for vendor specific conversion models for image-based auditing. The EPID measured profile and FSF data for Varian (vendor 1) and Elekta (vendor 2) systems are compared along with the performance of the existing Varian model (VM) and a new Elekta model (EM) for a series of audit IMRT fields measured on vendor 2 systems. **Materials and methods:** The EPID measured beam profile and FSF data were studied for the two vendors to quantify and understand their relevant dosimetric differences. Then, an EM was developed converting EPID to dose in the virtual water phantom using a vendor 2 water tank data and images from corresponding EPID. The VM and EM were compared for predicting vendor 2 measured dose in water tank. Then, the performance of the new EM was compared to the VM for auditing of 54 IMRT fields from four vendor 2 facilities. Statistical significance of using vendor specific models was determined. **Results:** Observed dosimetry differences between the two vendors suggested developing an EM would be beneficial. The EM performed better than VM for vendor 2 square and IMRT fields. The IMRT audit gamma pass rates were $(99.8 \pm 0.5)\%$, $(98.6 \pm 2.3)\%$ and $(97.0 \pm 3.0)\%$ at respectively 3%/3mm, 3%/2mm and 2%/2mm with improvements at most fields compared with using the VM. For the pilot audit, the difference between gamma results of the two vendors was reduced when using vendor specific models (VM: $p < 0.0001$, vendor specific models: $p = 0.0025$). **Conclusion:** A new model was derived to convert images from vendor 2 EPIDs to dose for remote auditing vendor 2 deliveries. Using vendor specific models is recommended to remotely audit systems from different vendors, however the improvements found were not major.

Key Words: Remote radiotherapy auditing, Elekta/Varian linacs, IMRT pretreatment dose verification

I. INTRODUCTION

Quality assurance (QA) is an essential procedure to assess accuracy of relevant parameters in radiotherapy [1] while an external audit is recommended to assess consistency of local QA and effectiveness of delivery and measurement systems [2]. The importance of external audits is emphasized in radiotherapy clinical trials where a consistent accuracy is essential [3-5]. Conventional audits are performed by site-visits or postal methods, which can be expensive and/or labour intensive [6-8]. Some virtual methods have been explored to reduce the audit cost using in-house QA methods [9].

Recently a novel approach was introduced to remotely assess intensity modulated radiotherapy (IMRT) deliveries using pre-treatment images from electronic portal imaging devices (EPIDs). The method was known as the Virtual Epid Standard Phantom Audit (VESPA) and designed for dosimetric auditing of clinical trials at remote facilities. The VESPA utilized an in-house software for analysis and provided a relatively consistent detection system for data acquisition [10]. Participating facilities were provided with CT data sets of the virtual water phantoms and transferred prostate and head and neck IMRT treatment plans onto these to calculate dose in their local treatment planning system (TPS). They electronically sent their images and planned dose to the auditing site for assessment.

The in-house software of the VESPA back-projects in-air acquired images from EPIDs into virtual water phantoms and converts the signals to dose at 10 cm depth within the phantoms [11, 12]. The conversion is performed based on a model developed by King et al at Calvary Mater Newcastle Hospital (CMNH). The software input includes a machine specific file, a beam model file and DICOM images and doses. The machine specific file refines the input and adapts it to each machine/delivery system using the facility calibration images. This file includes parameters defining central axis coordinate on the EPID and EPID-linac sag correction. Another software input is the beam model file referred to here as the Varian model (VM). The VM is not adjusted for each facility. It has been developed using aS1000 EPID acquired images from a Varian linac deliveries (vendor 1) of series of square fields. The beam profiles and field size factors (FSFs) of the deliveries were also measured in water tank and used for the VM optimisation. The VM has been extensively benchmarked and used for vendor 1 in-house QA.

Six facilities took part in a pilot study of the remote based auditing method. Three of the facilities acquired data from Varian delivery and measurement systems (vendor 1) and three from Elekta (vendor 2) [13]. The pilot study used the VM for both vendors but applied primary vendor differences to the machine specific file. Differences in the detector size and resolution

were applied; vendor 1: aS1000 EPIDs with 40×30 cm² active area, i.e. 1024×768 image resolution with 0.039 cm pixel resolution and, vendor 2: iViewGT EPIDs with 41×41 cm² active area, i.e. 1024×1024 image resolution with 0.040 cm pixel resolution [14]. Moreover, prior to analysis, acquired images at 160 cm source to detector distance (SDD) from vendor 2 were resampled to 100 cm. The '.HIS' format images acquired from iViewGT EPIDs were also converted to DICOM in consistent with the software input requirement. In spite of the applied differences to each machine file, slightly lower gamma pass rates were observed in the auditing results from vendor 2. The vendor 2 systems also demonstrated a different field size response for reconstructed dose at the phantom isocentre compared with those from vendor 1. These all could be due to the differences of relevant dosimetry characteristics between the two vendors. Ignoring the differences can result in significant uncertainties in the audit outcome [15]. Accordingly, this research studies relevant dosimetric variations between the two vendors and corresponding dose conversion models. Then, it investigates whether using vendor specific models could make the audit results independent from the vendors.

This research investigates differences of the beam profiles and FSFs, for the two vendors. The parameters are used in the development of the image to dose conversion model which in turn is applied for data analysis of the remote EPID based audit. The current study develops a model (EM) to convert images from EPID to dose inside the virtual phantom for vendor 2 deliveries. Then, the EM performance is compared with the VM for measured water tank data from vendor 2 deliveries. The EM is used for remote auditing of 54 IMRT fields from four vendor 2 facilities. Statistical study of the auditing results determines whether a vendor specific model is required for auditing of each vendor. This work will facilitate implementation of this new and efficient auditing procedure using a remote EPID based dosimetry with improved sensitivity.

II. MATERIALS AND METHODS

A. Dosimetry

A series of square field beams, 3×3 , 4×4 , 6×6 , 10×10 , 15×15 , 20×20 and 25×25 cm², were delivered by a vendor 1 and a vendor 2 linac and, in-air images were acquired by respectively an aS1000 and iViewGT EPID. The profiles and FSFs were acquired from the image signals to evaluate the differences of relevant dosimetric parameters between the two vendors. Note, the profiles and FSFs were later used for modelling signal to dose. The profiles were obtained from the pixel data in the crossplane through the central axis. The profiles penumbras were defined to quantify the profile differences. The penumbra widths were defined as the distance between 80% and 20% of the maximum dose for each side of the profile relative to central axis. The FSFs were directly extracted from the mean pixel value of the central 11×11 pixels of the image

signals and, the difference between FSFs of the vendors was quantified by percentage differences as $D = (D_{\text{vendor1}} - D_{\text{vendor2}}) * 100 / D_{\text{vendor1}}$.

An intra-vendor study was conducted on four vendor 2 facilities to evaluate variations of their parameters. The facilities were called C₁, C₂, C₃ and C₄. The percentage difference was calculated for each facility ($PD_{C2, C3, C4} = S_{C1} - S_{C2, C3, C4} \times 100 / S_{C1}$, (S: Signal). Later, the C₁ image data were used to develop a new model (EM) for vendor 2. The relative consistency for vendor 1 facilities has been reported elsewhere [16, 17].

B. Modelling

Following the method of King et al [11], which was used to develop a vendor 1 model (VM), a vendor 2 model (EM) was developed to convert images to dose onto the virtual phantom. Images from an iViewGT EPID and a vendor 2 measured dose in water tank (WT) were acquired. The images were acquired in-air from delivery of series of square field beams, 3×3, 4×4, 6×6, 10×10, 15×15, 20×20 and 25×25 cm². The water tank data were measured at 10 cm depth and used to optimize the model parameters. The water tank data were acquired at 100 cm SSD using a small cylindrical ionization chamber of CC01 for small field sizes, i.e. 3×3, 4×4, 6×6 cm², and a CC13 for the large field sizes, i.e. 10×10, 15×15, 20×20 and 25×25 cm². All images were acquired at 160 cm SSD and resampled to 100 cm SSD using interpolation. The images were truncated at about 1 cm of the detector edge to avoid the edge artefacts. As the images were found noisier than those from aS1000 EPIDs, an adaptive ‘wiener2’ filter in MATLAB was used to reduce the image noise and its impact on the model convolution function. The ‘wiener2’ low pass filters the images that have been degraded by a constant power additive noise. It uses a pixel wise adaptive method based on statistics estimated from a local neighbourhood of each pixel [18]. An initial trial EM could not consistently predict the FSFs for the four facilities. After investigation, an averaged FSF from the TPSs of the four facilities was used as the reference FSF for modelling purposes, see Supplementary file. The EM model accuracy was quantified via calculating discrepancy between the image and water tank dose for the profiles and FSFs

$$ST = \sum \left(\frac{\text{image dose} - \text{water tank dose}}{n_{\text{fields}}} \right)^2 \quad (1)$$

where ‘nfield’ was number of dose measurements/points. Furthermore, percentage differences were calculated for the EM dose compared with water tank measured dose (WT) via ($PD_{\text{EM}} = (D_{\text{WT}} - D_{\text{EM}}) \times 100 / D_{\text{WT}}$, (D: dose). The EM performance was then compared with the VM performance for estimating a vendor 2 water tank dose (WT). The percentage difference was calculated for both cases ($PD_{\text{EM, VM}} = (D_{\text{WT}} - D_{\text{EM, VM}}) \times 100 / D_{\text{WT}}$, (D: dose).

C. Auditing

The EM was used to convert pre-treatment images from IMRT deliveries, a post-prostatectomy (PP) and a head and neck (HN) plan, to dose for four vendor 2 facilities. Details of these plans and the audit procedures are detailed elsewhere [10, 13]. Each facility delivered (7-9) IMRT fields per patient plan. For each field, the converted EPID dose was compared to corresponding TPS dose. The comparisons were performed by an in-house developed gamma function at three different criteria, 3%/3 mm, 3%/2 mm and 2%/2 mm. The EM performance was compared with the VM performance for the IMRT audits at 1%/1 mm gamma criteria. Finally, a statistical study was conducted on the pilot audit including facilities from both vendors to compare performance of the vendor specific models and VM solely applied to all facilities.

III. RESULTS

A. Dosimetry

Figure 1 demonstrates relevant parameters for the two vendors measured by corresponding EPIDs. As Figure 1a demonstrates, the two vendors show some profile differences mainly in the horns and edge regions. Penumbras for vendor 2 and vendor 1 profiles were shown by respectively $+$ and $\square$. The penumbra values were demonstrated by the profile signal values but with a 'cm' unit. For vendor 2, larger penumbras were observed at all field sizes. The Figure 1a subplot magnifies the $10 \times 10 \text{ cm}^2$ profiles. It showed large differences in horn and edge of the profiles. As Figure 1b demonstrates, FSFs of the vendor 2 are larger at large fields, $>10 \times 10 \text{ cm}^2$, and smaller at small fields, $<10 \times 10 \text{ cm}^2$, than other vendor. The percentage difference (D%) between FSFs of the vendors was better demonstrated in the subplot. The subplot shows largest discrepancy at the largest field sizes, i.e. $20 \times 20 \text{ cm}^2$.

Figure 2 shows the signal response for four vendor 2 facilities measured by their iViewGT EPIDs. The signals were compared to the C1 values as the C1 was later used for the EM development. In addition to signal profiles, Figure 2a shows values for the profiles penumbras. The penumbras were relatively similar for C₁ and C₄ and, for C₂ and C₃. However, a relatively large discrepancy was observed in penumbras of all facilities at the very large field, i.e. $20 \times 20 \text{ cm}^2$. The subplot in Figure 2a shows percentage difference for the $10 \times 10 \text{ cm}^2$ profiles. The largest difference was observed for C₃ and the smallest for C₂. Relatively similar trend was observed for other field sizes (not plotted). Figure 2b demonstrates the FSFs response for the four facilities and the subplot shows their percentage differences. For FSF, C₄ shows a relatively large discrepancy at most fields and C₃ shows the largest difference at the very large field, i.e. $20 \times 20 \text{ cm}^2$.

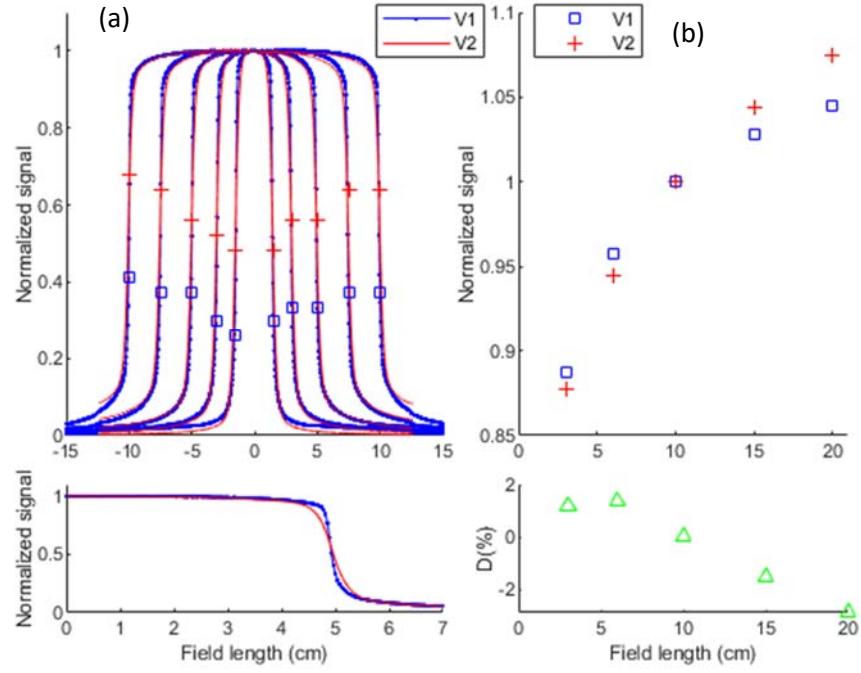


Figure 4-1- EPID measured signals for a vendor 1 and vendor 2 facility. a) Beam profiles. Penumbra for V2 and V1 profiles were shown by respectively + and □. Note, penumbra unit is 'cm'. The subplot magnifies the 10×10 cm² profiles for comparison. b) Field size factors (FSFs). The subplot demonstrates percentage differences for the FSFs. The profiles and FSF data were used to develop signal to dose conversion models (VM and EM).

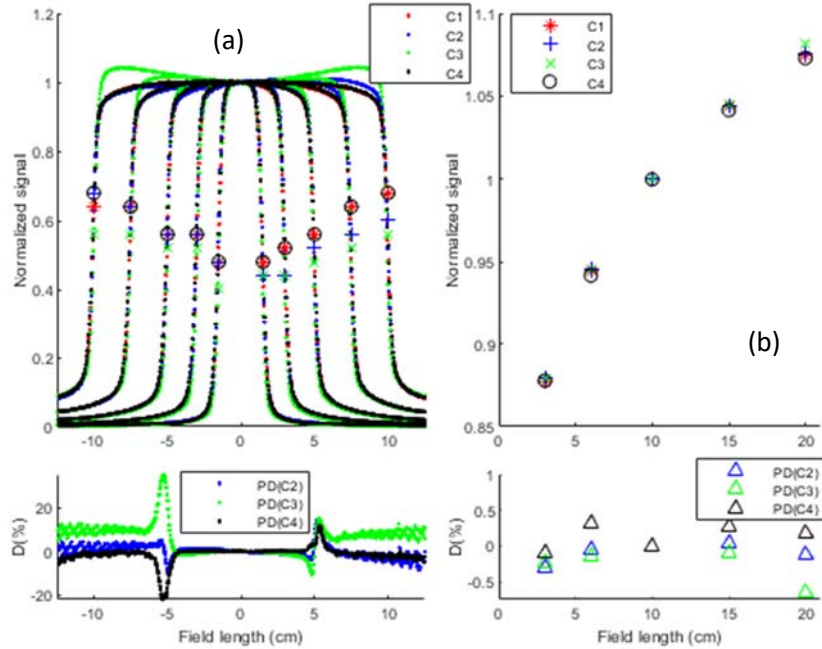


Figure 4-2- a) EPID measured signals for four vendor 2 facilities. a) Beam profiles. Penumbra for C₁, C₂, C₃ and C₄ profiles were shown by respectively *, +, × and ○. Note, penumbra unit is 'cm'. The subplot demonstrates percentage differences for the 10×10 cm² profiles. b) Field size factors (FSFs) for the four facilities. The subplot shows percentage differences for the FSFs. The percentage difference was calculated by $(PD_{C_2, C_3, C_4} = S_{C_1} - S_{C_2, C_3, C_4}) * 100 / S_{C_1}$, (S: Signal). Later, the C₁ image data were used to develop a new model (EM) for vendor 2.

B. Modelling

Figure 3 demonstrates the EM estimated dose compared with water tank (WT) measured dose for a vendor 2 facility. The ST values for the profiles and FSFs were respectively 3.7×10^{-6} and 1.9×10^{-6} which were close to the values for the established VM, 2.1×10^{-6} and 1.53×10^{-7} respectively [11]. The subplot of the Figure 3a shows percentage difference of the dose profiles for the 10×10 cm² profiles. The dips in the subplot came from the horns where the measured dose was smaller than the model dose. The peaks also originated from the profiles edge differences where the measured dose was larger than modelled dose. The dips/peaks demonstrated asymmetric response versus field size. Figure 3b shows the FSF dose measured by the EM and water tank (WT). The subplot showed the largest percentage difference at the very large field, i.e. 20×20 cm².

Figure 4a compares a vendor 2 water tank (WT) dose profiles estimated by both models, i.e. VM and EM. Penumbrae for the EM, VM and WT profiles were shown by respectively +, □ and ×. The EM penumbrae were closer to the WT penumbrae than the VM penumbrae. The subplot magnifies the 10×10 cm² profiles for a better visualization. A high agreement was observed between the EM and WT dose profiles. The Figure 4b demonstrates the models calculated FSFs compared with the WT dose and the subplot shows percentage differences for the FSFs. Slightly better FSF estimation was observed for the EM than VM dose.

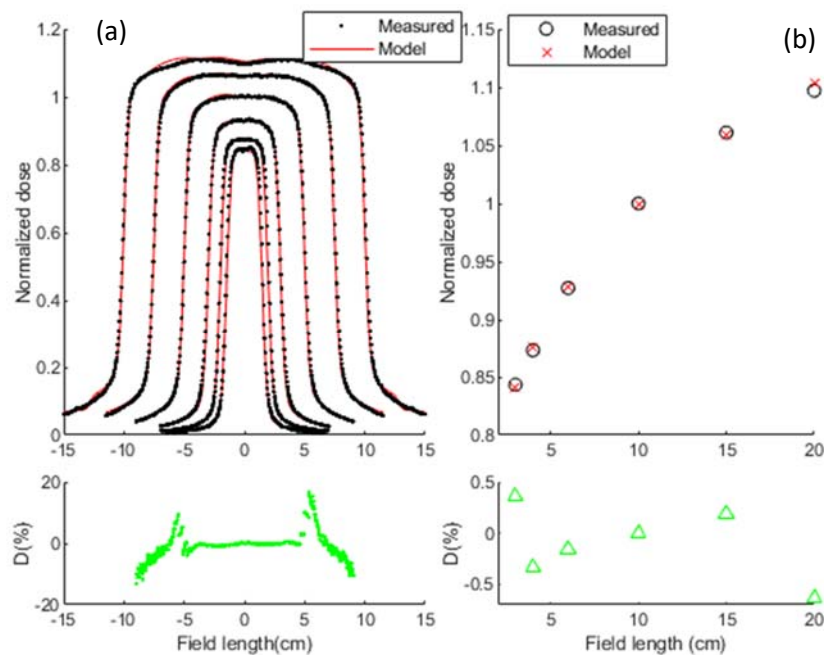


Figure 4-3- Measured dose by the new model (EM) compared with water tank measured data for a vendor 2 deliveries. a) Dose profiles. The subplot shows percentage differences for the 10×10 cm² profiles. b) FSF dose.

The subplot shows percentage differences for the FSFs. The percentage difference was calculated by $(PD_{EM} = D_{WT} - D_{EM}) * 100 / D_{WT}$, (D: Dose).

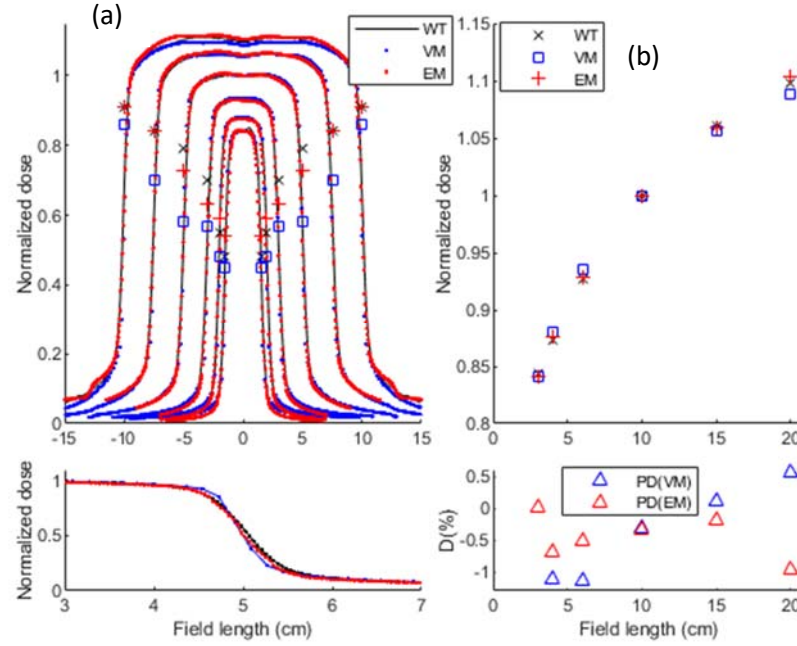


Figure 4-4- Performance of the two models (EM and VM) versus water tank (WT) dose for a vendor 2 deliveries. a) Dose profiles. Penumbra for the EM, VM and WT profiles were shown by respectively +, □ and X. Note, penumbra unit is 'cm'. The subplot magnifies the 10×10 cm² profiles for comparison. b) FSFs dose. The subplot shows percentage differences for the FSFs. The percentage difference was calculated by $(PD_{EM, VM} = D_{WT} - D_{EM, VM}) * 100 / D_{WT}$, (D: dose).

C. Auditing

Figure 5 summarizes the IMRT auditing results for vendor 2 facilities. The HN data from C₂ were not considered in any analysis as they had acquired calibration images at a different date from other EPID measurements. The audit result of each treatment site was assessed by pass rate boxplots and corresponding mean gammas. The HN mean gamma pass rates were $(99.9 \pm 0.2)\%$, $(98.8 \pm 1.7)\%$ and $(97.1 \pm 3.6)\%$ at respectively 3%/3 mm, 3%/2 mm and 2%/2 mm. The mean pass rates for the PP were $(99.8 \pm 0.7)\%$, $(98.4 \pm 2.7)\%$ and $(96.9 \pm 2.5)\%$ at the criteria. Interquartile ranges of the pass rates (mean gammas) at the gamma criteria were 0.1(0.05), 1.5(0.06) and 2.6(0.08) for the HN and 0.2(0.05), 1.3(0.06) and 2.9(0.06) for the PP. Figure 6 and Table 1 compare the auditing results for both the EM and VM using mean gamma values at 1%/1 mm criteria. Most of the HN and almost all PP fields from all facilities showed improved gamma results (lower mean gammas) for the EM than VM.

Figure 7 compares results of the pilot audit when using the VM for both vendors (blue boxplots) and when using vendor specific models (red boxplots) at 3%/3 mm criteria. Using analysis of variance (ANOVA) and Tukey-Kramer HSD methods for comparison of the mean gammas for the two scenarios, the former demonstrated a significant audit difference between two vendors

($p < 0.0001$). The mean gamma difference for the two vendors was reduced when using vendor specific models ($p = 0.0025$).

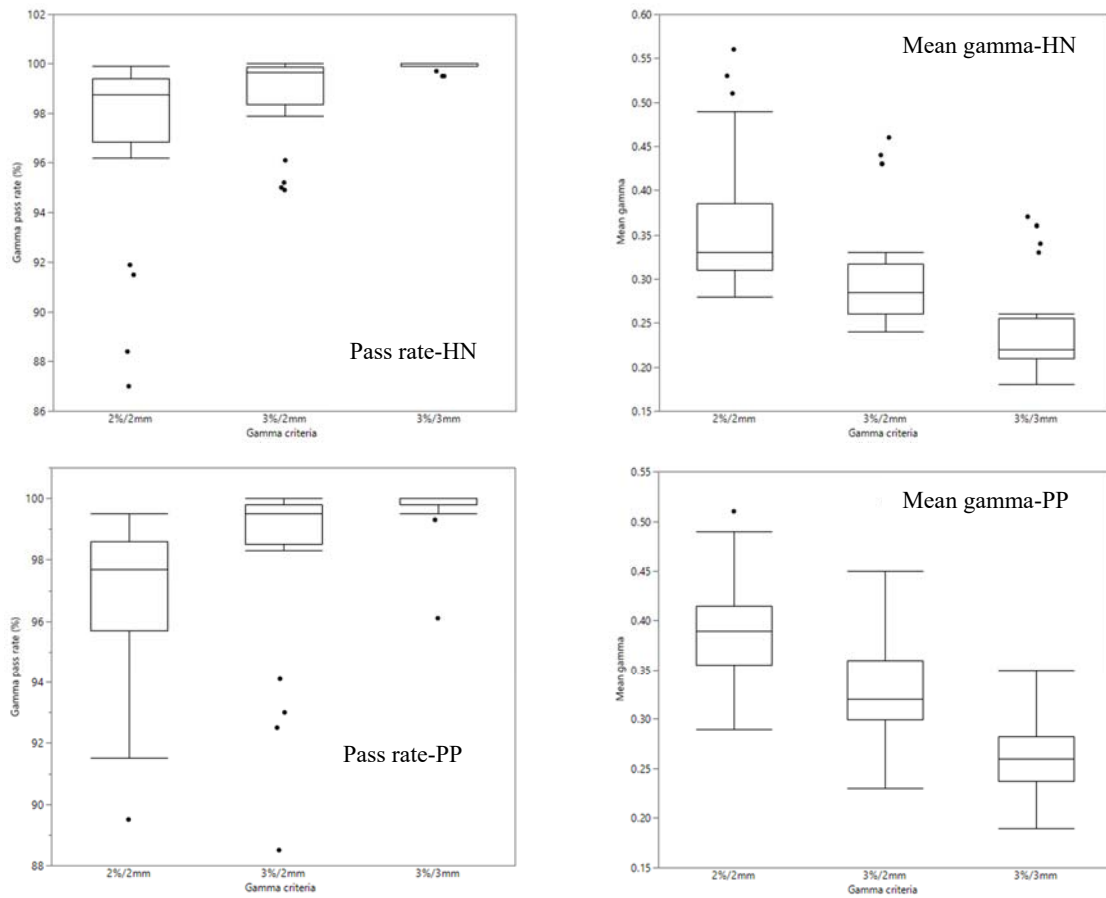


Figure 4-5- Auditing results of a post-prostatectomy (PP) and a head and neck (HN) plan from four vendor 2 facilities, C₁, C₂, C₃ and C₄, using the EM for analysis. Each facility has delivered (7-9) IMRT fields per treatment sites, totally 54 fields. The results include gamma pass rates and corresponding mean gammas for each patient plan.

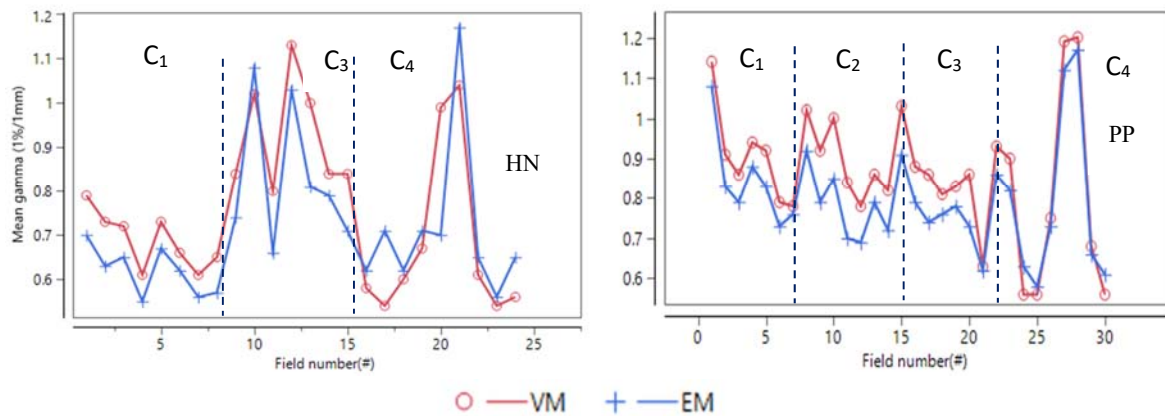


Figure 4-6- Mean gammas for the four vendor 2 centers for a a) head and neck (HN) and a b) Post-prostatectomy (PP) patient plan using both the EM and VM.

Table 4-1- Mean gamma pass rates at 1%/1mm for four vendor 2 facilities and two patient plans using both the EM and VM.

Centers	HN		PP	
	VM	EM	VM	EM
C ₁	78.5	83.5	66.1	69.4
C ₂	-	-	64.3	71.6
C ₃	63.1	69.3	68.5	73.1
C ₄	79.6	76.9	74.4	74.4

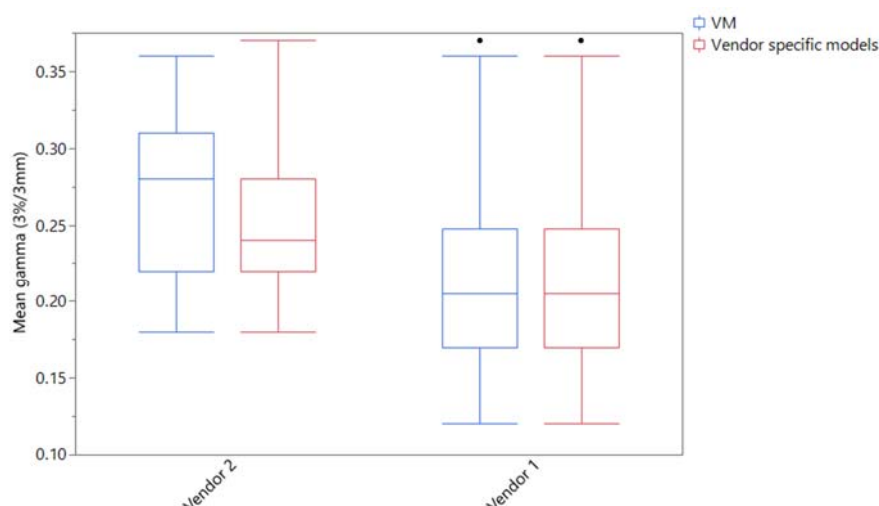


Figure 4-7- Auditing results for a study including two vendors. It uses either the VM or vendor specific models for dose conversion. The VM shows a significant difference between the two vendors ($p < 0.0001$). Using vendor specific models demonstrates less significant difference between the vendors ($p = 0.0025$).

IV. DISCUSSION

The VESPA auditing procedure is designed as an inexpensive and efficient auditing procedure that can be performed remotely with the time for the central site physicist generally being 2-3 hours to assess the results. The audit requires time from the local physicists to produce the IMRT verification plans and deliver the beams to the EPID however, all other auditing methods require local personnel time. The VESPA also does not include any equipment or transport costs. The studies on the method has been conducted on two vendors using one vendor verified model (VM) to convert the image signal to dose inside the phantom. Investigation for the need for vendor specific models makes the audit reliable over different vendors.

Studies on relevant EPID measured dosimetric parameters showed differences between the two vendors. The discrepancy increased between the vendors' profiles at the very small/large field sizes, $\sim 3 \times 3$ and 20×20 cm². The smaller penumbras observed for vendor 1 profiles indicate sharper profiles of corresponding images which may result in increasing the VM accuracy. The small penumbras for vendor 1 could be due to the proximity of the collimating system to the machine isocenter. For the FSFs of the two vendors, the discrepancy was increased by field size which was in accordance with the previous observations in the pilot audit. The FSF differences

between the vendors could be due to differences in either EPID scatter or head scatter beam as the EPID signals incorporate both effects.

The study on vendor 2 facilities showed some inconsistencies in their dosimetric parameters. The C_3 signals showed largest discrepancy with C_1 signals at profiles, penumbras and FSFs. The C_2 showed the minimum differences with the C_1 profile but for penumbras and, the C_4 showed the closest values to C_1 penumbras. However, the FSF influence seems more important than the profiles impact for the model accuracy since the FSFs are used in optimizing four out of six model parameters while two parameters are tuned by profiles. A comparison between Figures 1 and 2 shows larger inter-vendor discrepancy (vendor 1 and vendor 2) than intra-vendor variations (C_1 , C_2 , C_3 and C_4) for both parameters. This is in accordance with a report from Cozzi et al [19] and suggests developing a vendor 2 specific model may improve the auditing outcome.

A new model (EM) was developed for vendor 2 systems using a vendor 2 acquired parameters. The ST values for the EM were quite close to the values for the VM indicating high accuracy of the EM. Note, the VM has already been benchmarked and established as a reliable in-house QA tool. The model calculated dose is compared to corresponding TPS dose. High sensitivity of the model to the planned discrepancies ensures that clinically significant dosimetric errors are detectable. An in-house assessment demonstrated the method enough sensitivity to introduced MLC and/or collimator errors. However, a study on sensitivity of the gamma compared with a DVH approach is ongoing to determine the dose to the provided virtual patient CT dataset from the model. The model sensitivity to global dose differences is as expected dependent on the criteria with doses above the dose difference easily detected but those below it not.

The EM could accurately calculate water tank dose (WT) of a vendor 2 system. However, relatively large discrepancies were observed in horns and edges of the profiles. The EM dose also included small asymmetries in the profiles which may originate from the EPID image signals. Altogether, the EM was able to better calculate the WT dose profiles at all fields compare with the VM performance. For the FSFs, largest discrepancy of the EM with WT dose was observed at the very large field, i.e. $20 \times 20 \text{ cm}^2$. For most of the fields, the EM slightly better estimated the FSFs than the VM did.

The auditing pass rates for the two IMRT plans were relatively high for all facilities at the three gamma criteria and, their corresponding mean gammas showed similar behaviour. No significant difference was observed between the auditing results for the two treatment sites, the

HN and PP. For the HN results, more outliers were observed in the gamma results than for the PP audits. This could be due to relatively lower number of auditing fields included for the HN studies. In addition to analysis by treatment site, the results were analysed for each facility. Except for C4, mean gammas for all facilities and treatment sites were smaller for the EM than the VM. For C4, the VM demonstrated relatively better response for the HN. The VM, moreover, showed relatively similar response to the EM for the PP. In general, using the EM for auditing vendor 2 facilities reduced mean gammas though, the differences between the EM and VM performances were not easily observed unless a highly strict gamma criteria, i.e. 1%/1 mm, was used. This is in accordance with the above observations showing small improvement for calculating FSF dose.

The new EM and the VM were used to convert dose for deliveries from respectively vendor 2 and vendor 1 facilities in a study. The deliveries were also analysed using only VM for both vendors. Statistical studies of the two scenarios demonstrated a minor improvement when using vendor specific models ($p=0.0025$) than the VM ($p<0.0001$). Vendor dependency of the auditing results reduced when using vendor specific models (EM for vendor 2 and VM for vendor 1). However, mean gammas for vendor 2 were still larger than for vendor 1. This could be due to the impact of other variables such as facility TPS types which were not considered in this study.

V. CONCLUSIONS

Observed differences in relevant dosimetry parameters between vendor 1 and vendor 2 suggested using vendor specific models, to convert signal to dose onto the virtual phantoms, could account for dosimetry differences between the vendors. By developing a new model (EM) and using vendor specific models, the EM for vendor 2 and VM for vendor 1, the audit difference reduced between two vendors. The audit accuracy was improved and using vendor specific models was advised for future audits. The remote audit approach provides a highly automated method with significantly reduced cost.

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CONFLICT OF INTEREST STATEMENT

It is represented and warranted that, as at the date of this declaration, there is not any actual or perceived conflict of interest, or potential conflict of interest.

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SUPPLEMENTARY FILE

Observation

- Initial EM was developed using water tank data (profiles & FSFs).
 - (b3, b4): are short term parameters trained using crossplane profiles
 - (b5-b8): are long term parameters trained using FSFs
 - (b1-b2): are Terma and Attenuation factors trained by FSFs
- When using EM for both the HN and PP patients for the 4 Elekta facilities, C₁, C₂, C₃ , C₄:

The VM showed better performance than EM

For EM, the PP showed lower performance than H&N.

Main difference of patients: size.

Suggestion: The EM performance should be assessed for different field sizes.

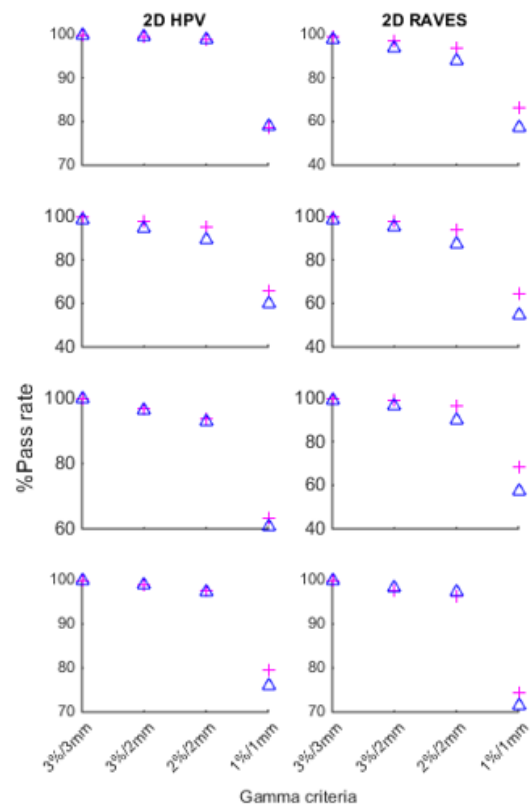


Figure 4-8- Gamma pass rates for both patients using both EM and VM. The VM shows better performance for most cases (Each row represents results of each facility, C₁, C₂, C₃, C₄ respectively).

Method

A- The performance of EM is compared with the VM performance for the FSFs of the facilities

B- Consistency of the EM performance is studied over different facilities

Results

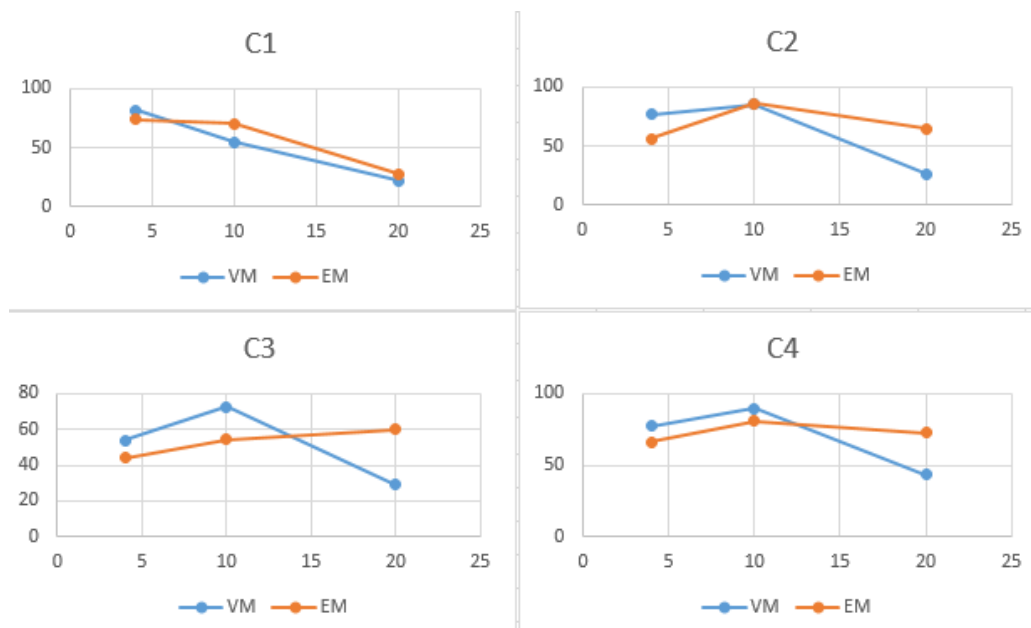


Figure 4-9- Gamma pass rates for the VM and EM vs field size for 4 facilities. The EM poor performance at fields ≤ 10 cm)

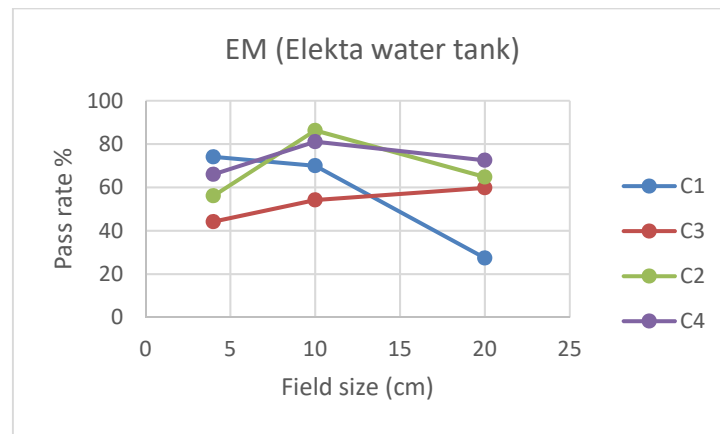


Figure 4-10- The EM performance for different field sizes for the 4 facilities. Inconsistent response of the facilities.

- The EM poor performance at small fields
- Inconsistent EM response over the facilities

Discussion

A- The EM performance was good at large fields (>10cm) but poor at small fields. Could be from dosimetry inaccuracy at small fields (FSFs).

Suggestion: The water tank FSFs used for the EM optimisation could be replaced with averaged FSFs from TPS of the 4 facilities.

B- For the EM, inconsistent gamma pass rates were observed between the facilities.

Gamma compares the EM with TPS and the comparison results shows the pass rates.

Then, the inconsistency could be from either the 1) EM or 2) TPS.

1) The EM investigation:

The EM was developed using the image data, profiles and FSFs, from one of the facilities.

The facilities images demonstrated small inconsistency in FSFs. The profiles however showed relatively large inconsistency.

Suggestion: Correct the profiles asymmetry manually. The suggestion was applied was no improvement was observed in the EM performance. These could be because 2 out of 6 parameters were optimized using profiles. The rest parameters were optimised using the FSFs. FSFs play more important role in model development while a small inconsistency was observed among FSFs. Then, the image data did not have a large impact on the EM development.

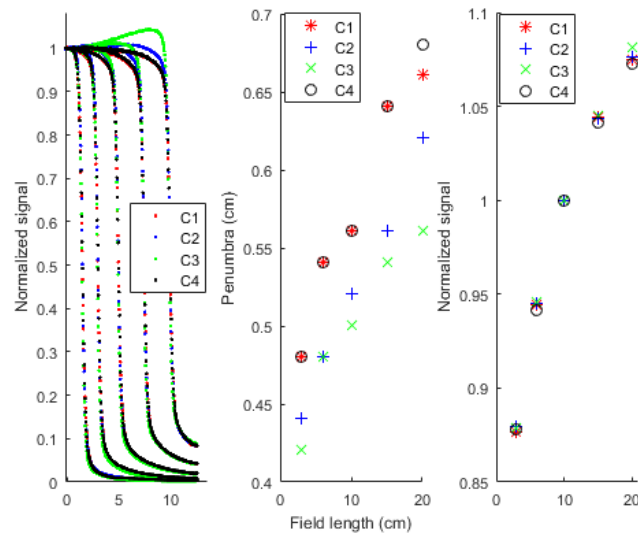


Figure 4-11- Signals from iView images from 4 facilities.

- The EM was trained/optimised using water tank data, profiles and FSFs,

Check for the accuracy of water tank data specially FSFs

Suggestion: replace the water tank FSFs with more accurate measurements

2) The TPS investigation: Calculated FSFs by the facilities' TPS showed relatively large differences between FSFs of the different TPSs.

Suggestion: Average over calculated FSFs by different TPSs and use them to optimize the EM.

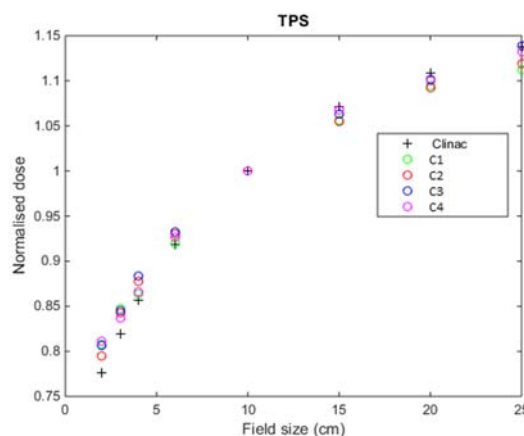


Figure 4-12- Field size factors (FSFs calculated by TPSs of the facilities. The Clinac FSF is a TPS data used for the VM modelling.

All above investigations resulted in training the EM by averaged FSFs of different TPSs calculations.

**Virtual EPID standard phantom audit
(VESPA) for remote IMRT and VMAT
credentialing**

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Abstract

A Virtual EPID Standard Phantom Audit (VESPA) has been implemented for remote auditing in support of facility credentialing for clinical trials using IMRT and VMAT. VESPA is based on published methods and a clinically established IMRT QA procedure, here extended to multi-vendor equipment. Facilities are provided with comprehensive instructions and CT datasets to create treatment plans. They deliver the treatment directly to their EPID without any phantom or couch in the beam. In addition, they deliver a set of simple calibration fields per instructions. Collected EPID images are uploaded electronically. In the analysis, the dose is projected back into a virtual cylindrical phantom. 3D gamma analysis is performed. 2D dose planes and linear dose profiles are provided and can be considered when needed for clarification. In addition, now using a virtual a flat-phantom, 2D field-by-field or arc-by-arc gamma analysis are performed. Pilot facilities covering a range of planning and delivery systems have performed data acquisition and upload successfully. Advantages of VESPA are (1) fast turnaround mainly driven by the facility's capability to provide the requested EPID images, (2) the possibility for facilities performing the audit in parallel, as there is no need to wait for a phantom, (3) simple and efficient credentialing for international facilities, (4) a large set of data points, and (5) a reduced impact on resources and environment as there is no need to transport heavy phantoms or audit staff. Limitations of the current implementation of VESPA for trials credentialing are that it does not provide absolute dosimetry, therefore a Level 1 audit is still required, and that it relies on correctly delivered open calibration fields, which are used for system calibration. The implemented EPID based IMRT and VMAT audit system promises to dramatically improve credentialing efficiency for clinical trials and wider applications found were not major.

Key Words: EPID, audit, IMRT, VMAT, clinical trials

I. INTRODUCTION

Quality assurance for clinical trials in Radiation Oncology is a key component to their success and to the validity of the clinical results. [1] For clinical trial credentialing dosimetry audits of at least two levels are employed to verify (1) the absolute dosimetry of linear accelerators participating in the trial and (2) the capability of the facility to accurately plan and deliver dose distributions relevant to the trial. Dosimetry audits are also performed outside of clinical trials for quality assurance purposes. [2]

Verification of absolute calibration, commonly known as Level I audits, are most often performed as postal audits [2-5]. However, onsite audits, which offer higher accuracy but are significantly more expensive, are available and in some countries, like the UK, they are the preferred option [2, 6-8]. Higher level audits, verifying delivery of specific treatment plans, have been successfully performed over many years using phantoms with embedded detectors shipped to a facility and irradiated there according to provided instructions [9, 10]. Alternatively, onsite higher level audits generally provide the opportunity for higher accuracy of the measurements and more data points to be collected. They also allow for immediate support in case of non-optimal results, saving time and bringing a facility faster to optimal treatment delivery [10-13]. However, onsite audits require an infrastructure [14, 15] which is not widely available and not easy to scale with changing needs. Cost differences between higher level onsite and postal audits depend on phantom costs and turnaround times in postal audits. Here extended turnaround time due to clinical priorities at facilities will increase the need for more phantoms in order to respond to requests for audits. This causes expenses for manufacturing, storage and logistics.

Higher level onsite audits have generally used commercial clinical quality assurance (QA) devices, either the way they were intended to be used or with some alterations in setup and application [13, 16]. These devices are often expensive and delicate electronic systems, which are not necessarily designed for ongoing travel. Special packaging for ground and air transportation needs to be obtained or designed and backup systems in case of breakage or loss need to be considered. Electronic portal imaging devices (EPIDs) are available at most linear accelerators. They have been making their way into machine specific QA [17-20] and patient specific QA [21-27]. Taking the use of EPIDs to the next level, this paper describes implementation of the Virtual EPID Standard Phantom Audit (VESPA) for dosimetric auditing of centres for the Trans Tasman Radiation Oncology Group (TROG).

II. MATERIALS AND METHODS

The VESPA audit methodology is completely remote and does not involve the transport of phantoms or personnel. A facility participating in the audit is provided by TROG with web-based instructions and patient CT data sets. A treatment planner from the facility creates a treatment plan according to the specifics of the clinical trial in question, as specified in the instructions. These trial plans are then transferred to virtual water-equivalent phantoms that are also provided, a flat-phantom for 2D field-by-field or arc-by-arc analysis and a cylindrical phantom for combined field 3D dose analysis.

After planning, the facility delivers the treatment beams to the EPID of the linear accelerator, free air and without a phantom. IMRT plans are delivered in EPID integrated mode, VMAT plans in EPID cine mode. Additionally, calibration fields are delivered to determine central axis and sag characteristics. The calibration plan is provided in DICOM format and consists of a series of open fields at different collimator and gantry angles. The calibration plan can be delivered in about 10 minutes.

The collected EPID images and the treatment planning data are electronically transmitted to TROG for analysis. To facilitate the upload and to improve data quality a directory structure is provided. Once filled with the corresponding files, the structure is packaged into a zip file. Upload is done via a university based server which can be accessed from hospital networks.

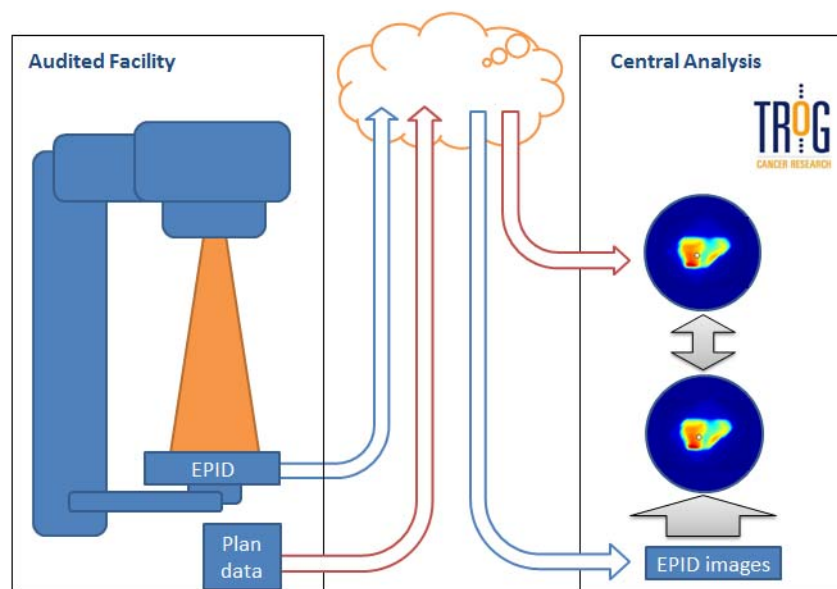


Figure 5-1- Fundamental process: Facility delivers treatment beams free air to EPID and uploads EPID images and plan data. Central analysis calculates dose in virtual phantoms from EPID images and compares with plan data.

To calculate the delivered dose distributions the EPID images are converted to dose-in-water using a previously developed and published algorithm [103, 193, 195-197] and clinically established patient specific pre-treatment IMRT and VMAT QA procedures at the Department of Radiation Oncology, Calvary Mater Newcastle, Australia. The calculated delivered dose distributions are compared to the corresponding distributions uploaded from the facility's treatment planning system. Dose is compared using industry standard gamma analysis [198] in 2D and 3D. 2D dose distributions can be displayed in the analysis software and are included in the audit report. The dose comparisons is assessed using the gamma pass-rate with a global dose difference and distance to agreement of 5%, 3 mm, 3%, 3 mm, 3%, 2 mm, and 2%, 2 mm. Currently the uncertainties and achievable pass-rates of this process in a widespread remote audit are not known. Initial tolerance levels applied for IMRT auditing will be greater than 90% pass-rate for 3%, 3 mm. The VMAT tolerance levels will be greater than 90% pass-rate for 5%, 3 mm to allow for the greater uncertainties in the cine imaging process; however 3%, 3 mm is desired. This might change in the future and can also depend on the application or clinical trial in question and its accuracy requirements.

Audit procedures for the remote centres were generated. They encompass instruction on how to create the necessary calibration images in addition to the images acquired from the tested treatment plans. A specific EPID guide was developed with instructions for each linac and EPID type for EPID positioning, flood and dark field calibration, cine mode setup, and image acquisition. To facilitate (and standardise) this process, DICOM plans for the calibration images are provided for the facility to download and import into the patient management / record and verify system for delivery. The specific centres linac name was inserted into the DICOM plan files to facilitate import.

III. Results

An EPID image to dose conversion model for the Varian Clinac was developed using the method described in King et al.[193] who developed a model for a prototype backscatter shielded EPID. All images used as input to the model parameter optimisation were first backscatter corrected[197]. These images then represent images as though acquired without the support arm present. The method was validated by comparison of EPID images converted to dose compared to measured MapCheck (Sun Nuclear Melbourne, FL, USA) dose planes. EPID images of 36 sliding window IMRT fields were acquired on the aS1000 EPID. These were backscatter corrected and converted to dose at 5 cm depth in water. Gamma comparison using an in-house implementation of the Low method was made to the Mapcheck measurements at 5 cm depth in solid water with 2% of maximum dose, 2 mm criteria and a threshold of 10% of maximum global dose. The resulting pass rates for 14 prostate fields were (mean $\pm$ 1 SD) 99.4

$\pm 1.0\%$ and 22 head and neck fields were $99.3 \pm 1.3\%$. The mean gamma pass rates were 0.31 ± 0.3 and 0.33 ± 0.5 respectively.

For VESPA the methodology was extended for use in an audit environment. This required expansion from a single planning system, single delivery system environment, to multi-vendor planning and delivery equipment. For the planning systems as all systems export 3D DICOM dose file format these were used. No major difficulties were encountered for Eclipse, Pinnacle, Monaco, and iPlan systems. The dose model described above was also tested for applicability to EPID data from Elekta linear accelerators. Results from three Varian units were compared to three Elekta units for Gamma criteria of 3%, 3 mm, 3%, 2 mm and 2%, 2 mm for head and neck IMRT fields with dose calculated at 10 cm depth in a 2D flat water phantom. The mean Varian (Elekta) pass-rates were 99.5% (99.7%); 98.9% (97.8%), and 95.9% (95.3%) for the respective criteria showing that the model can be used for Elekta systems with similar accuracy. The model has not to date been applied to Siemens linear accelerators or EPIDs. For the linear accelerators and record and verify systems the following table describes the major vendor specific issues that arose. The Elekta comments do not refer to the newer iView dose imaging software which has not been assessed for use in the study to date.

Table 5-1- Summary of vendor specific or other issues encountered with the VESPA audit process.

Problem	Solutions
Transfer of images to Mosaiq results in loss of pixel scaling information to obtain integrated dosimetric image.	▪ Varian Clinac - Images saved in Varian format in the cache on the linac used. ▪ Varian Truebeam –Image Processing Service used to store cumulative image frames. Last frame is integrated image in Varian format. Gantry angle for the image is the kV imager angle. ▪ Elekta – Images exported from iView EPID acquisition software in .his format with log file. Log file contains pixel scaling information DICOM images then created at central site for analysis.

Cine Mode imaging limitations or unavailable. Truebeam and Elekta cine imaging does not store dosimetric information. Elekta cine imaging using Perkin Elmer software does not store gantry angle.	- Varian Clinac - Requires large MU (300) for calibration of EPID signal to dose due to missing frames at start and end of acquisition. - Varian Truebeam –Image Processing Service required. This stores cumulative image frames from which cine images can be derived. - Elekta – Perkin Elmer XI service software is required. Individual frames stored. Separate inclinometer for obtaining gantry angle for frames.
Process and procedures	- The most common issue was incomplete data provided such as combined field 3D dose file not provided. - Some images were acquired with zero collimator angle but planned at actual collimator angle, or vise-versa.

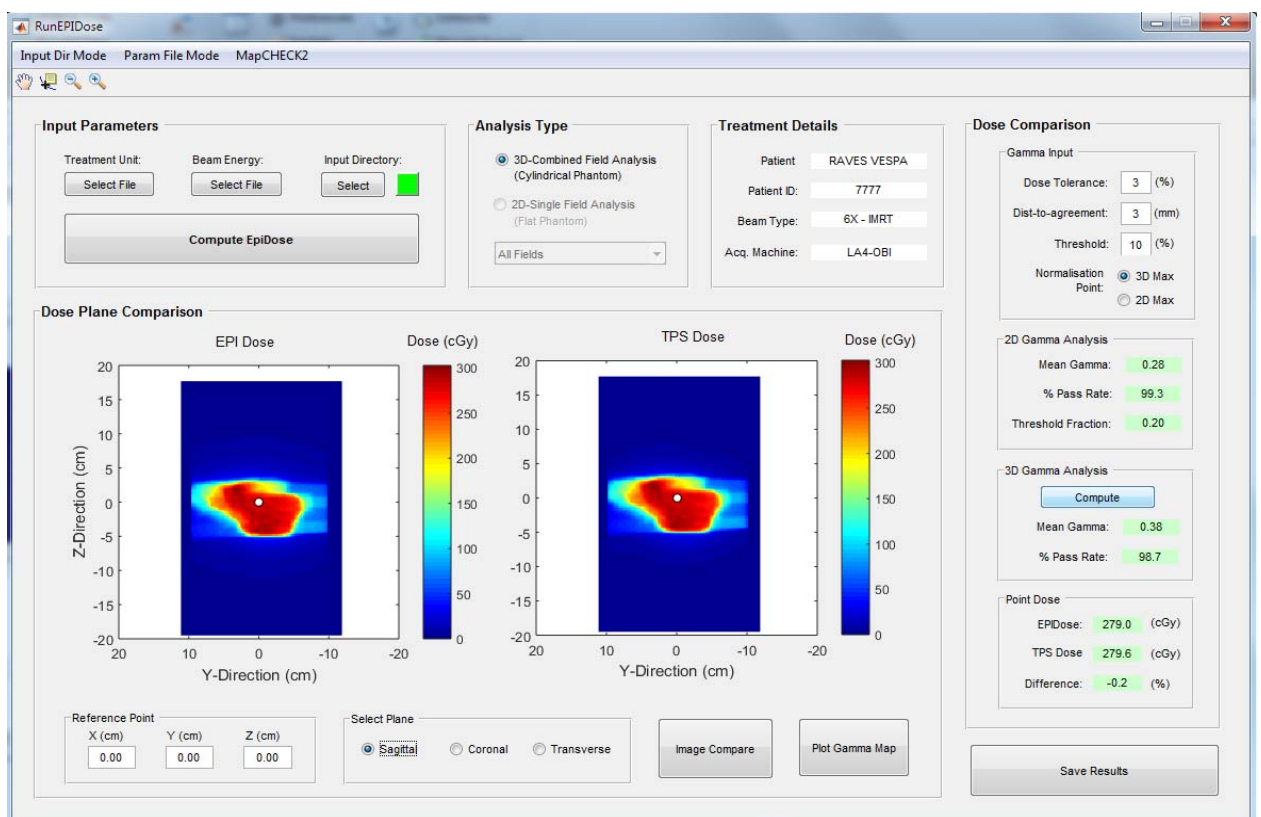


Figure 5-2-Screenshot of the VESPA software showing sagittal dose planes in the virtual cylinder phantom for the RAVES plan.

Following a pilot phase with six facilities each for IMRT and VMAT, VESPA has now been successfully rolled out for routine use for TROG facility credentialing. A total of 30 audits are underway at this point.

IV. Discussion & Conclusions

With VESPA we have implemented an EPID based, remote audit, where data acquisition is done by facility staff, results are transferred electronically and analysis is performed centrally in Newcastle, Australia.

The advantages of VESPA are (1) a fast turnaround, which is mainly driven by the facility, as the instructions are available at any time and the facility can work at their preferred pace, (2) the possibility for many facilities performing the audit in parallel, as there is no need to wait for a phantom to become available, (3) the availability of a large set of data points with higher accuracy than passive detectors, and (4) a reduced impact on resources and the environment as there is no need to ship heavy phantoms or transport audit staff.

Limitations of the current implementation of VESPA are that it does not provide absolute dosimetry, therefore a Level 1 audit still required, and that it relies on the open calibration fields to be delivered correctly, as they are used for calibration of the system. Other disadvantages are the implementation of EPID imaging by the vendors varies and the lack of transfer of pixel scaling information to Mosaic. Cine imaging implementation varies widely between the vendors. A potential future application of VESPA could be to determine the dose to the provided virtual patient CT dataset from the EPID images and perform dose-volume-histogram analysis.

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Remote dosimetric auditing for intensity modulated radiotherapy: A pilot study

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Abstract

Background and Purpose: Electronic portal imaging devices (EPIDs) can be used to reconstruct dose inside a virtual phantom. This work aims to study the feasibility of using this method for remote dosimetry auditing of clinical trials. **Materials and Methods:** Six centres participated in an intensity modulated radiotherapy (IMRT) pilot study of this new audit approach. Each centre produced a head and neck (HN) and post-prostatectomy (PP) trial plan and transferred the plans to virtual phantoms to calculate a reference dose distribution. They acquired in-air images of the treatment fields along with calibration images using their EPID. These data were sent to the central site where the images were converted to 2D field-by-field doses in a flat virtual water phantom and to 3D combined field doses in a cylindrical virtual phantom for comparison with corresponding reference dose distributions. Additional test images were used to assess the accuracy of the method when using different EPIDs. **Results:** Field-by-field 2D analysis yielded mean gamma pass-rates of 99.6% ($\pm 0.3\%$) and 99.6% ($\pm 0.6\%$) for HN and PP plans respectively (3%/3 mm, doses greater than 10% global max). 3D combined field analysis gave mean pass-rates of 97.9% ($\pm 2.6\%$) and 97.9% ($\pm 1.8\%$) for the HN and PP plans. Dosimetry tests revealed some field size limitations of the EPIDs. **Conclusions:** The remote auditing methodology using EPIDs is feasible and potentially an inexpensive method.

Key Words: Remote auditing, EPID dosimetry, EPID modelling, IMRT, pilot audit

I. INTRODUCTION

In radiotherapy clinical trials, the complex nature of planning and delivery systems may result in variations in the dose deliveries among participant centres. Studies have demonstrated the clinical relevance of poor quality planning and treatment outcomes [1]. The benefit of rigorous pre-treatment patient specific quality assurance (QA) and external dosimetric audits for clinical trials has been well demonstrated [2, 3]. On-site pre-treatment QA for intensity modulated radiotherapy (IMRT) has been shown to not always detect discrepancies between planning and delivery systems found in external audits [4]. Participating centres' treatment delivery should be assessed to reduce variability thus improving the reliability of trial results. Conventionally, an independent centre performs the audit by site visit(s) or by mailing phantoms and dosimeters [5-7]. The most comprehensive audit is an 'end-to-end audit' that tests the full treatment chain from CT scanning to delivery using an anthropomorphic phantom. For example, the Imaging and Radiation Oncology Core (IROC) sends a head and neck phantom to participant centres. It has used pass criteria of 7% point dose difference in the planning target volume (PTV) and 4 mm distance to agreement in the high dose gradient area [7]. While successfully established, the mailing audit approach is limited by the resources and costs involved in transporting equipment to and from each centre. As the measurement is responsibility of the local physicists, phantom and dosimeter set-up errors can result in measurements out of tolerance and therefore the need for repetition.

Site-visit audits, on the other hand, are performed by external auditors which reduce set-up errors and increase the consistency of measurements. More recent approaches targeting specific technology such as volumetric modulated arc therapy (VMAT) have used a 'TPS planned audit' approach where centres use benchmark CT data sets and planning instructions to produce a local treatment plan. This plan is then transferred to CT data sets of the audit QA phantoms or 2D/3D detectors, then the dose distribution is calculated and compared to measurements performed during the site-visit. Examples of this approach include an audit that used a head and neck IMRT plan transferred to a solid water block to perform film measurements, where 75% of the films passed a gamma criteria of 4%,3 mm [8]. Another audit used the Octavius II phantom with PTW array to measure 2D dose planes [9]. This audit found that 42 out of 43 trial plans achieved greater than 95% pass rates at 3%,3 mm criteria for measured dose planes. However, site visits can be expensive, time-consuming and logistically difficult to perform [10]. Some alternative remote methods have been proposed to reduce cost and increase efficiency. The European Organisation for the Research and Treatment of Cancer (EORTC) in conjunction with IROC, termed the use of "institutional virtual phantoms". In this method, participant centres sent CT data sets of their institutional phantom and their measured and planned dose

distributions to the auditing site for analysis with standardised software. Although 6 out of 12 centres demonstrated more than 95% pass rates at 3%, 3mm criteria, 1/3rd of the centres could not be analysed centrally due to the variation of employed techniques and dosimeters [11].

Recently we proposed a novel concept to perform remote dosimetric auditing for clinical trials using the 'TPS planned audit' model and EPID measurements [12]. In that work an overview of the concept was presented. In current work, the specific details of the auditing method are outlined including EPID image and calibration plan acquisition details, image processing and conversion to dose methodology. The results from a pilot study of IMRT for six centres are presented including a separate analysis of the dose conversion model performance for each centre using open-field image data. The method combines the cost and efficiency benefit of remote audits with a standardised measurement and analysis process using EPID. The approach is termed the Virtual Epid Standard Phantom Audit (VESPA) and is based on EPID to dose conversion model [13]. In this paper we investigate the feasibility of this concept for IMRT auditing using data from six participant pilot centres.

II. MATERIALS AND METHODS

A. Equipment

Six centres equipped with linear accelerators (linacs) from two vendors participated in this pilot study. Vendor 1 was Varian (Varian Medical Systems, Palo Alto, CA) with aS1000 type EPIDs, centres A, B and C and Vendor 2 was Elekta (Elekta AB, Stockholm, Sweden) with iViewGT EPIDs, centres D, E and F. Three different treatment planning systems (TPSs) were used. Comprehensive audit instructions were provided to the centres including a separate EPID guide to assist with correct calibration and operation. The Trans-Tasman Radiation Oncology Group (TROG) supplied IMRT head and neck (HN) and post-prostatectomy (PP) trial benchmarking plan instructions and CT data sets. Prescriptions, PTV and OAR constraints for both cases are shown in Supplementary table 1. CT datasets of two standard virtual water-equivalent QA phantoms were also provided; a virtual flat phantom (VFP) and a virtual cylindrical phantom (VCP). The VFP was 41 cm in length (superior-inferior direction) and 43 cm × 35 cm in cross-section and the VCP was 40 cm in length and 20 cm diameter in cross-section. The VESPA process has been summarised in Supplementary figure 1.

B. EPID to dose conversion method

2D dose planes in VFP

The conversion of EPID signal to dose at 10 cm depth within the virtual phantom was performed using an in-house software, based on the method of King et al. [13], and developed at Calvary Mater Newcastle Hospital (CMNH). The software included a model that does not require

parameter adjustment by the participating centres. However, an individual machine specific file, using information provided by each centre, was used to refine the model and adapt it to each machine/delivery type. The machine specific file used calibration images from each centre to determine central axis coordinate on the EPID and EPID sag as described below.

Two sets of dose conversion parameters have been developed for images acquired with a Varian aS1000 EPID and aS1200 EPID and validated by comparison to 2D dose-planes measured with MapCheck diode array (Sun Nuclear Corporation, Melbourne, FL) [14]. This study used the model developed for Vendor 1. The model required an additional EPID support-arm backscatter correction for the Varian images [15]. It was benchmarked for 14 prostate fields and 22 head and neck fields with mean gamma pass-rates of respectively 99.4% (SD 1.0%) and 99.3% (SD 1.3%) at 2%,2 mm criteria. The model was applied to images from Vendor 2 for the first time in this study. In conjunction with the IMRT field images, a series of dosimetry test images with varying field size were also acquired in this study to examine the model performance for auditing of both vendors.

3D dose distribution in VCP

For calculation of 3D dose in the VCP, the method of Ansbacher was used [16]. Images acquired at actual gantry angles were converted to planar dose at 10 cm depth at isocentre in the flat phantom as described above. The planar dose was converted to 3D dose inside a VCP by multiplying it by a 2D off-axis correction matrix and applying an exponential percentage depth dose (PDD) and buildup factor. The dose was calculated for each individual image at the actual gantry angle then it was added to the total dose matrix to give the combined 3D dose distribution. The combined dose was stored in coordinates of the TPS dose matrix so that the dose distributions could be quantitatively compared.

C. Treatment Planning

Each participating centre planned a prostate and a head neck trial case following the benchmarking instructions on the provided patient datasets. A dose of 70 Gy in 35 fractions were prescribed to the head and neck case with $D_{95\%}$ constraint for the PTVs and the PP plan prescribed a total dose of 64 Gy in 32 fractions using $D_{98\%}$ for the PTV; both being delivered at 6 MV energies. They then transferred the trial plan onto the VFP at perpendicular incidence for individual field analysis and onto the VCP at actual treatment gantry angles for the combined 3D dose distribution (Supplementary figure 1). The isocentre was placed at 10 cm depth at 90 cm source to surface distance (SSD), which was at the centre of the VCP. These verification plan doses were then exported in DICOM format.

A DICOM-RT format TPS calibration plan was also provided. This plan was only used for calculation of doses in the TPS. The plan consisted of a series of open jaw defined field sizes to calculate dose in the VFP, and a $10 \times 10 \text{ cm}^2$ field to calculate dose in the VCP, all at gantry zero incidence. The TPS open field doses were compared to doses derived with the model acquired from open field images to investigate model performance for each centre as described below. The dose at isocentre for the VCP provided a calibration dose for the 3D model.

D. EPID measurements

Integrated images of the IMRT fields were acquired both at gantry vertically downward and at actual gantry angles. All images were acquired with the clinical or QA mode operating. The images from Vendor 2 were acquired at 160 cm source to EPID distance and exported in Hamamatsu Image Sequence (HIS) format as the DICOM export does not retain pixel scaling information. Images from Vendor 1 however were acquired at 105 cm source to EPID distance. An EPID calibration plan in DICOM-RT format was provided for the centres. This plan consisted of a series of $10 \times 10 \text{ cm}^2$ fields at 45° gantry angles to provide data to 1) calibrate EPID response to dose; 2) determine EPID central axis position at gantry zero; and 3) correct EPID sag with gantry angle. The fields and analysis method are described below.

As this method was not previously applied to the systems from Vendor 2, a dosimetry test plan was also provided in DICOM-RT format consisting of a series of open jaw defined fields of size 2×2 , 3×3 , 4×4 , 6×6 , 10×10 , 15×15 , 20×20 , $25 \times 25 \text{ (cm}^2\text{)}$ to compare to TPS doses following image to dose conversion. The plan also included a set of $10 \times 10 \text{ cm}^2$ fields with different monitor unit (MU) settings for EPID linearity assessment.

The centres exported their images and TPS doses and uploaded them via the cloud to the central site for assessment. HIS format images were converted to DICOM format. For each centre, the following procedures were performed to determine the centre specific machine parameter file before dose was calculated in the virtual phantoms from the EPID images.

Coordinate system

Two EPID images of a $10 \times 10 \text{ cm}^2$ field with 90° and 270° collimator angles at gantry zero (gantry pointing vertically down) were used to determine the sub-pixel central axis (CAX) location on the EPID and hence an EPID coordinate system referenced to radiation isocentre. The field edges (50% dose points) of each image were determined using linear interpolation between pixels. The average mid-point of the two images gives the CAX location independent of jaw positioning [13].

EPID Sag correction

To characterise EPID sag, several methods have been presented [17]. In the current study, EPID images of a 10×10 cm² field were acquired at either 45° or 90° gantry angles. The field mid-point location was determined on each image as described above and compared to the mid-point at gantry zero to calculate sag relative to gantry zero (where the CAX position is known). The difference versus gantry angle showed best fit with a first order Fourier series, $Sag(\theta) = a_0 + a_1 \cos(\theta) + b_1 \sin(\theta)$. This fit was then used to correct the coordinate system for each acquired image depending on its gantry angle, Supplementary figure 2.

Calibration factor

The dose conversion method required a calibration factor [18]. Images of a 10×10 cm² field were acquired with 20 MU for Vendor 2 systems and 100 MU for Vendor 1 systems. This difference was related to the methods employed for IMRT image acquisition on these systems. Deshpande et al. demonstrated that calibrating pixel to dose at 100 MU for linacs from Vendor 2 would introduce calibration errors of 2-4% for the typical range of IMRT segment (4-20) MUs and recommended 20 MU for calibration [19]. The converted dose value in a region of interest at central axis was compared to the corresponding TPS value for calibration factor determination.

Dose analysis

IMRT images of each individual field delivery acquired at gantry zero were used to reconstruct planar dose at 10 cm depth in the VFP and compared to TPS calculations with 2D gamma analysis. The images acquired at actual gantry angles were used to reconstruct the 3D dose distribution in the VCP and compared to TPS calculations with 3D gamma analysis. An in-house developed gamma algorithm was used for the dose comparison. All doses above 10% of the maximum dose were assessed with a search region of 6 mm radius. The gamma function used a global dose difference (DD) criteria defined by percentage of maximum dose of each measured image. For individual fields, 2D gamma analysis was employed while for combined dose distributions, 3D gamma analysis was used. The dose comparisons in this work were performed with 2%,2 mm, 3%,2 mm and 3%,3 mm criteria.

To gain insight into the consistency of response and model performance and uncertainties for the different linac vendors, the dose converted from EPID images of open fields calculated in the VFP was compared to TPS calculations for each centre. Dose at isocentre at 10 cm depth was modelled for a set of square field images with different sizes, 2×2 , 3×3 , 4×4 , 6×6 , 10×10 , 15×15 , 20×20 , 25×25 (cm²), then compared with their corresponding TPS dose.

Finally, to ensure that the EPIDs from different centres were responding linearly to dose and to assess inter-centre response differences, each centre acquired a set of 10×10 cm² images at

incremental MU irradiations, (5-400) MU. The mean integrated pixel value (IPV) was calculated for 11×11 central pixel region of each image and normalised to the IPV at 100 MU.

III. RESULTS

Audit: 2D dose planes in VFP/3D dose planes in VCP

The gamma results for the pilot study audit are shown in Table 1. For the 2D field-by-field analysis, the mean of all centres was 95.6% at 2%,2 mm criteria with the lowest being 91.6% for Centre A. For the 3D combined dose analysis, the results were lower with the lowest being Centre E with 92.7% at 3%,3mm criteria. Table 2 shows examples of an axial plane of the 3D dose distributions in the VCP for both the HN and PP plans.

Table 6-1- Mean (with standard deviation) gamma pass rates of the pilot centres for head and neck (HN) and post-prostatectomy (PP) individual fields. 2D dose planes were compared at 10 cm depth in the VFP for each field.

Centres	HN pass-rate ($\pm$ SD) (%)			PP pass-rate ($\pm$ SD) (%)		
	3%/3mm	3%/2mm	2%/2mm	3%/3mm	3%/2mm	2%/2mm
A	99.2 (1.3)	97.9 (2.0)	91.6 (5.9)	100.0 (0.0)	99.9 (0.2)	99.4 (0.5)
B	100.0 (0.0)	100.0 (0.0)	99.7 (0.2)	100.0 (0.0)	100.0 (0.0)	99.8 (0.1)
C	99.3 (0.9)	98.8 (1.1)	96.5 (2.0)	99.8 (0.2)	99.7 (0.4)	99.2 (0.8)
D	99.5 (0.3)	97.4 (1.4)	94.6 (2.4)	99.4 (0.7)	98.0 (1.3)	93.4 (2.9)
E	99.8 (0.2)	96.7 (2.5)	93.2 (3.2)	99.8 (0.3)	98.9 (0.8)	95.8 (2.3)
F	99.8 (0.2)	99.3 (0.4)	98.2 (1.0)	98.5 (0.3)	96.1 (0.4)	90.7 (0.5)
Mean (SD)	99.6 (0.3)	98.4 (1.2)	95.6 (3.1)	99.6 (0.6)	98.8 (1.5)	96.4 (3.8)

Table 6-2- Mean gamma pass rates (with standard deviation) of the pilot centres for head and neck (HN) and post-prostatectomy (PP) combined dose distributions in the VCP, 3D dose gamma analysis.

Centres	HN pass-rate ($\pm$ SD) (%)			PP pass-rate ($\pm$ SD) (%)		
	3%/3mm	3%/2mm	2%/2mm	3%/3mm	3%/2mm	2%/2mm
A	99.8	97.7	87.5	98.7	93.4	79.2
B	98.9	96.9	89.4	98.7	98.0	95.5
C	98.1	95.7	87.6	99.8	99.3	96.8
D	98.9	87.3	72.0	95.6	92.2	82.3
E	92.7	77.6	54.4	98.9	93.6	78.5
F	99.1	94.0	79.6	95.8	88.7	77.0
Mean (SD)	97.9 (2.6)	91.5 (7.8)	78.4 (13.5)	97.9 (1.8)	94.2 (3.9)	84.9 (8.9)

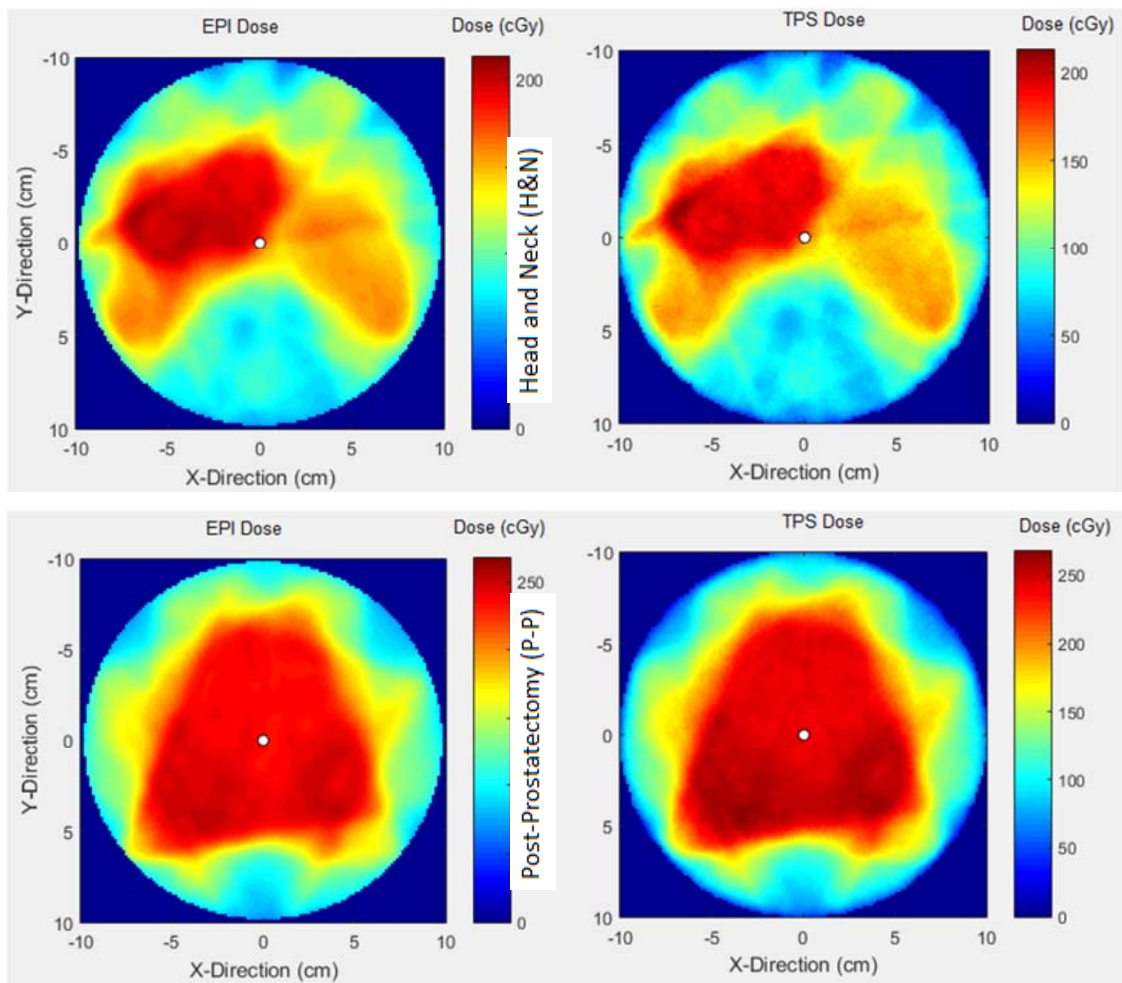


Figure 6-1-An example of an axial 2D plane of the head & neck (top row) and post-prostatectomy (bottom row) VCP doses. Left and right images show respectively delivery and treatment planning system (TPS) dose for each plan from centre F.

Dosimetry Verification

Figure 2 demonstrates the EPID converted dose versus field size for open jaw defined fields compared to TPS dose for each centre. The Vendor 1 centres showed consistent differences between EPID dose and TPS dose. The image converted dose was similar to the TPS calculation at fields smaller than $10 \times 10 \text{ cm}^2$ and slightly higher at larger fields. Similarly, the Vendor 2 centres showed consistent differences with these being larger than the differences for Vendor 1. The image converted dose was slightly lower than the TPS calculation at fields smaller than $10 \times 10 \text{ cm}^2$ and slightly higher at larger fields. Centres D and E demonstrated large differences at the largest field size, $25 \times 25 \text{ cm}^2$. The EPID response versus MU is shown in Figure 3. As expected the response of the EPID was not completely linear. Apart from the response at small MUs, the Vendor 1 centres showed similar response while for the Vendor 2 centres, centre D demonstrated different response than the other two centres.

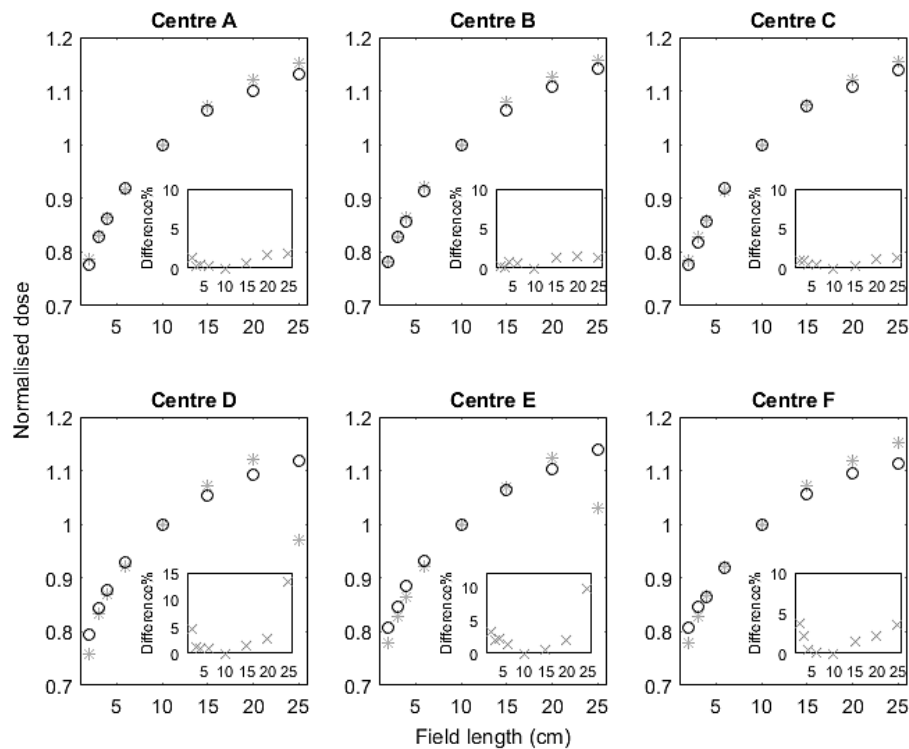


Figure 6-2-Calculated dose at isocentre in the VFP for jaw defined open field sizes using EPID images (stars) and TPS (circles) for different centres. The insets correspond to the difference defined by $(D_{\text{model}} - D_{\text{TPS}}) / D_{\text{TPS}} \%$. All values were normalised to the values of $10 \times 10 \text{ cm}^2$ field size.

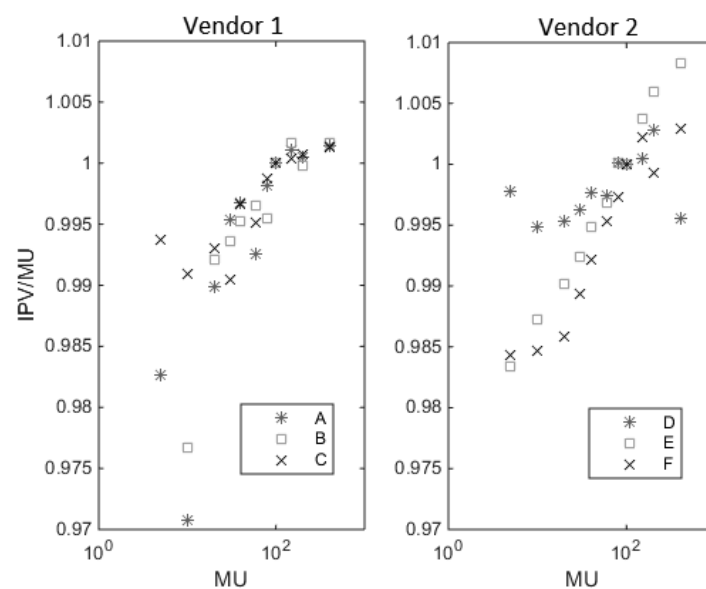


Figure 6-3- Central integrated pixel value (IPV) per MU versus MU for $10 \times 10 \text{ cm}^2$ images acquired at different MU settings. All IPV values were normalised to the values of 100 MU.

IV. DISCUSSION

The 2D field-by-field analysis resulted in mean (of all centres) gamma pass rates over 99.5% at 3%, 3 mm criteria and over 95.5% at 2%, 2 mm. The EPID signal to dose in water conversion model was not adapted for individual centres. The open field comparisons in Figure 2 suggest

that improvement to the results could potentially be made by deriving an Elekta specific set of model parameters. This could explain the slightly lower gamma results obtained for the Elekta systems in the study.

For 3D dose analysis of the centres, only 1 of the 12 plans had a gamma pass rate below 95% at 3%,3 mm criteria. The pass rates were lower for the 3D analysis reflecting the larger uncertainty in the 3D model where depth-dose modelling is required. The current algorithm does not use vendor or centre-specific beam information. A detailed investigation into the contributing uncertainty components of the VESPA model when implemented across multiple types of linacs is underway and shall be reported separately.

As Figure 2 demonstrates, a large discrepancy is observed at large field sizes for two centres from Vendor 2. The reason for this could be an EPID signal artefact introduced by scatter close to the peripheral electronics. The imager response from centre D was re-measured and it confirmed that the artefact exists for fields larger than $23 \times 23 \text{ cm}^2$, Supplementary figure 3. This did not influence the gamma results in this study as smaller field sizes were used for the HN and PP fields. Future studies will restrict measurements for systems from Vendor 2 to a maximum field size of $23 \times 23 \text{ cm}^2$. Furthermore, the EPID response versus MU demonstrated non-linearity at low MUs for the EPIDs. This could be due to the failure of the acquisition system in integrating all EPID frames [20], however the magnitude varies for the centres. Further investigations and data are required in order to determine the causes of these variations. A centre-specific calibration to dose that varies with irradiated MU could also be employed.

The VESPA method provides a potentially inexpensive and rapid method to perform dosimetric auditing for specific assessments of new technologies. To be consistent with previous auditing methodologies, each centre produced their own treatment plan using their own planning techniques. This can introduce variation in the deliveries compared with providing each centre with an identical plan. However, technically it would not be possible to deliver an identical plan on different vendor systems, and this auditing approach assesses the individual centres planning methods. The measurements can be performed in 2-3 hours while one calibration process suffices, if the measurements are performed in one session.

However, VESPA is not as comprehensive as an ‘end-to-end’ audit and cannot assess absolute beam output, beam profile or inhomogeneity modelling. In some cases, site visits or ‘end-to-end’ audits may still be preferable. The VESPA method has not yet been implemented for flattening-filter-free deliveries or small-field auditing. The method follows the TPS planned audit approach which specifically targets a new technique such as IMRT or VMAT. It aims to combine the cost effectiveness of, for example, the EORTC “institutional virtual phantoms” method [11] with a more standardised approach to the dosimetry and analysis. In principle, it

is attempting to mimic audits of IMRT performed with a pre-treatment verification type dosimeter but with extension to 3D dose estimation [9]. For the 3D dose volume analysis in the VCP the results are lower. It is likely due to uncertainties in the modelling of percentage depth dose as a single depth-dose model was used for these analyses. Improvement using a field-size specific and/or centre-specific depth dose model could be explored. Another approach that may improve results would be the use of a larger diameter virtual phantom to reduce high dose regions near the phantom surface.

The applied gamma tolerances should consider the expected dosimetric uncertainties of the treatment chain as well as the audit method [5]. Clark et al. have suggested 3%/3 mm and 4%/3 mm criteria to compare respectively field-by-field and combined field dose distributions [8] and some recent studies have used 7%/4 mm criteria for end-to-end audit frameworks [7, 11]. The current study however analysed the results at 3%/3 mm, 3%/2 mm and 2%/2 mm criteria for analysis of both 2D and 3D dose distributions. The gamma pass rates were higher compared to the results from a similar audit based, albeit with a smaller number of centres [9]. The results suggest that analysis of the 3D dose delivery is feasible at 3%,3 mm which is stringent compared with other audit methods. Improvement to depth-dose modelling may allow this criteria to be tightened. It may also be possible to assess field-by-field deliveries at 2%,2 mm criteria for higher sensitivity. However a sensitivity analysis of the method should be performed to ensure that clinically significant dosimetric errors can be detected. We are currently in the process of assessment of the sensitivity at our centre and comparison of this with a dose-volume histogram approach instead of a gamma assessment.

In conclusion, this pilot study assessed the methodology and feasibility of the VESPA method for remote verification of IMRT deliveries performed at different centres. Results of the current study demonstrate the feasibility of this method for clinical trial dosimetry auditing. The remote nature of the method promises a less expensive and more efficient alternative to those currently available. Further assessment and subsequent improvements will establish the method's capabilities as an alternative to current IMRT and VMAT dosimetric audit methods.

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Supplementary files:

Tables

Table 6-3- A summary of planning constraints for the two benchmarking plans of the pilot study: head and neck (HN) and post-prostatectomy (PP) plans.

PTV					
PP: Total Dose:V100% = 64 Gy			HN: Total Dose:V100% = 70 Gy		
Criteria	Dose range (Gy)	Value (Gy)	Criteria	Dose range (Gy)	Minor & Major violation (Gy)
D98%	>95%XV100%	>60.8	D95%	PTV70>=66.5	65.1<D<66.5 & D<65.1
				PTV67>=63.65	60.3<D<63.65 & D<60.3
				PTV63>=59.85	58.6<D<59.85 & D<58.6
				PTV54>=51.30	45.9<D<51.3 & D<45.9
Mean dose (Dmean)	-1%<Dmean<2%	63.4<Dmean<65.3	Median Dose for PTV70 (Dmedian)	68.8<Dmedian<71.4 (±2% of 70 Gy)	
Maximum D2%	<107%XV100%	<68.48	Maximum D2% for PTV70	<77.0	77<D<80.5 & D<80.5
Normal tissues					
Rectum: V60Gy & V40Gy	<40% & <60%	<24 & <24	(D1%) for Spinal cord & PRV Spinal cord	-	<45 & <50
Femoral heads: V35Gy, V45Gy and 60 Gy	<100%, <60% and <30%	<35, <27 & <18	(D1%) for Brachial plexus	-	<66

Table 6-4- Details of the participants in the pilot study. TPS: treatment planning system. EPID: electronic portal imaging device.

Centre	(A)	(B)	(C)	(D)	(E)	(F)
Linac type	Varian Trilogy	Varian iX	Varian Trilogy	ELEKTA Synergy	ELEKTA Synergy	ELEKTA Axesse (+agility head)
TPS algorithm	Pinnacle CCC	Eclipse AAA	Eclipse AAA	Pinnacle CCC	Monaco MC	Monaco MC
EPID model	aS1000	aS1000	aS1000	iViewGT	iViewGT	iViewGT
Record & Verify system	Aria	Aria	Aria	MOSAIQ	MOSAIQ	MOSAIQ

Figures

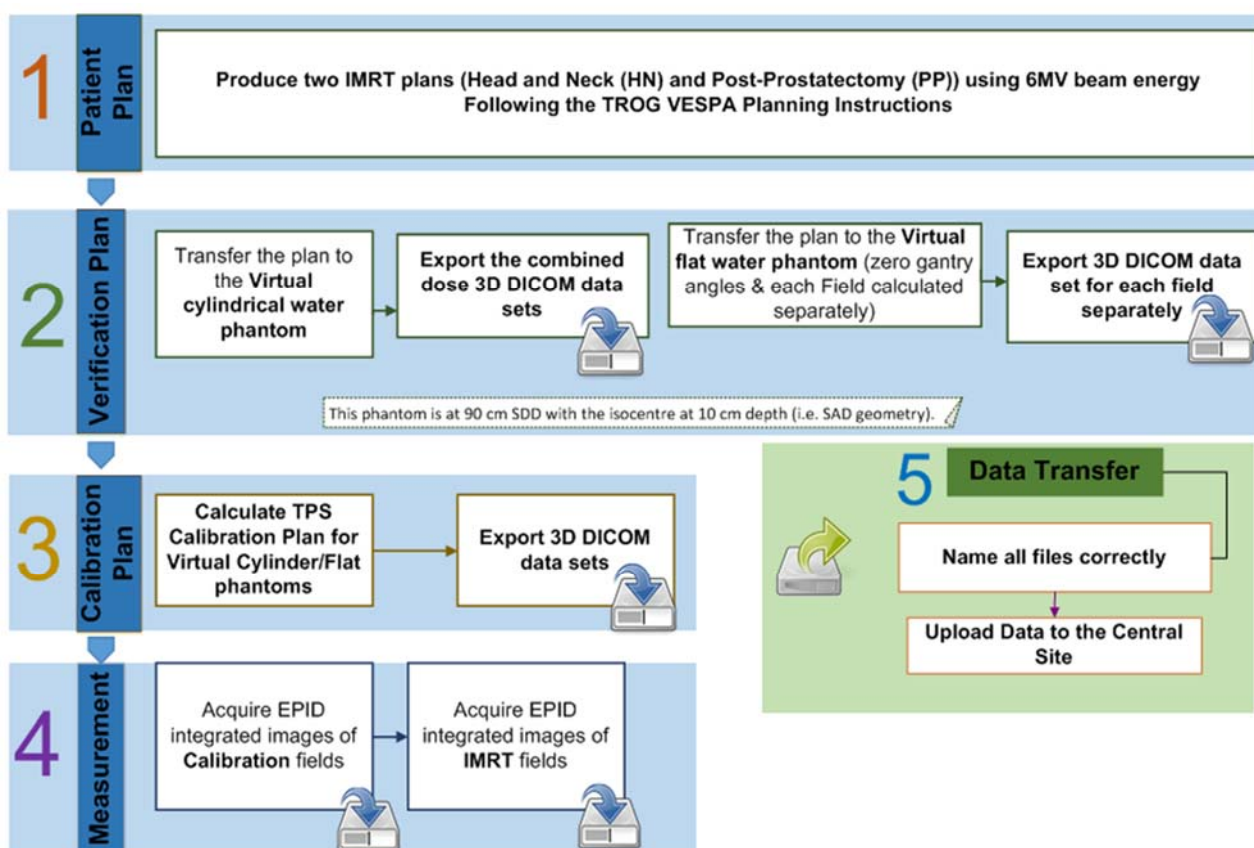
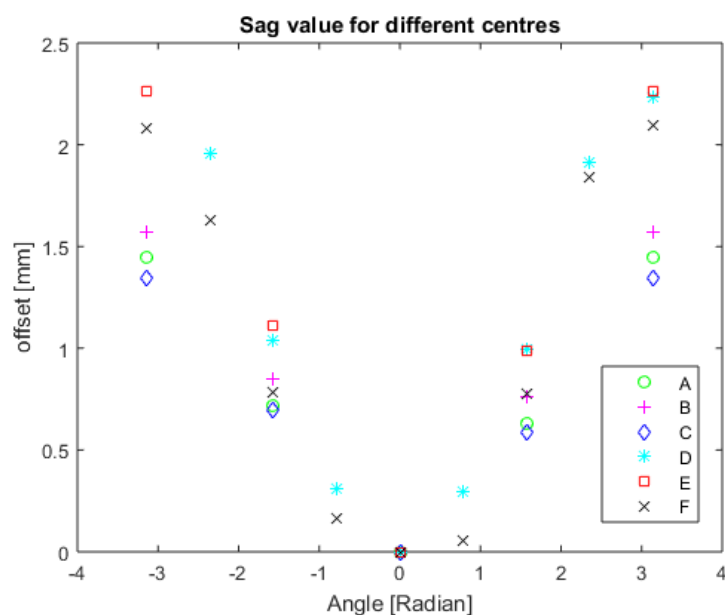


Figure 6-4-An overview of VESPA.

Figure 6-5-In-plane pixel offset (i.e. sag) of 10×10 cm² images versus gantry angle.

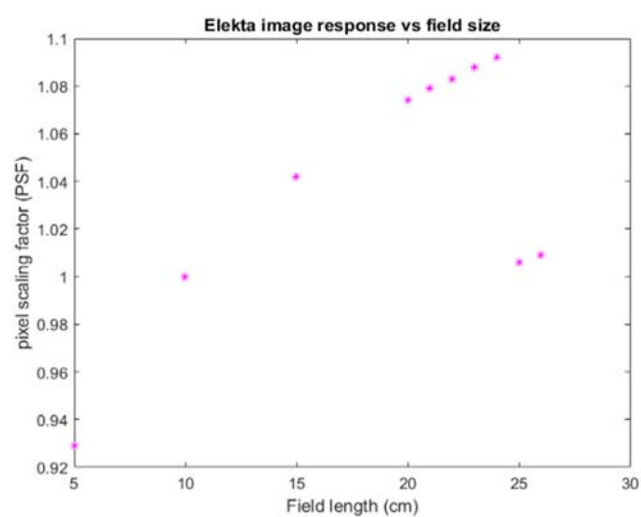


Figure 6-6- Imager response at different field sizes for an ELEKTA imager (Centre D).

A remote EPID-based dosimetric TPS-planned audit of centers for clinical trials: outcomes and analysis of contributing factors

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Abstract

Background: A novel remote method for external dosimetric TPS-planned auditing of intensity modulated radiotherapy (IMRT) and volumetric modulated arc therapy (VMAT) for clinical trials using electronic portal imaging device (EPID) has been developed. The audit has been applied to multiple centers across Australia and New Zealand. This work aims to assess the audit outcomes and explores the variables that contributed to the audit results. **Methods:** Thirty audits were performed of 21 radiotherapy facilities, 17 facilities underwent IMRT audits and 13 underwent VMAT audits. The assessment was based on comparisons between the delivered doses derived from images acquired with EPIDs and planned doses from the local treatment planning systems (TPS). Gamma pass-rate (GPR) and gamma mean value (GMV) were calculated for each IMRT field and VMAT arc (total 268 comparisons). A multiple variable linear model was applied to the GMV results (3%/3mm criteria) to assess the influence and significance of explanatory variables. The explanatory variables were Linac-TPS combination, TPS grid resolution, IMRT/VMAT delivery, age of EPID, treatment site, record and verification system (R&V) type and dose-rate. Finally, the audit results were compared with other recent audits by calculating the incidence ratio (IR) as a ratio of the observed mean/median GPRs for the remote audit to the other audits. **Results:** The average (± 1 SD) of the centers' GPRs were: $99.3 \pm 1.9\%$, $98.6 \pm 2.7\%$ & $96.2 \pm 5.5\%$ at 3%,3mm, 3%,2mm and 2%,2mm criteria respectively. The most determinative variables on the GMVs were Linac-TPS combination, TPS grid resolution and IMRT/VMAT delivery type. The IR values were 1 for seven comparisons, indicating similar GPRs of the remote audit with the reference audits and >1 for four comparisons, indicating higher GPRs of the remote audit than the reference audits. **Conclusion:** The remote dosimetry audit method for clinical trials demonstrated high GPRs and provided results comparable to established more resource-intensive audit methods. Several factors were found to influence the results including some effect of Linac-TPS combination.

Key Words: auditing, dosimetry, electronic portal imaging device

I. INTRODUCTION

Starting in the mid-1990s, multileaf collimators (MLCs) were introduced to linear accelerators (linacs) to deliver a highly conformal dose to the patients. Inverse planning algorithms were added to treatment planning systems (TPSs) to plan the delivered dose when MLCs were used to modulate the profiles of beams. Intensity modulated beams formed the foundation of intensity modulated radiotherapy (IMRT) and volumetric modulated arc therapy (VMAT) deliveries [1]. Machine and patient specific quality assurance (QA) measurements are taken by local physicists to ensure accuracy and stability of IMRT/VMAT deliveries. The European Society for Radiotherapy and Oncology (ESTRO) recommends an additional external audit for independent verification [2]. Additionally, in the context of clinical trials, a dosimetry audit provides a controlled environment to minimize dependency of the outcome on stochastic and systematic errors that can reduce the trial cost and enhance the outcome reliability [3]. Conventionally, an auditing center performs the assessment by site visit(s) or by mailing phantoms and dosimeters [4, 5].

Remote auditing can significantly reduce the audit costs while enhancing the efficiency. Recently, a novel approach was introduced to remotely audit IMRT and VMAT deliveries [6, 7]. The method is termed the Virtual EPID Standard Phantom Audit (VESPA) and it is based on images from electronic portal imaging devices (EPIDs) and image to dose conversion models [8-11]. In VESPA, the audit center provides instructions and CT data for participants to produce benchmarking plans using their local TPS. These plans are then transferred to two provided virtual water phantoms and the doses exported. The participants deliver the dose in air to their EPID and send the corresponding images together with calibration images and their planning data to the audit center. The image signals are converted to dose in the virtual phantoms using in-house developed software. The method combines the cost and efficiency benefits of remote audits with a standardized measurement and analysis process using EPIDs. Details of the method and feasibility of the approach have been reported in a pilot study for six centers [7].

This work aims to assess the VESPA audit outcomes and explores the contribution of several explanatory variables to the overall outcomes of the audit. Results are presented for 30 audits from 21 treatment centers in terms of gamma analysis for multiple criteria. A multi-variable model was developed to understand whether the audit was sensitive to differences in equipment of the centers or other factors. Finally, the audit outcome was compared with other recent audits to assess whether the VESPA audit is consistent with conventional audit approaches.

II. MATERIALS AND METHODS

2.1. Equipment

Participants were radiotherapy centers from Australia and New Zealand who were already treating patients with IMRT or VMAT and required credentialing for clinical trials by the Trans-Tasman Radiation Oncology Group (TROG). Additional file 1 provides details of the centers and their planning and treatment equipment. Of the 21 centers, 17 participated in the IMRT audit and 13 in the VMAT audit. TROG supplied a head and neck (HN) and a post-prostatectomy (PP) trial benchmarking plan case including CT datasets and planning instructions (TROG trials 12.01 HPV and 08.03 RAVES). Additionally CT datasets of two standard virtual water-equivalent QA phantoms were also provided; a virtual flat phantom (VFP) of 30 cm height, 40 cm width, 40 cm length and a virtual cylindrical phantom (VCP) of 20 cm diameter and 40 cm length. A separate EPID guide was included in the audit instructions to assist with calibration and data acquisition. As centers either submitted one or two plans for their audit, a total of 27 IMRT plans and 19 VMAT plans were submitted resulting in 268 individual IMRT fields or VMAT arcs.

2.2. Planning and measurements

Each center planned the HN and PP trial patients on the provided patient datasets for IMRT or VMAT following the benchmarking instructions. A dose of 70 Gy was prescribed in 35 fractions for the HN plan and 64 Gy in 32 fractions for the PP plan. Except for one case at 10 MV these all were planned and delivered at 6 MV energy. The plans were then transferred onto the two supplied virtual phantoms within the local planning system. For 2D planar dose calculations the individual IMRT fields and VMAT arcs were transferred to the VFP at perpendicular incidence (zero gantry angle). This required collapsing all gantry angles to zero for the VMAT plans. For calculation of composite 3D dose the plans were transferred to the VCP at actual gantry angles. The phantoms were positioned at 90 cm source to surface distance (SSD). These verification plan doses were then exported in DICOM format. A DICOM-RT format TPS plan was also provided for calibration purposes.

All EPID measurements were made in-air with no phantom or treatment couch present. For the IMRT audit an integrated image for each field was acquired both at gantry zero and at actual gantry angles. For the VMAT audit EPID cine-images with 5 frames averaged per image were acquired continuously throughout the delivery. These were summed to obtain an integrated image for each arc. A calibration plan was also provided to determine EPID positioning and

sag with gantry angle as well as to calibrate EPID signal to dose. The centers exported their images and TPS doses and uploaded them via the cloud to the auditing site for assessment.

2.3. Analysis

All analysis was performed by the auditing site using in-house software developed in MATLAB (The Mathworks, Natick, USA). Integrated images of each individual IMRT field and VMAT arc delivery were used to reconstruct 2D dose planes at 10 cm depth in the VFP. Details of the method to calculate dose in phantom from EPID images have been detailed previously [7, 8]. For calculation of composite 3D dose in the VCP, a similar method to Ansbacher [12] was used with the IMRT images at actual gantry angles and the cine images for VMAT delivery. These were converted to dose in the VCP using the same dose conversion model as for the 2D individual field analysis but with additional contour correction and percentage depth dose modelling to derive 3D dose.

An in-house developed gamma (γ) algorithm was used for the dose comparison. All doses above 10% of the maximum dose were assessed with a search region of 0.6 cm radius. The gamma function used a global dose difference criteria defined as a percentage of the maximum dose. For 2D dose planes from individual fields or arcs, 2D gamma analysis was employed with the TPS dose map interpolated to the EPID resolution. Gamma pass-rate (GPR) and gamma mean values (GMV) were calculated for each 2D dose plane comparison for the individual IMRT fields and VMAT arcs (268 comparisons). The GPR is the percentage of assessed points that have a gamma score of less than or equal to 1. The GMV is the mean of the gamma scores of all assessed points in the 2D distribution. Similarly GPR and GMV were calculated for the composite 3D dose distributions using 3D gamma analysis with both dose distributions interpolated to 0.4 times the distance-to-agreement metric.

A multivariable linear model was made for quantitative assessment of the significance/contribution of different (explanatory) variables on the overall outcome of the audit. This is a standard statistical technique to examine the influence of different variables on an overall result. Explanatory variables that were chosen were Linac-TPS combination, TPS calculation grid resolution, IMRT or VMAT delivery, age of EPID, treatment site (HN or PP), record and verification (R&V) system type and nominal dose-rate. The EPID to dose conversion method was developed using measured doses in water and EPID images from Varian Clinac linear accelerator for aS1000 type EPID [8] and Truebeam linear accelerators for aS1200 type EPID [10, 11] at a center with Eclipse planning system. Therefore this will examine whether

the Varian and Truebeam combinations with Eclipse produce higher pass-rates than other combinations. The other variables were chosen based on available data from each center for the audit. The linear model was based on analysis of least squares of the GMVs for the 268 2D dose planes in the audit. The influence of the explanatory variables was studied through both visual and statistical assessment. The visual assessment was made by scatterplot of the audit GMVs versus each variable. The Tukey-Kramer honest significance test (HSD) and student's t-test were used for assessment of the significance of the differences in results due to the explanatory variables. Statistical studies were performed in JMP software [13].

Finally, to assess the consistency of the VESPA audit with other reported audits, the results were compared with published results. To this purpose, the incidence ratio (IR) was calculated as the ratio of the observed GPR for the VESPA audit to the reference audit. Comparisons should be 'stable' if the range for the 95% confidence interval is 'small', i.e. < 0.5 . The 95% confidence interval was calculated using:

$$IR \pm 1.96 \left(\frac{IR}{\sqrt{(\# \text{ of observed planes})}} \right)$$

III. RESULTS

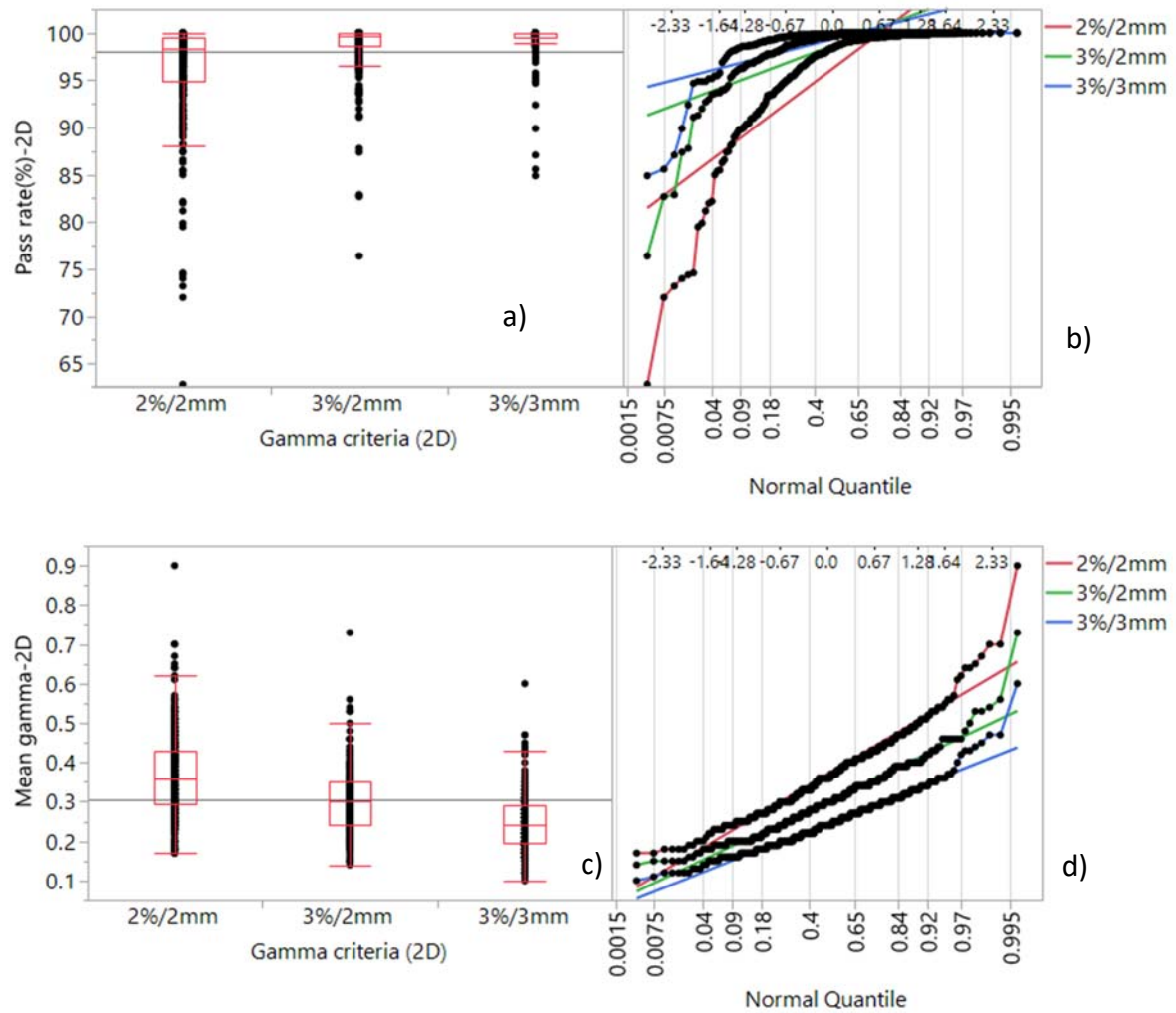


Figure 7-1- Gamma analysis results and normal quantile linearity for the 2D dose plane comparisons for 268 IMRT fields and VMAT arcs at 3%/3mm, 3%/2mm and 2%/2mm criteria. The normal quantile linearity indicates normality of distributions. (a) GPRs; (b) GPR normal quantile; (c) GMVs; (d) GMV normal quantile.

Figure 1(a) and (c) demonstrate the spread of GPRs and GMVs for different criteria for the measured planar IMRT fields and VMAT arcs. Maximum GPRs were 100.0% and minimum GPRs were 84.9%, 76.4% and 62.7% for 3%/3mm, 3%/2mm and 2%/2mm criteria respectively. The mean GPRs and GMVs are shown in Table 1. Normal quantiles are plotted for both GPRs and GMVs in Figure 1(b) and (d). As these figures suggest, more linearity is visually observed for GMV than GPR, indicating better normal distribution of GMV.

Table 7-1- Summary of the 2D audit gamma results.

Gamma criteria	GPR (1 SD)	GMV (1 SD)
2%,2mm	96.2 (5.5)%	0.37 (0.11)
3%,2mm	98.6 (2.7)%	0.30 (0.09)
3%,3mm	99.3 (1.9)%	0.25 (0.07)

The composite 3D dose distributions were analysed for the HN and PP plans in the VCP. Figure 2(a) illustrates the GPRs and Figure 2(b) the GMVs for the 3D gamma analysis. The maximum GPRs were 100.0%, 99.9% and 99.1% and the minimum GPRs were 80.6%, 56.6% and 26.4% for 3%/3mm, 3%/2mm and 2%/2mm criteria respectively. Mean GPRs (± 1 SD) were 97.7 (3.3)%, 92.5 (8.0)% and 80.8 (14.5)% for the same criteria.

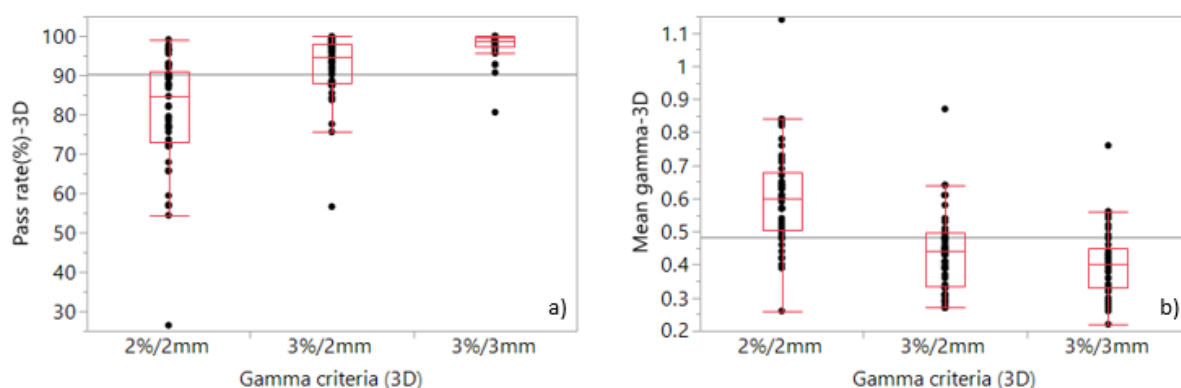


Figure 7-2- (a) GPRs and (b) GMVs for composite 3D dose analysis for the plans.

A multiple variable linear model was made using the GMVs for the 2D dose plane comparisons. Table 2 summarizes the model outcome for the explanatory variables. The most influential variables in determining the results were Linac-TPS combination, TPS grid resolution and delivery type (IMRT or VMAT). The least significant variables were EPID age, treatment site, record and verification system and dose-rate.

Table 7-2- Effect of the explanatory variables on overall audit results. The columns have been ordered according to the significance of each variable on the results.

Variable	LogWorth	p
Linac - TPS	12.824	0.00000
TPS grid resolution	4.782	0.00002
IMRT/VMAT Delivery	3.855	0.00014
EPID age-5ys	2.030	0.00933
Treatment site	0.976	0.10561
R&V	0.814	0.15353
Dose rate	0.011	0.97501

Figure 3 shows GPS and GMV scatterplots for the three most significant explanatory variables. The 1st plot for Linac-TPS combination shows some apparent distinctions between results for different combinations of linear accelerator and TPS type. The other two variables were colored according to the Linac-TPS combination.

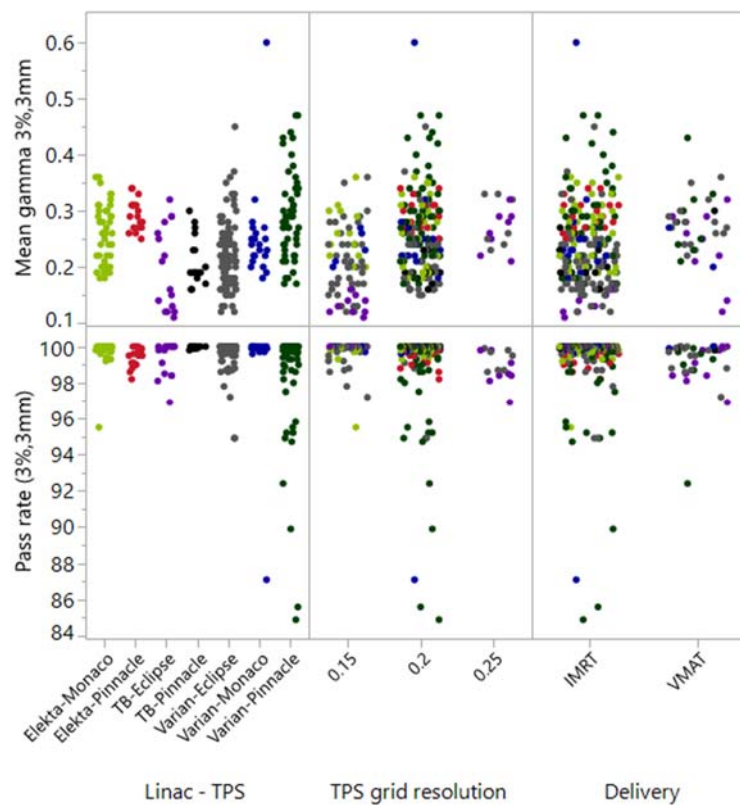


Figure 7-3- Scatterplot of the GMVs and the GPRs for the 2D dose plane comparisons of the audit versus the most significant explanatory variables (Linac-TPS combination, dose grid resolution and delivery type).

Figure 4 contains plots of GMVs (3%,3mm) estimated marginal means (EMM, named lsmeans in JMP) and 95% CIs for the three variables with significant effects in the model. Follow-up testing of the significant differences (Tukey-Kramer HSD/student's t test) between the means for the significant variables led to the following interpretations. For Linac-TPS, TB-Eclipse had a significantly lower mean than all other combinations (except TB-Pinnacle). The 4 combinations Elekta-Monaco, Elekta-Pinnacle, Varian-Monaco and Varian-Pinnacle were not significantly different to each other and appear to form a group with similarly high levels. There was some support for TB-Pinnacle and Varian-Eclipse having somewhat lower levels than the high group of 4 with 4 instances of significantly lower means (TB-Pinnacle lower than Varian-Monaco, Varian-Pinnacle and Elekta-Pinnacle, Varian-Eclipse lower than Varian-Pinnacle). For TPS grid, resolution 0.25 cm had higher GMV than the other two conditions which were both the same. For delivery, VMAT was higher than IMRT. Additional file 1 lists the test results.

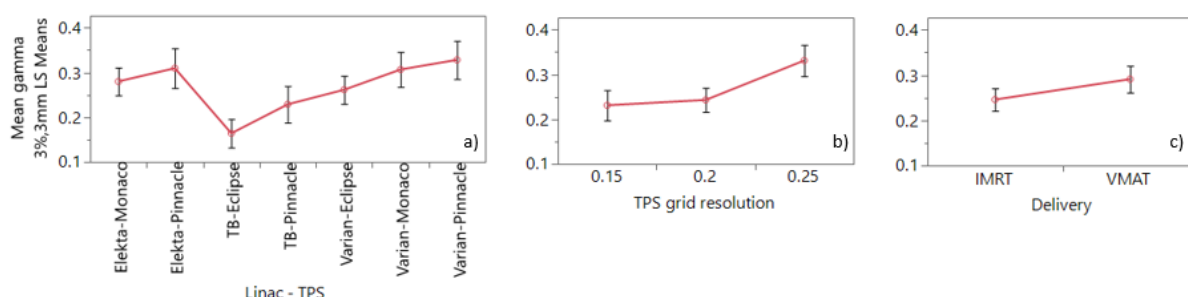


Figure 7-4- Plot of GMV for the three explanatory variables that showed most influence on the audit results (Linac-TPS combination, TPS dose grid resolution and IMRT/VMAT delivery)

Table 7-3- Comparison of the VESPA audit results with other recent audits. The GPRs are compared at 2%/2mm criteria.

Ref	Variable	Compare group	VESPA study γ	IR(95% CI)	Range	Significance/stability
		γ GPR %(no)	GPR %(no)			
1-[14]	Linac type		Median			
	Varian	96.7 (25)	96.8 (26)	1.0(0.8-1.2)	0.4	Insignificant/stable
	TB		96.2 (12)	-		
	TPS type		Median			
	Eclipse	97.3 (22)	96.3 (26)	1.0(0.8-1.2)	0.4	Insignificant/stable

	Monaco	98.8 (4)	98.5 (2)	1.0(0.9-1.1)	0.2	Insignificant/stable
	Pinnacle	88.7 (6)	96.1 (10)	1.1(1.0-1.2)	0.2	Significant/stable
	Delivery type	Mean				
	IMRT	90.0 (23)	96.3 (230)	1.1(0.9-1.3)	0.4	Significant/stable
	VMAT	93.0 (31)	95.5 (38)	1.0(0.8-1.2)	0.4	Insignificant/stable
	TPS type	Mean				
2-	Eclipse	95.0 (21)	98.0 (113)	1.0(0.8-1.2)	0.4	Insignificant/stable
[15]	Monaco	84.0 (5)	96.4 (68)	1.1(0.9-1.4)	0.5	Significant/unstable
	Pinnacle	91.7 (19)	93.7 (87)	1.0(0.8-1.2)	0.4	Insignificant/stable
	Treatment site	Mean				
	H&N	90.0 (25)	95.2 (135)	1.1(0.8-1.3)	0.5	Significant/unstable
	Pelvic	93.0 (10)	97.2 (133)	1.0(0.8-1.3)	0.5	Insignificant/unstable
5-	Delivery type	Mean				
[16]	IMRT	92.0 (155)	96.3 (230)	1.0(0.8-1.3)	0.5	Insignificant/unstable
6-	IMRT/VMAT	Mean				
[17]		90.0 (1265)	96.2 (268)	1.1(0.9-1.3)	0.4	Significant/stable
7-	VMAT	Mean				
[18]		88.0 (118)	95.5 (38)	1.1(0.9-1.3)	0.4	Significant/stable

Table 3 summarises the gamma comparisons at 2%/2mm between VESPA and five conventional audits. Comparisons were made as variable specific as possible based on the published data, resulting in 16 comparisons. For the comparisons, 5 out of 16 were unstable as their interval range was quite ‘wide’, ≥ 0.5 , and no conclusion was made for them. Among stable comparisons, 7 indicated similar pass rates of the VESPA with other audits (IR=1) and 4 comparisons demonstrated higher pass rates for VESPA than the other audits (IR>1).

IV. DISCUSSION

The 3D composite dose audit results showed lower GPRs and larger GMVs than the 2D individual field/arc dose plane audit results. The 3D analysis could not currently be performed with 3%,2mm criteria as recommended by TG218 report for pre-treatment QA methods while the 2D analysis would meet this criteria. However the 3D analysis is sensitive to gantry angle errors as the dose for each image is calculated with the acquired gantry angle and is therefore an important component of the audit. The EPID measurement is inherently 2D and to estimate a 3D dose distribution in the virtual cylindrical phantom requires modelling of percentage depth

dose. For the VESPA audit a single field percentage depth dose model was used which was also center independent. Improvement using a field-size specific and/or center-specific depth dose model could be explored. As a result the GMVs from the 2D individual field/arc dose comparisons were used for the statistical analysis in this paper.

Linac-TPS combination was found to influence the audit results. The Linac-TPS combination was used in the analysis rather than as separate variables due to the lack of spread of TPS type across all linac types which could bias results. The TB-Eclipse and TB-Pinnacle combinations were particularly found to result in lower GMV results. There are many potential reasons for this related both to this combination and the audit methodology. The Truebeam systems are a modern linac platform with high specifications for isocentre accuracy and other parameters. They have very accurate EPID positioning with active correction of EPID sag with gantry angle. The newer aS1200 imager does not have significant backscatter artifact which improves their performance for dosimetry. Plan complexity was not captured in the audit but could potentially have an effect. Future audits will include this parameter.

Centers were requested to produce the VFP and VCP plans at 0.2 cm or lower resolution although some submitted 0.25 cm resolution plans. The statistical analysis showed that the 0.25 cm resolution gave inferior results. The gamma algorithm used interpolates the TPS data to a high resolution to match the EPID resolution however this is clearly insufficient to counter the effect of the poor TPS resolution. Future audits will mandate the 0.2 cm or lower resolution based on these results. Another interesting finding was that the IMRT results showed lower GMV than the VMAT results. A possible explanation for this could be that the IMRT fields are acquired at fixed gantry angles and the data are corrected for EPID sag at these angles. However the VMAT acquires cine images during rotation and combines these into a single integrated image for 2D analysis. The individual cine images are not corrected for EPID sag and so the effect of this is likely to be greater and result in some blurring of the dose in the integrated image.

The VESPA audit is a TPS-planned audit not an end-to-end audit. These type of audits target a specific technology such as IMRT or VMAT and the CT scan of the phantom is typically provided to the center for planning. Comparing VESPA to other TPS-planned audits, the GPRs were similar to those from Clark et al. [14] for their audit of Varian VMAT deliveries conducted with the Octavius dosimetry system. While the VESPA results were higher for Elekta systems, the variability of these results meant that conclusions could not be drawn. For TPS systems the results were similar except for Pinnacle systems where the VESPA results had higher GPRs.

For IMRT audits as well as the Monaco TPS system, significantly higher GPRs were found for VESPA compared to the ArcCheck based audit of Eaton et al. [15].

For the VESPA audit as for most audits and in-house pre-treatment quality assurance the pass/fail criteria were arbitrarily set. It was not possible to know the uncertainties in a particular centers' TPS data or linac measurements. Pass/fail criteria could be set for future audits based on a statistical analysis of the current audit so that outlier centers could be identified. However the future audit would have to use a similar methodology and the same EPID to dose conversion model.

There are some limitations of this study. The measurement equipment are not completely standardized with differences between Varian EPID types (aS1000, aS1200) and Elekta imagers as well as equipment age. Data was collected on EPID response linearity as part of the study to ensure consistent results. 2D dose plane analysis is not ideal particularly for VMAT deliveries where a composite dose analysis would be preferred. By improving the 3D calculation model, it should be possible to audit centers using 3D dose distribution methods with more sensitive criteria (e.g. 3%, 2mm). Another possibility is to use dose-volume-histogram methods where the ratio of 2D doses is backprojected through the benchmark CT plan and hence percentage depth dose modelling is not required. Elekta linacs are not currently audited for VMAT using VESPA due to difficulties in obtaining cine images and gantry angle information. This is possible with Elekta's newer hardware and software (version 3.41) and a licence to access pixel scaling information, however these were not available at the time of the audit.

The EPID to dose conversion models were developed based on Varian Clinac and Truebeam measured beam data for single linacs and has been applied to multiple linacs of the same type. The model derived on Varian Clinac was applied to the Elekta linacs for this audit. Comparison of the field-size responses for the Elekta and Varian linacs using the TPS calibration plan data (2×2 to 25×25 cm² fields) showed that there was a small difference in the average field size factor for the two linac types of maximum 2.1% for the smallest field, and average 0.6%. The field size factors from dose derived from Elekta images of the above fields compared to the TPS data showed greater differences than for Varian/TB centers' data. The average of the absolute difference was 1.2% for Elekta and 0.6% for Varian/TB. Recently a model derived with Elekta measured data was compared to the Varian derived model for Elekta linacs in a separate study that has been submitted for publication. The improvement in results was small and not sufficient to affect the results of the current study.

Though not an end-to-end audit, the VESPA method provides a potentially inexpensive and rapid method to perform dosimetric auditing for specific assessments of new technologies. It takes about 2-4 hours to do the planning and delivery. Currently the analysis is based on in-house software. This software has several advantages over commercial systems in that it has a sophisticated backscatter correction; it accounts for EPID sag with gantry angle; and it allows 3D dose determination particularly for VMAT using cine-imaging. In principle the VESPA method can be applied in exactly the same way to flattening filter free (FFF) beams however there are currently hardware limitations for imaging these high-dose rate beams on the older EPID systems that are still prevalent. The newer Varian and Elekta EPID systems have FFF imaging capability. The EPID to dose conversion method has also not to date been developed or benchmarked for small field dosimetry auditing.

V. CONCLUSIONS

A new EPID-based remote dosimetric TPS-planned auditing method (VESPA) has been successfully applied to 30 audits of IMRT and VMAT for 21 centers across Australia and New Zealand. 2D dose-plane analysis was found to give more consistent results than 3D analysis. Statistical analysis of the results showed that there was some influence of Linac-TPS combinations on the results. This work shows that the remote EPID method can be used to audit centers with gamma pass-rates comparable or higher than other recent audits.

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Abbreviations

GPR: Gamma pass rate
GMV: Gamma mean value
EPID: electronic portal imaging device
VESPA: Virtual epid standard phantom audit
TPS: treatment planning system
IMRT: intensity modulated radiation therapy
VMAT: volumetric modulated arc therapy
VFP: virtual flat phantom
VCP: virtual cylindrical phantom
IR: incidence ratio

R&V: record and verify

Linac: linear accelerator

HN: head and neck

PP: post-prostatectomy

MLC: multi-leaf collimator

QA: quality assurance

CT: computed tomography

TROG: Trans-Tasman Radiation Oncology Group

TB: Truebeam

2D: two-dimensional

3D: three-dimensional

CI: confidence interval

SSD: source-surface distance

HSD: honest significance test

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Supplementary Files

Table 7-4- Participating centers in the VESPA audit and explanatory variables details for each center.

Center No	TPS-Linac	Grid Resolution (cm)	IMRT delivery type		EPID age (5ys)	R&V	Treatment site				Dose rate (MU/min)
							IMRT		VMAT		
1	Pinnacle-Varian	0.20	IMRT	-	1	Aria	HN	PP	-	-	400
2	Eclipse-Varian	0.15	IMRT	VMAT	1	Aria	HN	PP	HN	PP	600
3	Eclipse-Varian	0.15	IMRT	VMAT	2	Aria	HN	PP	HN	PP	400
4	Eclipse-Varian	0.20	IMRT	VMAT	2	Aria	HN	PP	HN	PP	400
5	Eclipse-Varian	0.25	IMRT	-	2	Mosaiq	-	PP	-	-	400
6	Eclipse-Varian	0.20	IMRT	-	2	Mosaiq	HN	PP	-	-	400
7	Pinnacle-TB	0.20	IMRT	VMAT	1	Mosaiq	-	PP	-	PP	600
8	Monaco-Varian	0.15	IMRT	VMAT	2	Aria	-	PP	HN	-	600
9	Pinnacle-Varian	0.20	IMRT	VMAT	2	Aria	-	PP	HN		400
10	Pinnacle-TB	0.20	IMRT	VMAT	1	Mosaiq	-	PP	-	PP	600
11	Pinnacle-Varian	0.20	IMRT	VMAT	1	Aria	HN	PP	HN	PP	600
12	Eclipse-TB	0.15	IMRT	VMAT	1	Mosaiq	-	PP	-	PP	600
13	Pinnacle-Elekta	0.20	IMRT	-	1	Mosaiq	HN	PP	-	-	400
14	Monaco-Elekta	0.20	IMRT	-	1	Mosaiq	HN	PP	-	-	600
15	Monaco-Elekta	0.15	IMRT	-	1	Mosaiq	HN	PP	-	-	600
16	Monaco-Elekta	0.20	IMRT	-	1	Mosaiq	HN	PP	-	-	600
17	Monaco-Varian	0.20	IMRT	VMAT	2	Mosaiq	HN	-	-	-	600

18	Eclipse-Varian	0.20	-	VMAT	1	Mosaik	-	-	HN	-	600
19	Eclipse-TB	0.25	-	VMAT	1	Aria	-	-	HN	PP	300
20	Eclipse-TB	0.25	-	VMAT	1	Aria	-	-	HN	PP	600
21	Eclipse-Varian	0.15	-	VMAT	1	Aria	-	-	-	PP	600

Table 7-5- Statistical testing of the differences between audit results (GMV) for the explanatory variables. Results with asterisk indicate significant differences where Variable 1 (V1) has lower GMV than Variable 2 (V2).

Linac-TPS			TPS grid resolution			Delivery type		
V1	V2	p	V1	V2	p	V1	V2	p
Elekta-Monaco	Elekta-Pinnacle	0.7933	0.15	0.20	0.5113	IMRT	VMAT	0.0001*
Elekta-Monaco	Varian-Pinnacle	0.1888	0.15	0.25	<.0001*			
Elekta-Monaco	Varian-Monaco	0.8233	0.20	0.25	<.0001*			
Elekta-Pinnacle	Varian-Pinnacle	0.9718						
TB-Eclipse	Elekta-Monaco	<.0001*						
TB-Eclipse	Elekta-Pinnacle	<.0001*						
TB-Eclipse	TB-Pinnacle	0.1057						
TB-Eclipse	Varian-Eclipse	<.0001*						
TB-Eclipse	Varian-Monaco	<.0001*						
TB-Eclipse	Varian-Pinnacle	<.0001*						
TB-Pinnacle	Elekta-Monaco	0.0632						
TB-Pinnacle	Elekta-Pinnacle	0.0175*						
TB-Pinnacle	Varian-Eclipse	0.7148						
TB-Pinnacle	Varian-Monaco	0.022*						
TB-Pinnacle	Varian-Pinnacle	0.0003*						
Varian-Eclipse	Elekta-Monaco	0.92						
Varian-Eclipse	Elekta-Pinnacle	0.1113						
Varian-Eclipse	Varian-Monaco	0.4193						
Varian-Eclipse	Varian-Pinnacle	<.0001*						
Varian-Monaco	Elekta-Pinnacle	1						
Varian-Monaco	Varian-Pinnacle	0.9845						

Chapter 8

Discussion

Safety and efficacy of a biological intervention should be assessed when an intervention is introduced to a radiotherapy treatment. The assessment known as clinical trials is performed through research in a controlled environment. The trials require minimum dependency to systematic and stochastic errors. Verification of dose delivery ensures low dependency of the trials to the errors and reduces the cost of the trials.

The complex nature of planning and delivery systems used in the trials using VMAT and IMRT may result in variations in dose deliveries among participants. Discrepancies between planning and delivery systems found in external audits are not always detectable by local pre-treatment QAs. Conventionally, an independent site performs the audit by site-visit(s) or by mailing tools. Though consistent and comprehensive, the site-visit method is expensive and time consuming. The mailing audit approach is also limited by the resources and costs involved in transporting equipment to and from each centre. Alternative introduced remote methods were also not able to analyse all the data centrally. This research presented and implemented a novel concept for remote dosimetric auditing of clinical trials using the 'TPS planned audit' model and EPID measurements. The approach used a model to convert in-air acquired IMRT/VMAT images to delivered dose inside the virtual phantom.

In Chapter 3, the dosimetry response of an aS1200 EPID system was studied and compared to the response from an aS1000 EPID system. The chapter outlined major dosimetry tests performed to commission the new EPID system on the Varian Truebeam accelerator. The linearity of the EPID dose response was investigated within 0.4% above 5 MU, and ~ 1% above 2 MU. The lag was measured as approximately 3% for small MU settings, which was considerably smaller than for the aS1000 imager. The symmetry of the profiles for the EPID was considerably improved over the aS1000 imager, indicating the effectiveness of the backscatter shielding in the new system. The study also developed a kernel based model using EPID images to determine delivered dose to a VFP. The model inputs were images acquired with the aS1200 high resolution EPID, and the model outputs were dose onto the phantom. For the modelling, the TPS fluence and measured dose of square fields inside a rectangular water phantom were utilised. A rectangular phantom was selected due to its simple structure, while being easily integrated with the image acquisition software. The fluence and dose profiles were modelled using square-field images so they could closely follow corresponding measured profiles. To validate the model performance for clinical application, the doses of 9 IMRT fields were modelled using their pre-treatment images from the EPID, and they were compared with the corresponding TPS measured doses. The method successfully back-projected the 2D dose inside the virtual phantom with a mean gamma pass rate of more than 99% for all beam energies

and modes, though an improvement was required for the FFF modes. Performance of the Truebeam model developed for the FFF modes was less than the one developed for flattened beams. The reduced performance was due to the more complex structure and sharp dose gradient of the FFF beam profiles and their different field size response. The current kernel-based model is unable to accurately capture dose at penumbra regions and a new algorithm should be developed.

The signal to dose conversion model is developed using EPID images from square field deliveries and it is trained with beam profiles and field size factors (FSFs) of a series of square field deliveries measured in water tank. For the audit, the vendor independency of the method (i.e. insignificant differences between vendors performance) is desired. Chapter 4 investigated the need for vendor specific conversion models for the audit. Comparing the EPID measured profiles and FSF data for a Varian (vendor 1) and Elekta (vendor 2) system showed some relevant dosimetry differences between the two vendors. Large discrepancies were observed at very large and small fields, $\sim 3 \times 3$ & 20×20 cm², for the two measurements. Small penumbras for the vendor 1 indicated sharper profiles of the images, though this could be due to the proximity of the collimating system to the machine isocenter of the vendor. Different FSFs were observed for the two vendors, which could be due to either EPID scatter or head scatter as the EPID signal incorporates both effects. Observed dosimetry differences between the two vendors suggested that developing an individualised model for each vendor would be beneficial. Accordingly, a ‘signal to dose’ model was developed for vendor 2, EM, and its performance was compared with those from the model for vendor 1, VM. The EM agreement with water tank data of vendor 2 was better than the VM’s agreement of vendor 1, though the agreement reduced at mid-profiles and at the edges of small fields. The performance of the existing VM and the new EM were studied for a series of audit IMRT fields measured on vendor 2 systems. The pass rates for dose verification and auditing of the deliveries were relatively high using the EM model for three gamma criteria, and their corresponding mean gammas showed similar behaviour. No significant difference was observed between the auditing results for the two treatment sites, head and neck (HN) and post-prostatectomy (PP). The mean gammas for all vendor 2 deliveries and treatment sites were better for the EM than the VM, and only one centre demonstrated relatively similar response for both the EM and VM. It would have provided an insight, if the EM performance was assessed on the Varian data. Although using a vendor specific model reduced mean gammas, it did not demonstrate a major improvement compared with the VM, and the differences were mainly observed in the most stringent gamma criteria at 1%/1 mm. Using vendor specific models reduced the audit

differences for vendors 1 and 2, though the mean gammas for vendor 2 were still larger than for vendor 1. This could be due to the impact of other variables such as the centre TPS type, which was not considered in this study, or due to the relative dosimetry inconsistencies observed among vendor 2. Differences of the EPID sags could be a reason too. In this study, an individualised sag was calculated for each EPID and the sag parameters were determined from calibration images of corresponding EPID. However, algorithm of the sag model was considered the same for all EPIDs while it was benchmarked against aS1000 EPID.

In Chapter 5, the in-house IMRT QA method was proposed to be extended to multi-vendor facilities for remote auditing of centres, and the challenges/limitations were studied. Details of the instructions for delivery, calibration and dosimetry were presented, and the analysis details were explained. This chapter summarised the advantages of VESPA as (1) fast turnaround mainly driven by the facility's capability to provide the requested EPID images, (2) the possibility for facilities to perform the audit in parallel, as there is no need to wait for a phantom, (3) simple and efficient credentialing for international facilities, (4) a large set of data points, and (5) a reduced impact on resources and environment, as there is no need to transport heavy phantoms or audit staff. The potential limitations of the VESPA for auditing purposes were also presented as (1) it does not provide absolute dosimetry, therefore a Level 1 audit is still required, and (2) it relies on correctly delivered open calibration fields, which are used for system calibration. The chapter generated the audit procedures for the remote centres and instructions on how to create the necessary calibration images in addition to the images acquired from the tested treatment plans. A specific EPID guide was also developed with instructions for each linac and EPID type for EPID positioning, flood and dark field calibration, cine mode setup, and image acquisition. To facilitate (and standardise) this process, DICOM plans for the calibration images were provided for the facilities to download and import into the patient management/record and verify system for delivery. The specific centres linac name was inserted into the DICOM plan files to facilitate import. Two main vendor specific challenges were identified for the linear accelerators and for the record and verify systems: 1- Transfer of images to Mosaiq results in loss of pixel scaling information to obtain integrated dosimetric images. The solution was identified as saving the images in Varian format in the cache on the linac used. For Varian Truebeam, using an Image Processing Service was suggested to store cumulative image frames. For Elekta linacs, images were exported from the iView EPID acquisition software in '.his' format with the 'log' file, with the 'log' file containing pixel scaling information. DICOM images were then created at the central site for analysis. 2- The Truebeam and Elekta cine imaging modes did not store dosimetric information. For Truebeam,

it was suggested to use an Image Processing Service to store cumulative image frames from which cine images could be derived, and for Elekta, Perkin Elmer XI service software was used to store individual frames. For Varian Clinac, using large MU (300) was advised for calibration of the EPID signal to dose to not miss frames at the start and end of the acquisition. As Elekta cine imaging mode using Perkin Elmer software does not store gantry angle for deliveries, using a separate inclinometer or independent video recording were suggested to obtain the gantry angle for delivery frames. Mans et al. also suggest that a field match of aggregated EPID data with the nearby field shape of the partial VMAT arc may render that obsolete [134].

Acquired results from chapters 3, 4 and 5 provided some potential action levels for the VESPA audit and extension of the IMRT QA to multicentre auditing. The developed models could be used for Varian Clinac/Truebeam and Elekta deliveries. The deliveries should include a flattening filter on the beam path. For Elekta deliveries, both EM and VM could be used at gamma criteria of looser than 3%/3mm but using the EM is recommended at 1%/1mm. The auditing method for Elekta deliveries seems irresponsive at the very small/large field size deliveries, though the patient sizes of the VESPA lies within the limit. Moreover, to avoid losing the image signals, it is suggested to save the images in Varian format when they are transferred to Mosaicq, to use Image Processing Service (IPS) for Truebeam deliveries and to use '.his' format for Elekta image recording. For Elekta VMAT deliveries, cine images should be acquired using Perkin Elmer XI Service to store frames individually. However, an independent video recording or inclinometer measurement is suggested to record gantry angles of each frame.

Chapter 6 assessed the methodology and feasibility of the VESPA concept for auditing of IMRT deliveries at six pilot centres. The 2D field-by-field analysis resulted in mean gamma pass rates over 99.5% at 3%/3 mm criteria and over 95.5% at 2%/2 mm criteria. For 3D dose analysis of the centres, only 1 out of 12 plans had a gamma pass rate below 95% at 3%/3 mm criteria. The pass rates were lower for the 3D analysis than the 2D analysis, reflecting the larger uncertainty in the 3D model where depth-dose modelling was required. The employed algorithm did not use vendor or centre-specific beam information, and a detailed investigation was suggested into the contributing uncertainty components of the VESPA model when implemented across multiple types of linacs. For the pilot study, three of the participants acquired data from Varian delivery and measurement systems (vendor 1), and three from Elekta (vendor 2). A slightly lower gamma pass rate and different field size response at the phantom isocentre was observed for vendor 2 compared with vendor 1, though preliminary differences between the two vendors had been applied to the audit analysis software. Using square field images, a separate analysis

of the dose conversion model performance was presented for each centre, and a comparison of the field-size responses was performed for the two vendors using the TPS calibration plan data (2×2 to 25×25 cm² fields). A small difference of maximum 2.1% for the smallest field, and average 0.6% in the average FSF was observed for the two linac types. The FSFs from dose derived from vendor 2 images of the fields compared to the TPS data showed greater differences than those from vendor 1 centres. The average of the absolute difference was 1.2% for vendor 2 and 0.6% for vendor 1. This suggested that improvement to the results could potentially be made by using vendor specific models. This was addressed later in a separate study as explained in above paragraphs on results from Chapter 4. A large discrepancy was also observed at large field sizes for two vendor 2 centres. The reason for this could be an EPID signal artefact introduced by scatter close to the peripheral electronics. The imager response from one centre was re-measured and it confirmed that the artefact exists for fields larger than 23×23 cm². This did not influence the gamma results in this study, as smaller field sizes were used for both of the studied plans. Future studies will restrict measurements for systems from Vendor 2 to a maximum field size of 23×23 cm². Furthermore, the EPID response versus MU demonstrated non-linearity at low MUs for the EPIDs. This could be due to the failure of the acquisition system in integrating all EPID frames, however the magnitude varied for the centres. Further investigation and data are required in order to determine the causes of these variations. This study analysed the results at 3%/3 mm, 3%/2 mm and 2%/2 mm criteria for both 2D and 3D dose distributions, though the sensitivity of the method could be assessed and compared with a DVH approach.

In Chapter 7, the audit was applied to multiple centres across Australia and New Zealand, and comprehensive results were presented for 268 fields/arcs from 21 centres. The 3D composite dose audit results showed lower gamma pass rates and larger gamma mean values than the 2D individual field/arc dose plane audit results. The 3D analysis could not currently be performed with 3%/2 mm criteria as recommended by the TG218 report for pre-treatment QA methods, while the 2D analysis would meet this criteria. However, the 3D analysis is sensitive to gantry angle errors as the dose for each image is calculated with the acquired gantry angle and is therefore an important component of the audit. The EPID measurement is inherently 2D and to estimate a 3D dose distribution in the VCP, it requires modelling of the percentage depth dose. For the VESPA audit, a single field percentage depth dose model was used, which was also center independent. Improvement using a field-size specific and/or center-specific depth dose model could be explored. The gamma mean values from the 2D individual field/arc dose comparisons were used for the statistical analysis, as they showed better linearity. The most

determinative variables on the gamma mean values were Linac-TPS combination, TPS grid resolution, and IMRT/VMAT delivery type. The Linac-TPS combination was used in the analysis rather than as separate variable due to the lack of spread of TPS types across all linac types, which could bias results. The Truebeam-Eclipse and Truebeam-Pinnacle combinations in particular were found to result in lower gamma mean value results. There are many potential reasons for this, related both to this combination and to the audit methodology. The Truebeam systems are a modern linac platform with high specifications for isocentre accuracy and other parameters. They have very accurate EPID positioning, with active correction of EPID sag with gantry angle. The newer aS1200 imager does not have significant backscatter artefacts which improves their performance for dosimetry. Plan complexity was not captured in the audit but could potentially have an effect. Furthermore, the centers were requested to produce the VFP and VCP plans at 0.2 cm or lower resolution, although some submitted 0.25 cm resolution plans. The statistical analysis showed that the 0.2 cm resolution gave superior results. The gamma algorithm used interpolated the TPS data to a high resolution to match the EPID resolution, however this was clearly insufficient to counter the effect of the poor TPS resolution. Future audits will mandate the 0.2 cm or lower resolution based on these results. Another interesting finding was that the IMRT results showed lower gamma mean values than the VMAT results. A possible explanation for this could be that the IMRT fields were acquired at fixed gantry angles and the data was corrected for EPID sag at these angles, but the VMAT acquired cine images during rotation and combined them into a single integrated image for 2D analysis. The individual cine images were not corrected for EPID sag, so the effect was likely to be greater and resulted in inaccuracies for dosimetry of the integrated image. The remote method demonstrated high gamma pass rates and provided results comparable to more resource-intensive audit methods. Comparing VESPA to other TPS-planned audits, the gamma pass rates were similar to those from Clark et al. [116] for their audit of Varian VMAT deliveries conducted with the Octavius dosimetry system. While the VESPA results were higher for Elekta systems, the variability of these results meant that conclusions could not be drawn. For TPS systems the results were similar, except for Pinnacle systems where the VESPA results had higher gamma pass rates. For IMRT audits, as well as the Monaco TPS system, significantly higher gamma pass rates were found for VESPA compared to the ArcCheck based audit of Eaton et al. [199].

Though the pilot audit showed lower gamma pass rates for Elekta than Varian deliveries, the overall audit outcome demonstrated similar performance for the two linacs when using similar TPS. In this thesis, the Chapter 6 was performed earlier than Chapter 4 and it suggested more

tightened tolerances to compare sensitivity of the VM and EM for Elekta deliveries. The overall pass rates for the VESPA were very close to other ‘TPS-planned’ audits which indicate a close sensitivity of the two groups. There is no need to use different tolerances for audits of departments with linacs from different vendors since the corresponding models are tolerance independent. However, the smaller tolerances, the lower pass rates. Though no strict action level was considered for this thesis, following TG218, if pass rates were less than 95% at 3%/2mm for 10% dose threshold, an investigation was performed. However, a parallel study is in progress on correlation of the audit tolerances with DVH response to make the evaluation clinically more relevance.

Chapter 9

Conclusions

In support of facility auditing for the trials using IMRT and VMAT, this research presented a novel concept for remote dosimetric auditing of clinical trials. The approach was termed the Virtual Epid Standard Phantom Audit (VESPA) which used EPID to dose conversion models for assessment.

The VESPA provided an inexpensive and rapid method to perform dosimetric auditing for specific assessments of new technologies. The analysis used an in-house software with several advantages over commercial systems; it embedded a sophisticated backscatter correction; it accounted for EPID sag with gantry angle; and it allowed 3D dose determination, particularly for VMAT using cine-imaging. It took about 2-4 hours to do the planning, delivery and assessment. The method accuracy was comparable with more resource intensive audits.

Chapter 10

Future Work

The VESPA can be applied to FFF beams in a similar procedure to the FF beams. However, most EPIDs do not have the imaging ability for the FFF beam. Moreover, the current model performance was reduced for the FFF beams, and inaccuracies were observed for their beam profiles. This is due to the complex structure of the beam profiles with the field size and the sharp dose gradients of the fields, which will require improvement of the model. Perhaps a different kernel model better estimates the FFF beam distribution. The kernel could be modelled by Monte Carlo or any similar modelling method.

The VESPA is also inappropriate for dosimetry auditing fields smaller than $3 \times 3 \text{ cm}^2$, since the developed models were optimised using dosimetric data of larger size fields. The beam profiles and FSFs at small field sizes do not follow similar trends of the larger field sizes, and it is better explained by a non-linear behaviour. Absolute dosimetry is also challenging in small fields due to technical deficiency. Developing a pre-treatment model in the future that estimates dose at small fields is useful for dose verification of small field treatments, e.g. stereotactic radiosurgery audits.

Another limitation of the current implementation for the VESPA is that it does not provide absolute dosimetry and it relies on relative dose from ratio of calibration images to TPS dose at isocentre. Therefore, a Level 1 audit is still required. In future, the dose to the provided CT dataset of the patients could be determined using the images.

For the 3D dose volume analysis in the VCP, the results were lower than for 2D analysis in the VFP. It is likely due to uncertainties in the modelling of percentage depth dose, as a single depth-dose model was used for these analyses. Improvement using a field-size specific and/or centre-specific depth dose model could be explored. Another approach that may improve the results is the use of a larger diameter virtual phantom to reduce high dose regions near the phantom surface. An alternative for a 3D model is also using DVH methods where the ratio of 2D doses is backprojected through the benchmark CT plan, and hence percentage depth dose modelling is not required. A sensitivity analysis of the method should be performed to ensure that clinically significant dosimetric errors can be detected. Sensitivity of the gamma assessment can be compared with a DVH approach. Therefore in future, the VESPA could be used to determine the dose to the provided virtual patient CT dataset from the EPID images performing a DVH analysis.

A lower performance was observed for VMAT than IMRT. This could be due to a greater sag for the VMAT cine images, as the images were combined into single integrated images that were not corrected for EPID sag, resulting in some dose blurring in the integrated image. This could be addressed in the future. Furthermore, the Elekta linacs were not audited for VMAT

using the VESPA method, due to difficulties in obtaining cine images and gantry angle information. This would be possible with the Elekta's recent hardware and software, v 3.41, and a licence to access pixel scaling information. Correcting sag for cine images of VMAT and VMAT delivery assessment for Elekta systems could be considered in future VMAT auditing.

Similar to most audits and pre-treatment QAs, the pass/fail criteria were arbitrarily set. It was not possible to know the uncertainties in a particular centres' TPS data or linac measurements. The criteria and action level could be set for future audits based on the statistical analysis of the audit so that outlier centres could be identified. However, this setting requires that the future audit would have to use a similar methodology and the same EPID to dose conversion model.

Abbreviations

TPS	Treatment planning system
QA	Quality assurance
EBRT	External beam radiotherapy
MLC	Multi-leaf collimator
3DCRT	Three dimensional conformal radiotherapy
IMRT	Intensity modulated radiotherapy
IMAT	Intensity modulated arc therapy
VMAT	Volumetric modulated arc therapy
SBRT	Stereotactic body radiotherapy
DVH	Dose volume histogram
MRI	Magnetic resonance imaging
OAR	Organ at risk
PTV	Planning target volume
CTV	Clinical target volume
TCP	Tumour control probability
NTCP	Normal tissue complication probability
SM	Setup margin
IM	Internal margin
SD	Standard deviation
TLD	Thermoluminescent dosimeter
OSLD	Optical stimulated luminescence dosimeter

MOSFET	Metal oxide semiconductor field effect transistor
EPID	Electronic portal imaging device
SDD	Source to detector distance
DICOM	Digital imaging and communications in medicine
FF	Flattening filter
FFF	Flattening filter free
MV	Mega electron volt
KV	Kilo electron volt
HDR	High dose rate
LDR	Low dose rate
PDR	Pulsed dose rate
VESPA	Virtual epid standard phantom audit
TB	TB

Appendix

VESPA instruction



Trans-Tasman Radiation Oncology Group

Virtual EPID Standard Phantom Audit (VESPA)

IMRT/VMAT Dosimetry Assessment

Version 4.11

26 April 2017

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Communication

Initially contact Peter Greer with any issues. Feel free to text, phone or email.

Wiki

www.newmedphys.org

Information will be added to this wiki as the project progresses and results become available. login vespa, password vespa123

Acronyms

<u>Acronym</u>	<u>Term</u>
EPID	Electronic portal imaging device
TPS	Treatment planning system
VFP	Virtual Flat Phantom
VCP	Virtual Cylinder Phantom
MU	Monitor Units

Table 2 Acronyms

Background

Historically dosimetric auditing of treatment centres participating in clinical trials has been performed by site visits. For these visits a unified measurement and analysis approach is possible to ensure that they are delivering high quality radiation therapy with accurately reported dose distributions. These can be Level 3 audits where an anthropomorphic phantom is planned and treated as a patient with dosimeters inserted to determine dose, usually point dose and a 2D plane or planes measured with film. However this approach is expensive and logistically difficult to perform regularly which is particularly a problem with the rapid introduction of new advanced and complex delivery methods (1). The IROC in Houston has sent centres a head and neck anthropomorphic phantom and dosimeters which they plan and treat and then send back the phantom with dosimeters (2, 3). The tolerance limits have been large at 7% dose and 4 mm distance to agreement and the success rate of this method has not been very high showing that producing accurate dosimetry in this way is challenging.

More recently some alternative methods using site visits have been utilised that use digital IMRT/VMAT type dosimetry systems rather than anthropomorphic phantoms and film. A site visit based audit used a head and neck IMRT plan transferred to a solid water block to perform film measurements (4). They determined that 75% of the films passed a Gamma criteria of 4%,3mm. (5, 6) used the Octavius II phantom with PTW array to measure 2D dose planes in the Octavius II phantom. Both point dose and Gamma pass rates were analysed with multiple Gamma criteria 2%,2mm, 3%,2mm and 3%,3mm. Haworth et al. (private communication)] performed a similar type of audit as a pilot but with the Octavius 4D system which rotates the detector with the gantry and calculates a 3D dose distribution from the 2D midplane dose measurements using golden percentage depth dose data. They analysed both 2D dose planes and 3D whole volume Gamma results from the centres also at the above criteria.

An alternative lower-cost approach has been proposed recently by the European Organisation for the Research and Treatment of Cancer (EORTC) in conjunction with IROC which they termed the use of “institutional virtual phantoms” (7). This method uses the actual institutions phantoms and measurement devices. The centre send the CT data set of their phantom and the planned dose distribution in the phantom along with their measured data to a central site for analysis. Although this is a low cost solution there are still many problems including the wide variety of techniques and dosimeters used making scientific evaluation of the results difficult. 1/3rd of centres could not be analysed centrally with this method. The validity of in-house QA methods has been questioned by Kry et al. who found that the in-house results did not predict for poor results using a standard phantom audit.

VESPA (Virtual Epid Standard Phantom Audit) is a novel method that aims to remotely credential centres using measurements made at each centre of their delivery using their electronic portal imaging device (EPID). As all current vendors’ EPIDs are mostly consistent, and the different vendors EPIDs are very similar in structure, the advantage of this approach is greater consistency of measurement methods. As the data is centrally analysed the analysis methods are standardised. This method combines the cost advantages of the virtual phantom method with the more rigorous methods used in site visits with consistent measurement and analysis. It is most similar to the Octavius phantom studies mentioned earlier in that a 3D dose

distribution in a cylindrical water-equivalent phantom is estimated from the EPID measurements and compared to the planning system calculation. A comparison of the scope of the VESPA audit compared to the recent TROG VMAT pilot audit and generic Level 3 audits is presented below.

Dosimetric Test	Level 3	TROG VMAT Pilot (Octavius)	VESPA	Comment
CT scanner produces accurate data for the phantom to load into TPS	possible	X	X	Some "Level 3" provide the CT data. This could potentially be assessed with other tests.
Dose calculation inhomogeneity corrections	possible	X	X	The VESPA verification plan is in a water-equivalent phantom. Where inhomogeneity corrections are important and known to be problematic (e.g. lung) then Level 3 with inhomogeneity phantom should be considered.
IMRT and VMAT fluence modelling and dose calculation	✓	✓	✓	
Couch modelling	possible	✓	X	Depends if couch included in Level 3 audit
Phantom setup at linac	✓	✓	X	Has limited role in dosimetric testing, especially if the phantom is not anthropomorphic
IMRT/VMAT delivery accuracy e.g. MLC leaf motion, dose-rate	✓	✓	✓	VESPA likely more comprehensive due to larger number of assessed points and potentially higher resolution
3D dose distribution assessed	possible	✓	✓	3D dose assessment with anthropomorphic phantoms is generally limited to a small number of discrete measurement points or a film without expansion to the entire volume, as done with QA type phantoms eg Octavius, ArcCheck
Effect of target motion on the delivery e.g. interplay effect	possible	X	X	Requires moving phantom with dosimeter inserted.

Table 3 Scope of the VESPA audit in comparison to Level 3 audits and the TROG VMAT pilot audit.

Comparison Metrics, Action Levels and Uncertainties

It should be noted that there is no Level 1 absolute dose assessment associated with VESPA therefore the dose results are relative to the users stated beam output. Errors in this output if present are not included in the VESPA results.

The Gamma criteria (8) and pass-rate acceptance criteria that have been used generally depend on what is achievable with any particular dosimetric equipment and audit procedure. The use of digital IMRT/VMAT type dosimetry systems has allowed for higher pass rates with tighter Gamma criteria compared with Level 3 anthropomorphic type phantoms and film

measurements. The TG119 report on the implementation of IMRT recommended the use of 3%,3mm with a pass-rate of greater than 90% for field-by-field analysis and 88% for combined field dose analysis (9). This recommendation is for in-house commissioning of IMRT. A recent survey of IMRT QA in Australia and New Zealand (J.Barber, private communication) found that the most common criteria for in-house IMRT QA measurements was 3%,3mm with acceptance pass-rate of 95%, however a considerable number of centres were using 90% pass-rate. The Australian Clinical Dosimetry Service when assessing Level 3 dose does not regard the uncertainties in their measurements as being quantifiable as these depend on the treatment planning system uncertainties which are not well described. They therefore set an Tolerance Level of 5% as being clinically relevant and set Action Levels from these to trigger investigations.

Audit Recommended Outcomes

VESPA using remote EPID based assessment for a wide variety of user planning and treatment equipment is a new technique and the uncertainties are similarly not easily quantified. We do not in this report state a within tolerance or out-of-tolerance as this is a judgement that a particular clinical trial should determine based on their requirements. However we make initial recommendations as given in the Table below that with 3%,3mm criteria that an average pass rate greater than 95% should be achieved for field-by-field analysis and given the greater uncertainties of the 3D dose reconstruction that with a 3%,3mm criteria pass rate of greater than 90% should be achieved. However we would not currently recommend that a centre is out-of-tolerance unless the average field-by-field at 3%,3mm and the 3D dose at 5%,3mm criteria were below 90%. These recommendations will change over time with increasing information, application and improvement of the method. There is also currently development and use of dose-volume-histogram type analysis of delivered dose distributions however there is very limited experience with this for auditing studies. Ongoing development of the VESPA system will include this capability.

Range	Audit Outcomes		Actions
	2D Gamma (Field-by-field)	3D Gamma (Combined dose)	
Acceptable	3%,3mm criteria (10% threshold). Average pass rate greater than 95%	3%,3mm criteria (10% threshold). Pass rate greater than 90%	Final report to Facility. Recommend to TROG that facility passes audit.
	3%,3mm (10% threshold). Average pass rate greater than 90%	5%/3mm (10% threshold) Pass rate greater than 90%	Repeat procedure. Follow up with Facility.

Table 4 Initial VESPA Audit Recommendations

Methods

VESPA Process

An overview of the VESPA process is shown in Figure 1. The local Centre produces a RAVES and HPV trial plan and copies the trial plans onto the Virtual Cylinder Phantom for a combined 3D dose distribution and the Virtual Flat Phantom for individual field or arc analysis. These doses are exported to the Central Site. EPID measurements are made of each field in the case of IMRT and with cine-mode imaging during delivery for VMAT. EPID images of calibration plans and EPID Dosimetry test plans are also acquired. All images are uploaded to the central analysis site. Combined with calibration images the IMRT/VMAT images are used to reconstruct the delivered dose in the virtual phantoms for comparison at the central analysis site.

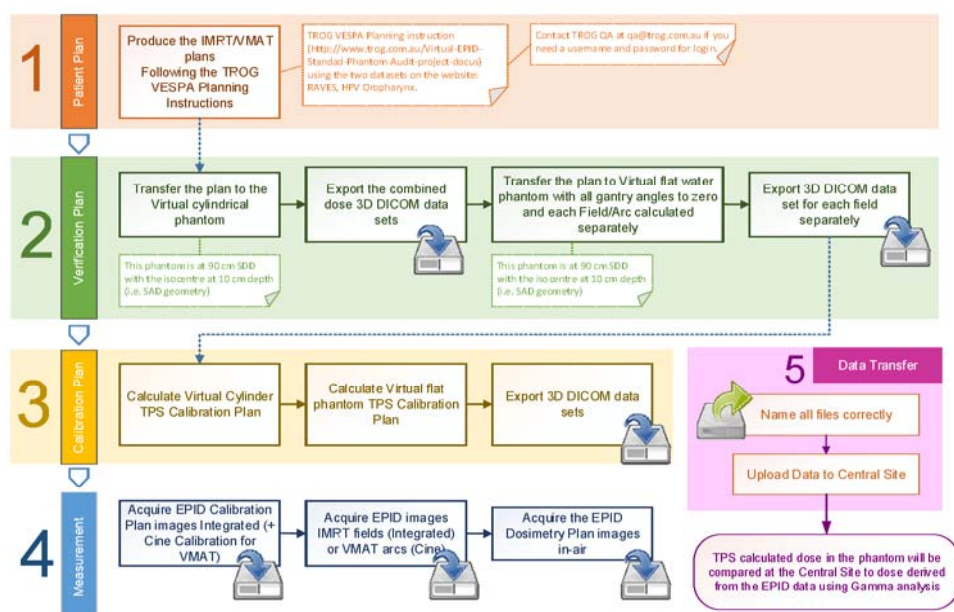


Figure 1 Overview chart of VESPA process

Trial and Verification Plans

The local Centre produces two plans with patient CT data sets provided by TROG. These are currently the RAVES and HPV benchmarking exercise plans. These plans are then transferred to the virtual phantoms and the 3D dose distributions exported in DICOM format. Simple calibration plans are also produced on the two phantoms. These calibration plans are required to determine reference doses for the EPID to dose conversion, and to verify the accuracy of the EPID to dose conversion model for the local users EPID and treatment planning system combination. DICOM Plan files are provided to each Centre to automatically set up these calibration plans.

EPID Measurements

Two EPID delivery plans are required in addition to the EPID images for the trial plans. The first is an EPID calibration delivery plan. This is required to determine pixel value to dose scaling for dose conversion, determine the central axis pixel location on the EPID and to determine the EPID sag with gantry angle so that a correction function can be derived. The second is an EPID dosimetry delivery plan. This plan is designed to ensure the dosimetric integrity of the local EPID, for example linearity with MU, and in combination with the TPS calibration plans verify the accuracy of the EPID to dose conversion model for the local users EPID and treatment planning system combination.

The EPID images acquired for IMRT and VMAT differ in that for IMRT fields a single integrated image is obtained for each field at the planned gantry angle, whereas for VMAT cine-images are required during the arc delivery. This enables a 3D dose to be determined for VMAT delivery and verifies the accuracy of delivery with gantry angle.

EPID to Dose Conversion

The method converts EPID images measured in-air to a dose plane in a virtual flat water phantom at 10 cm depth using previously reported methods (10, 11). These dose planes can be used to verify individual IMRT field doses.

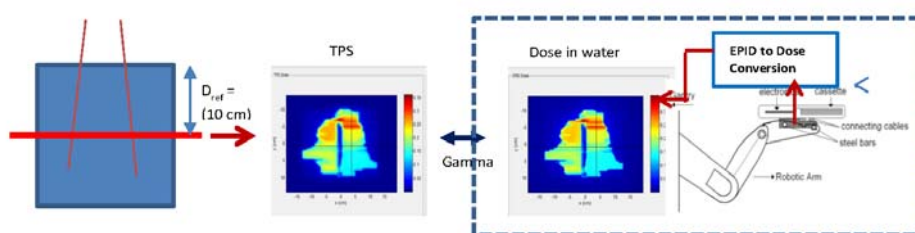


Figure 2 Illustration of VESPA process for a virtual flat phantom used for 2D field-by-field dosimetry

For combined field analysis (2D planes and 3D dose) this dose plane is then modified to estimate the dose plane at the same depth but in a virtual cylindrical phantom. This is then converted to a three-dimensional (3D) dose distribution in the virtual phantom using percentage depth dose modelling (12). This conversion is performed for each image acquired and the dose is projected along the gantry angle for the image. For VMAT this means the dose is calculated for each acquired cine image and the gantry angle for the image must be known. The virtual cylindrical phantom currently used has a 20 cm diameter and is 40 cm long.

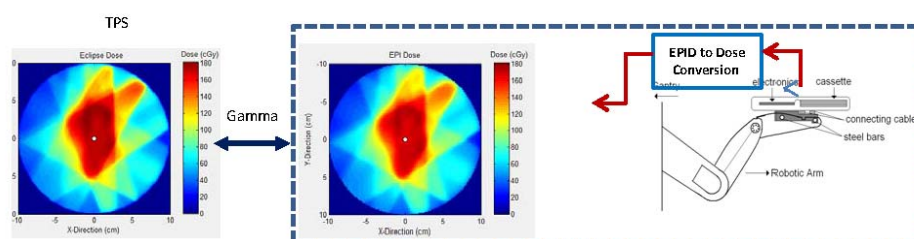


Figure 3 Illustration of VESPA process for a virtual cylinder phantom used for 3D combined field dose analysis

These doses are then compared to the TPS calculations. Software has been developed at Calvary Mater Newcastle (CMN) to perform this analysis. The conversion to dose and backscatter correction (for Varian EPID) models have been developed using EPID and water-tank measured data at CMNH on both C-Series and Truebeam linacs and the Central Coast Cancer Centre. The software is used at both these treatment Centres for pre-treatment IMRT verification. Extensive validations of the dose conversion at depths in flat water phantom have been performed with comparison to MapCheck 2D planar dose distributions in solid water phantoms. Limited verification measurements have been made to date of 3D dose distributions in water equivalent cylinders. Verification for this geometry have been with comparison to the Eclipse planning system calculations.

Analysis of data

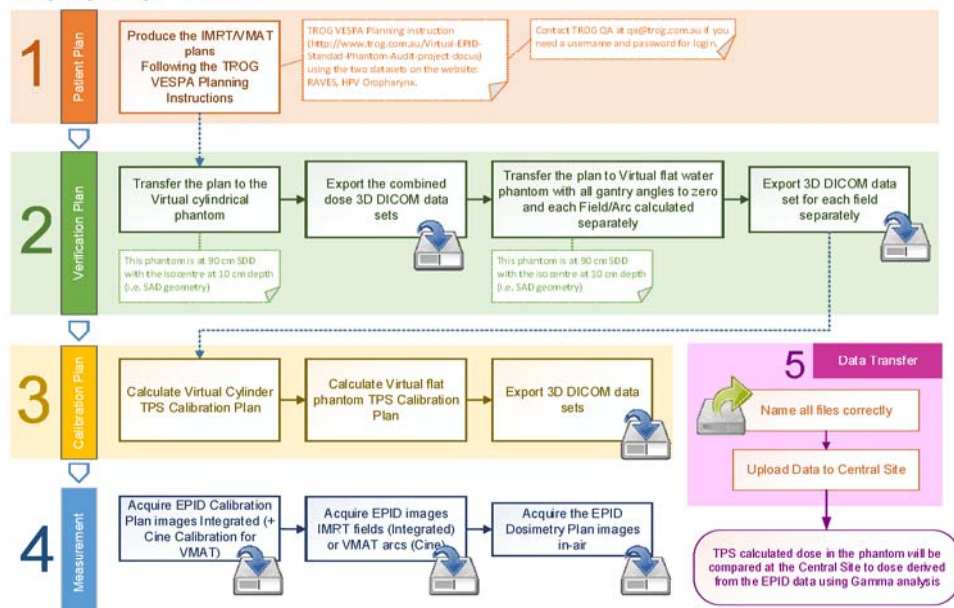
A three-dimensional Gamma calculation is calculated for the 3D dose distributions for the whole-volume by comparison to the TPS dose. For individual field analysis a 2D Gamma analysis is performed on the 2D dose planes at 10 cm depth in the virtual flat water phantom. Results for 2D Gamma results for coronal and sagittal planes through the centre of the phantom (isocentre) are also reported for interest but are not used for the final assessment.

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Step by Step Overview



Data Management

Site-identifiable results from dosimetry measurements for the purposes of this project will remain confidential and will not be released for external distribution or review. If a deviation is detected during this project, and repeat measurements confirm the deviation, the VESPA study group will provide assistance to identify the cause of the variation, but total responsibility for further action remains that of the treatment centre. The VESPA study group does not assume any reporting responsibility (other than to the site physicist and Director of Radiation Oncology) or regulatory role.

Relevant Documents

Download from [TROG VESPA](#) web-page

- Planning Guidelines (.pdf)
- TROG 08.03 RAVES Benchmarking Case (.zip)
- TROG 12.01 HPV Benchmarking Case (.zip)
- VESPA Virtual Flat Phantom (.zip)
- VESPA Virtual Cylindrical Phantom (.zip)

Provided by email/Cloud:

- Procedure Guide (i.e. This Procedure) (.pdf)
- Epid Guide (.pdf)
- Data Entry Sheet (.xls)
- TPS Calibration Plans and EPID Delivery Plan Dicom Files zip files
- Empty folder structure zip file
- (Virtual Phantoms are also on Cloud site)

Overview of Delivered and Exported Plans

This table shows the plans that must be calculated or delivered for each Audit. There is a lot of commonality for the IMRT/VMAT procedure.

<u>Plan Name</u>	<u>Dose Exported</u>	<u>Plan Delivered</u>
TPS Calibration Plan Virtual Cylinder	✓	X
TPS Calibration Plan Virtual Flat Phantom	✓	X
Verification Plan Virtual Flat Phantom	✓	✓ (IMRT only)
Verification Plan Virtual Cylinder	✓	✓
EPID Calibration Delivery Plan	X	✓
EPID Calibration Cine Mode Delivery Plan	X	✓ (VMAT, Clinac only)
EPID Dosimetry Delivery Plan	X	✓

Table 5 Plans to Deliver and Export Doses

“Trial” Plans

- Plan two IMRT or VMAT (depending on audit) plans following the TROG VESPA Planning Instructions (RAVES, HPV Oropharynx).
- Use the energy and dose-rate used at your centre for IMRT/VMAT treatments. Currently VESPA can accommodate 6 MV and 10 MV but not higher energies or FFF modes.
- Plan these **with the isocentre centrally within the high dose region** if at all possible. The EPID dose conversion method is most accurate close to isocentre and is not as effective if the dose distribution is away from isocentre.
- Note that the EPID size limitations (at the isocentre plane) are:
 - Varian (± 14 cm from central axis in sup-inf direction at 105 cm and ± 19 cm in left-right direction),
 - Elekta (± 12.5 cm from central axis in sup-inf, left-right directions at 160 cm). Do not exceed these dimensions (at the isocentre level) with the plans.

IMRT Audit Only

- Please export the Total Trial Plan Dose (RD), Plan file (RP) and Structure file (RS) as well as the individual Trial Plan Field Doses (Separate RD. Dose File for each IMRT Field e.g. RD.HPV_Trial_130, RD.HPV_Trial_180 etc)
- This will allow us to conduct further research on DVH type analysis.

TPS Calibration Plans

TPS Calibration Plan Virtual Cylinder

- A DicomRT Patient is provided to import into the TPS/R&V to setup these Plans (See Appendix C). Otherwise follow instructions below. Separate plans may be required on some TPS systems for each field.
- Change fields to your IMRT/VMAT energy and dose-rate.
- Manual setup.
 - Attach a 10x10 cm² beam to the **Virtual Cylinder**
 - The isocentre for all plans should be at the midpoint of the phantom 10 cm depth. This is DICOM zero position.
 - gantry=0, collimator=0, Fixed 100 MU.
 - Your IMRT/VMAT energy and dose-rate
 - Use Fractions = 1.
 - Use a high resolution grid eg 0.15 or 0.2 cm³.
 - Make sure to include at least 4 cm of the out-of-field penumbra dose on each side of the field if possible
 - Repeat for fields 25x25 cm² and 40x40 cm², 100 MU each field.
 - Attach a “four-field box” plan with each field a 10x10 cm² field with 100 MU with cardinal gantry angles, 0, 90, 180 and 270 degrees.
- Calculate dose – make sure that MU’s are correct as below.
- Record the doses at the isocentre on the Data Entry Sheet.
- Export the 3D DICOM RT data sets. If possible a single plan export with separate field doses exported, but can export separate plans. Use RD. for the dose files and RP. for the plan files. **Do not export Dose Planes.**
- In Monaco these fixed MU plans may be difficult to create. An approach to this is to calculate one plan with a beam in it and then create a QA plan from this. In these QA plans beams can be added and MU set.
- Note - **this plan is not delivered at the linac.**

Filename		MU	Field size jaws (cm ²)	Gantry	Coll
RD.Cyl_10x10	RP.Cyl_Cal	100	10x10	0	0
RD.Cyl_25x25		100	25x25	0	0
RD.Cyl_40x40		100	40x40	0	0
RD.Cyl_Box	RP.Cyl_Box	100 each field	10x10	180, 90, 0, 270	0

Table 6 Virtual Cylinder TPS Calibration Plan doses to export

TPS Calibration Plan Virtual Flat Phantom

- A DicomRT Patient is provided to import into the TPS/R&V to setup this Plan (See Appendix C). Otherwise follow instructions below. Separate plans may be required on some TPS systems for each field.
- Change fields to your IMRT/VMAT energy and dose-rate.
- Manual setup.
 - Attach open field jaw defined fields as below to the **Virtual Flat Phantom** with
 - isocentre at 10 cm depth, DICOM zero position.
 - gantry=0, collimator=0, fixed 100 MU.
 - IMRT/VMAT energy and dose-rate.
 - Use Fractions = 1.
 - Use a high resolution grid eg 0.15 or 0.2 cm³.
 - Make sure to include at least 4 cm of the out-of-field penumbra dose on each side of the field if possible
- Record the dose calculated by the TPS at the isocentre on the Data Entry Sheet for the 10x10 field. This field is essential for calibration.
- Export the 3D DICOM RT data sets. If possible a single plan export with separate field doses exported, but can export separate plans. Use RD. for the dose files and RP. for the plan files. Do not export Dose Planes.
- Note - this plan is not delivered at the linac.

Filename		MU	Field size jaws (cm ²)	Gantry	Coll
RD.Flat_2x2	RP.Flat_Cal	100	2x2	0	0
RD.Flat_3x3		100	3x3	0	0
RD.Flat_4x4		100	4x4	0	0
RD.Flat_6x6		100	6x6	0	0
RD.Flat_10x10		100	10x10	0	0
RD.Flat_15x15		100	15x15	0	0
RD.Flat_20x20		100	20x20	0	0
RD.Flat_25x25		100	25x25	0	0
RD.Flat_40x40		100	40x40	0	0

Table 7 Virtual Flat Phantom TPS Calibration Plan Doses to Export

IMRT/VMAT Verification Plans

Verification Plan Virtual Cylinder

Overview

- Import the 3D data set for the Virtual Cylindrical Phantom (VCP) into the TPS.
- Transfer the IMRT/VMAT plan on to the cylindrical VCP i.e. create a verification plan of the combined 3D dose distribution

Import and setup of Virtual Cylinder

- A 3D data set called "VCP" represents a cylindrical water-equivalent volume with diameter 20.0 cm. This can be downloaded from TROG Website
- Import this into the TPS. If the HU value is not equal to water, then override to water within the phantom. Alternatively if the relative or physical density is not equal to 1 then override to 1 within the phantom. The phantom is an RT Structure set and may need density override.
- Do **not** add a couch. There will be no couch present for the EPID in-air measurements. If asks to remove couch do nothing.

Procedure

- Create verification plan with all fields combined to give the total dose distribution on the cylindrical phantom.
 - The isocentre for all plans should be at the midpoint of the phantom 10 cm depth. This is DICOM zero position
 - Do not modify collimator angles or gantry angles.
 - IMRT/VMAT energy and dose-rate
 - Do not zero the collimator.
 - Use Fractions = 1.
 - **Use a high resolution grid eg 0.15 or 0.2 cm³.**
 - Make sure to include **at least 4 cm** of the out-of-field penumbra dose on each side of the field if possible

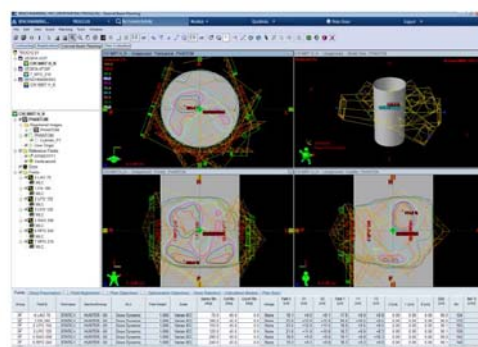


Figure 4 Illustration of combined IMRT fields planned on the virtual cylindrical phantom in Eclipse.

- Once the verification plans have been created, export the 3D DICOM RT data set including RD (Combined) Dose File, RP Plan File (**Note –the Combined Dose File is required**)

- **Prepare this Verification Plan for delivery** at the linac using actual gantry angles
- IMRT use Integrated imaging (see EPID Guide)
- Cine Mode imaging (Varian Clinac) or IPS imaging (Varian Truebeam) for VMAT imaging.(see EPID guide for precise imaging instructions for each equipment combination).
- Make sure that the EPID position is the same as used for the Calibration Delivery Plan images.

Filename		MU	Field size jaws (cm ²)	Gantry	Coll
RD.Cyl_RAV	RP.Cyl_RAV	actual	actual	planned	planned
RD.Cyl_HP	RP.Cyl_HP	actual	actual	planned	planned

Table 8 Cylindrical Phantom Dose Files to export

Verification Plan Virtual FLAT Phantom

Import and setup of Virtual Flat Phantom

- A 3D data set called “VFP” represents a flat water-equivalent volume phantom (zzz9999). This can be downloaded from the TROG VESPA website page.
- Import this into the TPS. If the HU value is not equal to water, then override to water within the phantom. Alternatively if the relative or physical density is not equal to 1 then override to 1 within the phantom. The phantom is comprised of an RT Structure set and may need density override.
- Do **not** add a couch. There will be no couch present for the EPID in-air measurements. If asks to remove couch do nothing.

Procedure

- Create a verification plan with all gantry angles at zero. [For information on how to collapse angles for VMAT arcs see Appendix D.](#)
- If possible use a **Single Verification Plan** with all the fields and export each Field/Arc dose separately with the Single Plan file. This is much more efficient for analysis. (If necessary you can create separate verification plans for each Field/Arc).
 - At gantry zero incidence on the phantom
 - **Isocentre placed at 10 cm depth, DICOM zero**
 - IMRT/VMAT energy and dose-rate
 - **Make sure that MU are the treatment MU**
 - **Do not zero the collimator.**
 - Use Fractions = 1.
 - **Use a high resolution grid eg 0.15 or 0.2 cm³.**
 - Make sure to include **at least 4 cm** of the out-of-field penumbra dose on each side of the field if possible

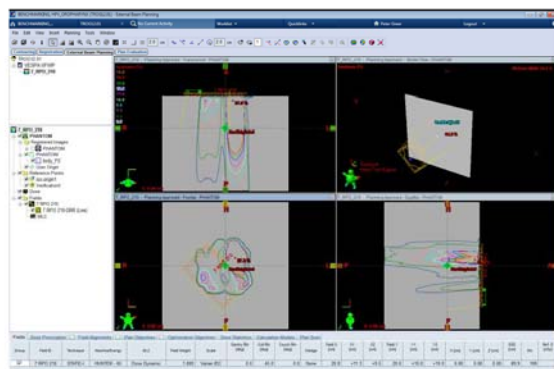


Figure 5 Illustration of an IMRT beam planned on the virtual flat phantom in Eclipse.

- Once the verification plans have been created, **export the DICOM RT data set** including RD. Dose files for each Field/Arc separately and the RP. Plan file as in **Table 9. (Not Dose Planes).**

- IMRT '###' = Trial Plan Field gantry angle, followed by _0 to denote at gantry zero, If there are two fields at the same gantry angle use _1 and _2 e.g. 'RD.Flat_RAV_135_1_0' & 'RD.Flat_RAV_135_2_0'
- VMAT '###' = 'Arc#' e.g. 'Arc1', 'Arc2' etc

IMRT Audit

- **Prepare this plan for delivery at the linac** to record on EPID with Integrated imaging.

VMAT Audit

- **This plan is not delivered at the linac** for VMAT. Only TPS doses are required.

Filename		MU	Field size jaws (cm ²)	Gantry	Coll
RD.Flat_RAV_###_0	RP.Flat_RAV	planned	planned	0	planned
...		...	...	...	...
RD.Flat_HP_V_###_0	RP.Flat_HP_V	planned	planned	0	planned
...		...	...	...	...

Table 9 Flat Phantom Field Dose Files to export

EPID Delivery Plans

EPID Calibration Delivery Plan

- A DicomRT Patient is provided to import into the TPS/R&V to setup this Plan. This may need to be adapted to local equipment, dose-rate etc. See Appendix C. Otherwise create this.
- **Make sure to set the dose-rate and energy here to be the same as for your IMRT/VMAT delivery.**
- Note that the CWD image field size is set to 40x30 cm in this plan and will have to be changed for Elekta, aS1200 etc as per below.
- **Setup this plan for delivery - Integrated mode** imaging (see Epid Guide for details).
- No dose export is required.

Image Filename	MU	Field size jaws (cm ²)	Gantry	Coll	Purpose
RI.CWD_0_100	100	Varian aS1000 X=40,Y=30 cm Varian aS1200 X=Y=40 cm Elekta X=Y=23 cm	0	0	Uniformity
RI.C10x10_0_090_100	100	10x10	0	90	Calibration and CAX position
RI.C10x10_0_270_100	100	10x10	0	270	Calibration and CAX position
RI.CWD_0_20	20	Elekta X=Y=23 cm	0	0	Uniformity Elekta
RI.C10x10_0_090_20	20	10x10	0	90	Calibration and CAX Elekta
RI.C10x10_0_270_20	20	10x10	0	270	Calibration and CAX Elekta
RI.C10x10_180	100	10x10	180	0	EPID Sag/Flex
RI.C10x10_135	100	10x10	135	0	EPID Sag/Flex
RI.C10x10_90	100	10x10	90	0	EPID Sag/Flex
RI.C10x10_45	100	10x10	45	0	EPID Sag/Flex
RI.C10x10_0	100	10x10	0	0	EPID Sag/Flex
RI.C10x10_315	100	10x10	315	0	EPID Sag/Flex
RI.C10x10_270	100	10x10	270	0	EPID Sag/Flex
RI.C10x10_225	100	10x10	225	0	EPID Sag/Flex
RI.C10x10_180E	100	10x10	180	0	EPID Sag/Flex

Table 10 EPID Calibration Delivery Plan Fields/Images

EPID Calibration Cine Mode Delivery Plan (only Clinac)

- This is only required for VMAT on the Clinac
- A DicomRT Patient is provided to import into the TPS/R&V to setup this Plan. This may need to be adapted to local equipment, dose-rate etc. See Appendix C. Otherwise create this.
- **Make sure to set the dose-rate and energy here to be the same as for your VMAT delivery.**
- **Setup this plan for delivery – Cine mode mode** imaging (see Epid Guide for details).
- No dose export is required.

<u>Image Filename</u>	<u>MU</u>	<u>Field size jaws (cm²)</u>	<u>Gantry</u>	<u>Coll</u>	<u>Purpose</u>
RI.CWD_0_100	100	X=40,Y=30 cm	0	0	Cine-Mode (WD=Whole-Detector)
RI.### RI.### RI.### . .	300	10x10	0	0	Cine-Mode

Table 11 EPID Cine-Mode Calibration Delivery Plan Fields

EPID Dosimetry Delivery Plan

- A DicomRT Patient is provided to import into the TPS/R&V to setup this Plan. This may need to be adapted to local equipment, dose-rate etc. See Appendix C. Otherwise create this.
- **Make sure to set the dose-rate and energy here to be the same as for your IMRT/VMAT delivery.**
- **Setup this plan for delivery - Integrated mode imaging** (see Epid Guide for details).
- No dose export is required.

<u>Image File Name</u>	<u>MU</u>	<u>Field size jaws (cm²)</u>	<u>Gantry</u>	<u>Coll</u>	<u>Purpose</u>
RI.E10x10_5	5	10x10	0	0	Linearity
RI.E10x10_10	10	10x10	0	0	Linearity
RI.E10x10_20	20	10x10	0	0	Linearity
RI.E10x10_30	30	10x10	0	0	Linearity
RI.E10x10_40	40	10x10	0	0	Linearity
RI.E10x10_60	60	10x10	0	0	Linearity
RI.E10x10_80	80	10x10	0	0	Linearity
RI.E10x10_100	100	10x10	0	0	Linearity
RI.E10x10_150	150	10x10	0	0	Linearity
RI.E10x10_200	200	10x10	0	0	Linearity
RI.E10x10_400	400	10x10	0	0	Linearity
RI.E2x2	100	2x2	0	0	Field size response
RI.E3x3	100	3x3	0	0	Field size response
RI.E4x4	100	4x4	0	0	Field size response
RI.E6x6	100	6x6	0	0	Field size response
RI.E10x10	100	10x10	0	0	Field size response
RI.E15x15	100	15x15	0	0	Field size response
RI.E20x20	100	20x20	0	0	Field size response
RI.E25x25	100	25x25	0	0	Field size response

Table 12 EPID Dosimetry Delivery Plan Fields/Images

Linac EPID Measurements

EPID Preparation

- Read the EPID Guide (EG) before continuing
- For this audit Integrated Mode imaging must be setup for the energy and dose-rate used for IMRT/VMAT delivery.
- Also set up cine mode imaging for VMAT audit with Clinac.
- Dark and Flood-Field Calibration is recommended.
- All images are acquired in-air with no couch or phantom present and no added buildup on the EPID.
Once positioned do not move EPID. Same EPID position for all images.
- 105 cm for Varian Clinacs (100 cm for Truebeam). Varian use console controls **not pendant**.
- 160 cm for Elekta and carefully ensure at zero position.
- Position the EPID (**and do not move it again**):
- **The EPID Calibration Plan(s) and IMRT/VMAT Verification Plans must be delivered at the same session on the same linac.**

IMRT Audit (All Integrated Imaging)

The EPID calibration plan images and the IMRT images must be at the **SAME ENERGY AND DOSE-RATE** and **measured in the same session** with same EPID position.

Varian

Clinac

- Use the clinical system to irradiate all images. (This is important as the systems can apply different scaling to the images in the Clinical system. **Do not use AM-Maintenance.**)
- For Mosaic users the DICOM images must be captured in the Cache on the 4DITC.

Truebeam

- For ARIA use clinical Dosimetry Imaging for every image and export DICOM from ARIA.
- For Mosaic use clinical Dosimetry Imaging with IPS running. Export last IPS image frame for integrated images in DICOM format.

Elekta

- For Elekta use the iView software for Integrated Imaging. Follow instructions in EPID guide. Elekta= HIS16bit file format + log file, See EG for details.

IMRT Verification Plan Virtual Flat Phantom

- First record Integrated Images for the IMRT fields at **gantry zero** with the clinical MU, jaw, collimator positions.

File Name	Gantry	Coll	Comment
RI.Flat_RAV_###_0	0	planned	Gantry zero
...	...	...	...
RI.Flat_HP_V_###_0	0	planned	Gantry zero
...	...	...	...

Table 13 Virtual Flat Phantom IMRT Verification Plan fields at Gantry Zero

EPID Calibration Delivery Plan

- Record **Integrated** images as outlined in Table 10.
- **Use IMRT energy and dose-rate.**

IMRT Verification Plan Virtual Cylindrical Phantom

- Record **Integrated** images for the IMRT delivery at **actual planned gantry angles** (IMRT – ‘###’=field gantry angle), with the clinical MU, and jaw positions etc.

<u>File Name</u>	<u>Comment</u>
RI.Cyl_RAV_###	planned gantry
...	...
RI.Cyl_HP_V_###	planned gantry
...	...

Table 14 Virtual Cylinder IMRT Verification Plan Fields at actual Gantry angles

EPID Dosimetry Delivery Plan

- Record **Integrated** images as outlined in Table 12.
- **Use IMRT energy and dose-rate.**
- Gantry zero and collimator zero
- Wait at least 10 seconds between each image acquisition.
- Export the flood-field calibration image if possible (Varian only, see EG).

VMAT Audit (Varian only)

The calibration field images and the VMAT images must be at the **SAME ENERGY AND DOSE-RATE** and measured in the same session with same EPID position.

Clinac

- For Clinac use the clinical system to irradiate and record all the images. **Do not use AM-Maintenance**

VMAT audit with ARIA (Aria)

- For integrated images use clinical imaging and export DICOM from ARIA
- For Cine Mode Calibration and VMAT delivery use Cine mode and export DICOM from Aria.

VMAT audit with Mosaic (Cache)

- For integrated images Use clinical imaging and get the DICOM images from the Cache on the 4DITC.
- For Cine Mode Calibration and VMAT delivery use Cine mode and get the DICOM images from the Cache on the 4DITC

Truebeam

VMAT audit with ARIA (Clinical and IPS)

- For integrated images use clinical dosimetry imaging and export DICOM from ARIA. (IPS last frame can also be used).
- For VMAT delivery use clinical dosimetry imaging with IPS running and export all image frames.

VMAT audit with Mosaic (IPS)

- For integrated images Use clinical dosimetry imaging with IPS running and export last IPS image for integrated images in DICOM format.
- For VMAT delivery use clinical dosimetry imaging with IPS running and export all image frames.

EPID Calibration Delivery Plan

- Record **Integrated** images as outlined in Table 10.
- Use VMAT energy and dose-rate.**

EPID Calibration Cine Mode Delivery Plan (only Clinac)

- Record **Cine Mode** images (Table 11). Use the clinical system. **Do not use AM Maintenance.**
- Use VMAT energy and dose-rate.**
- If system overflow occurs on the Clinac when acquiring cine images see Epid Guide.
- Do not rename the individual cine images.

VMAT Verification Plan Virtual Cylindrical Phantom

- Cine Mode images (Clinac) and Dosimetry Mode IPS image frames (Truebeam)
- Record images for the VMAT arcs (– ‘###’=Arc1, Arc2 etc) with the clinical MU, and jaw positions etc.
- Use the **clinical system** (do not use AM-Maintenance).
- Do not rename the individual cine images/frames, place in a folder named as below. If system overflow occurs when acquiring Clinac images see EG.

<u>File Name</u>	<u>Comment</u>
RI.Cyl_RAV_Cine_###	
...	...
RI.Cyl_HP_V_Cine_###	
...	...

Table 15 Virtual Cylinder VMAT Verification Plan Fields at actual Gantry angles

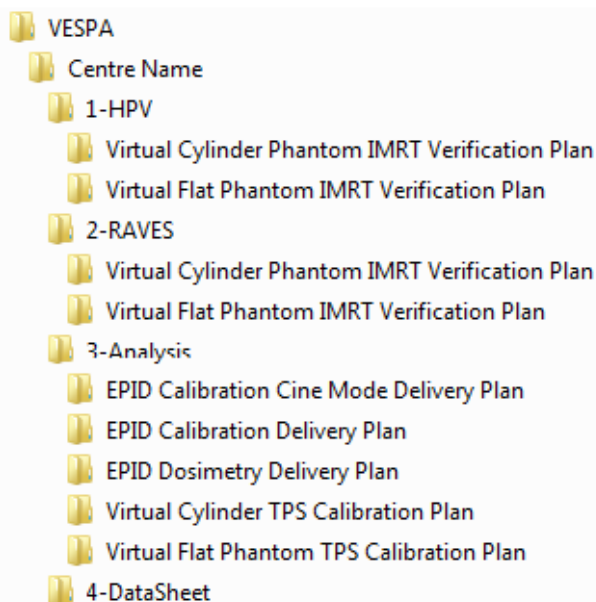
EPID Dosimetry Delivery Plan

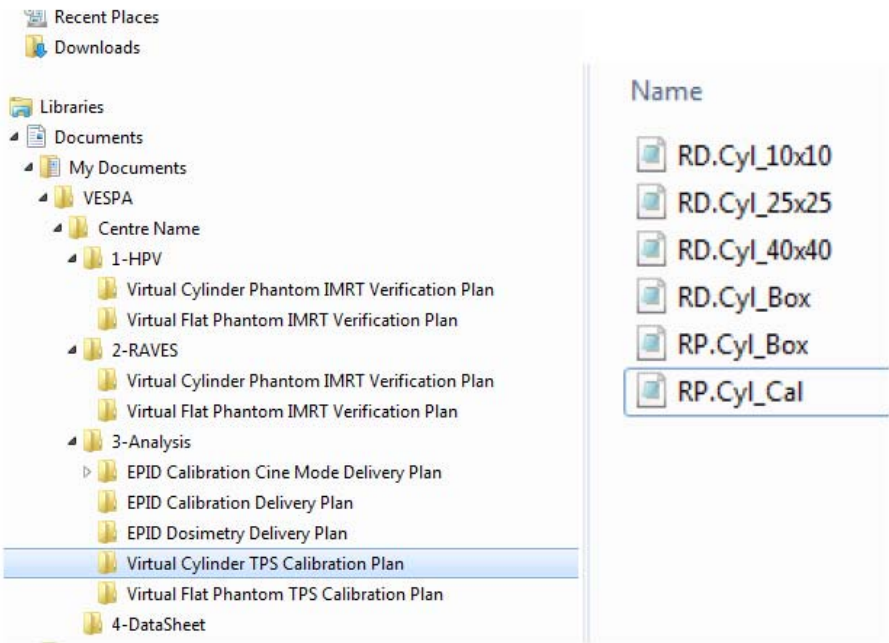
- Record Integrated Mode images as outlined in Table 12.
- **Use VMAT energy and dose-rate.**
- Gantry zero and collimator zero
- Wait at least 10 seconds between each image acquisition.
- Export the flood-field calibration image if possible (Varian only, see EG).

Appendix A - Data Files for Upload

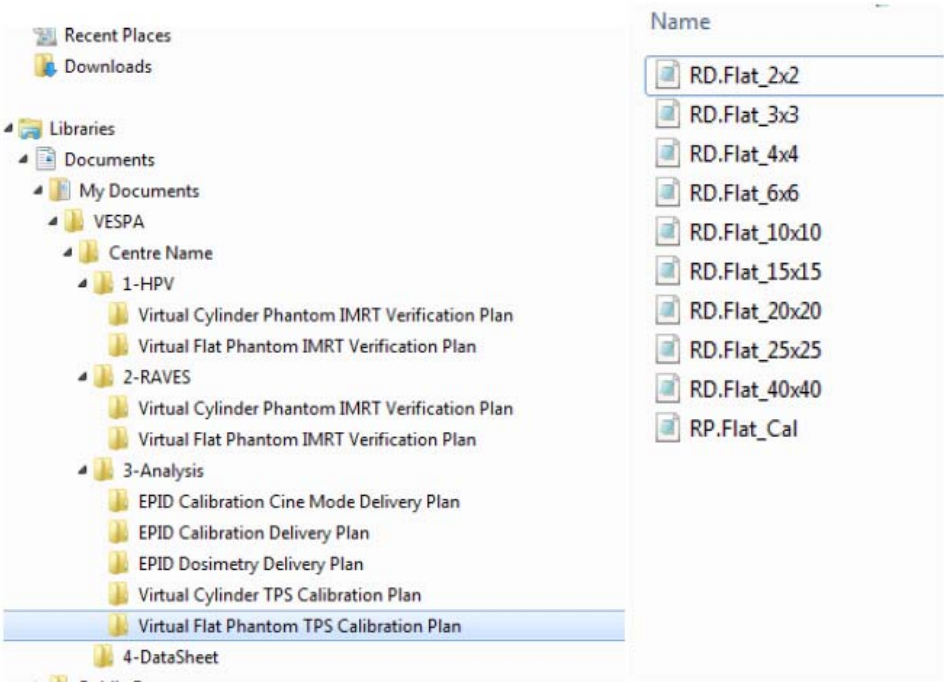
Please put files in the folder structure outline below and make sure that all files are uploaded. For Elekta the image files will be .his/.log pairs not DICOM.

Also include the Datasheet please.

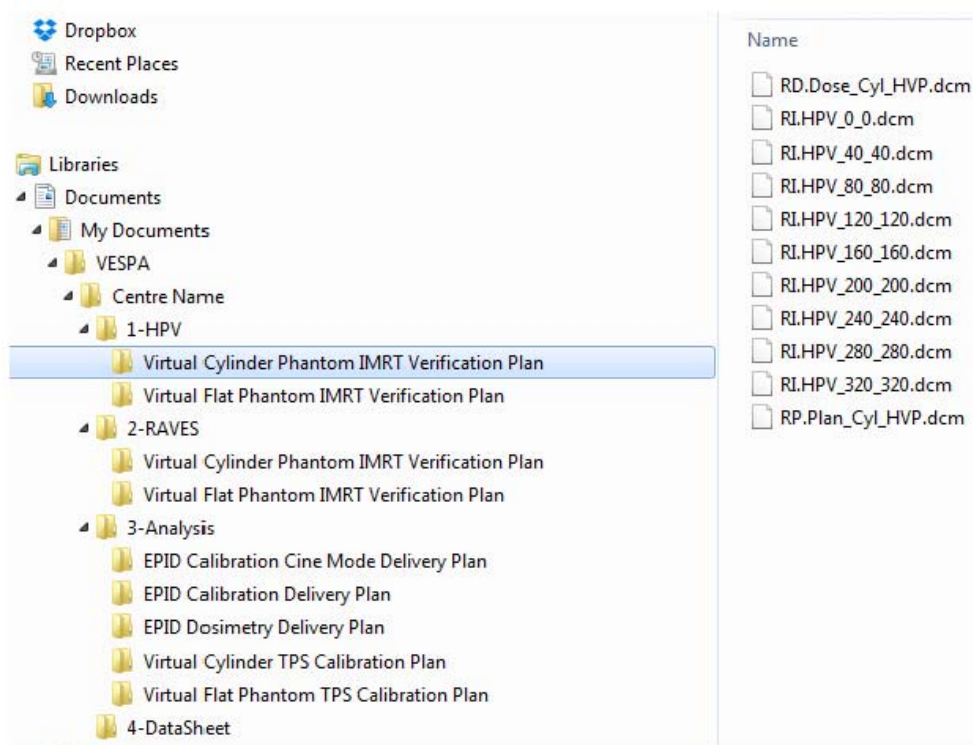




Virtual Cylinder TPS Calibration Plan Files (note multiple plan files for _Box and _Cal plans is OK)



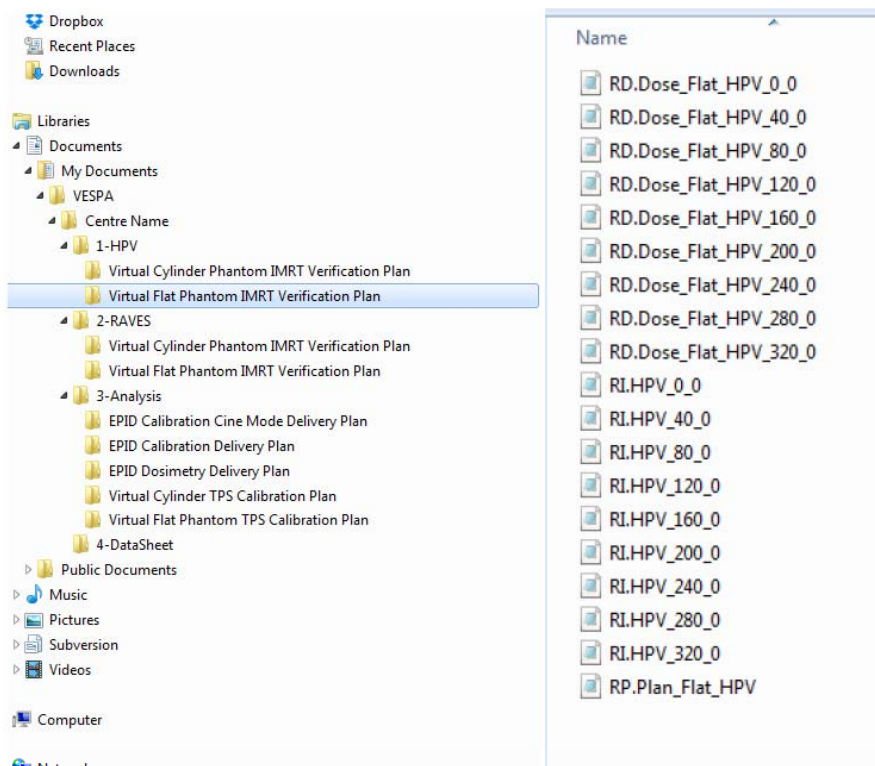
Virtual Flat Phantom TPS Calibration Plan Files (note multiple plan files is OK)



Virtual Cylinder IMRT Verification Plan (IMRT audit)

Same as above for IMRT except with Cine images for each arc

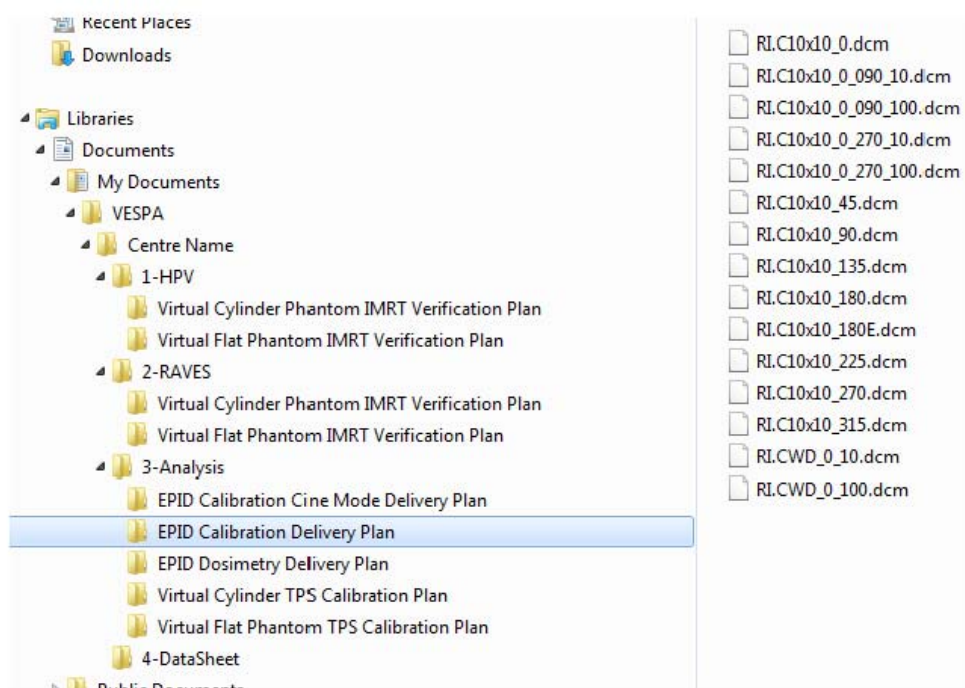
Virtual Cylinder VMAT Verification (VMAT audit)



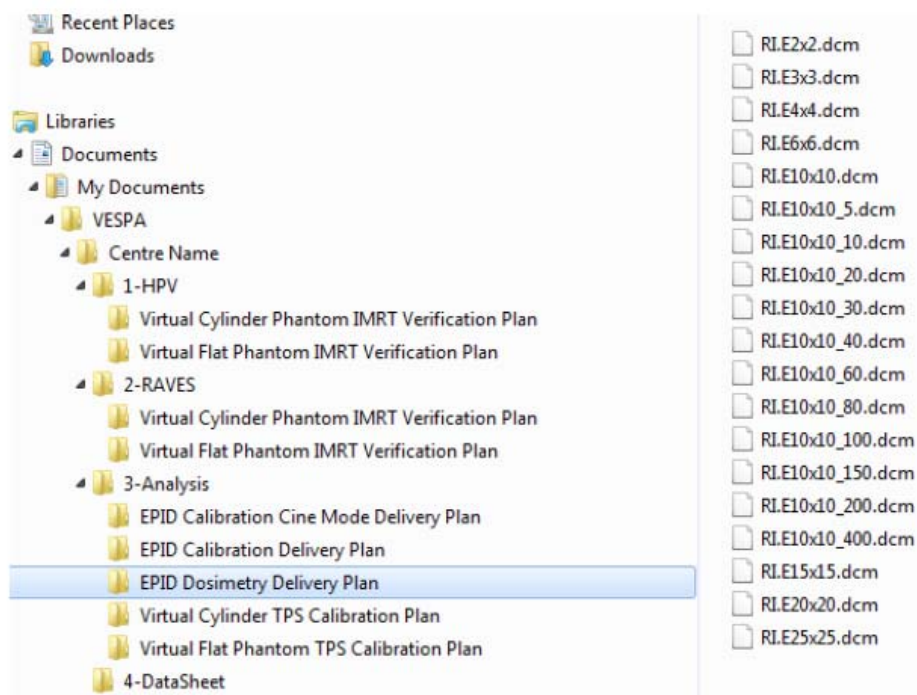
Virtual Flat Phantom IMRT Verification Plan Files (IMRT audit)

Just the RD dose files for the arcs and the plan files. We will create the integrated arc images from the cine images.

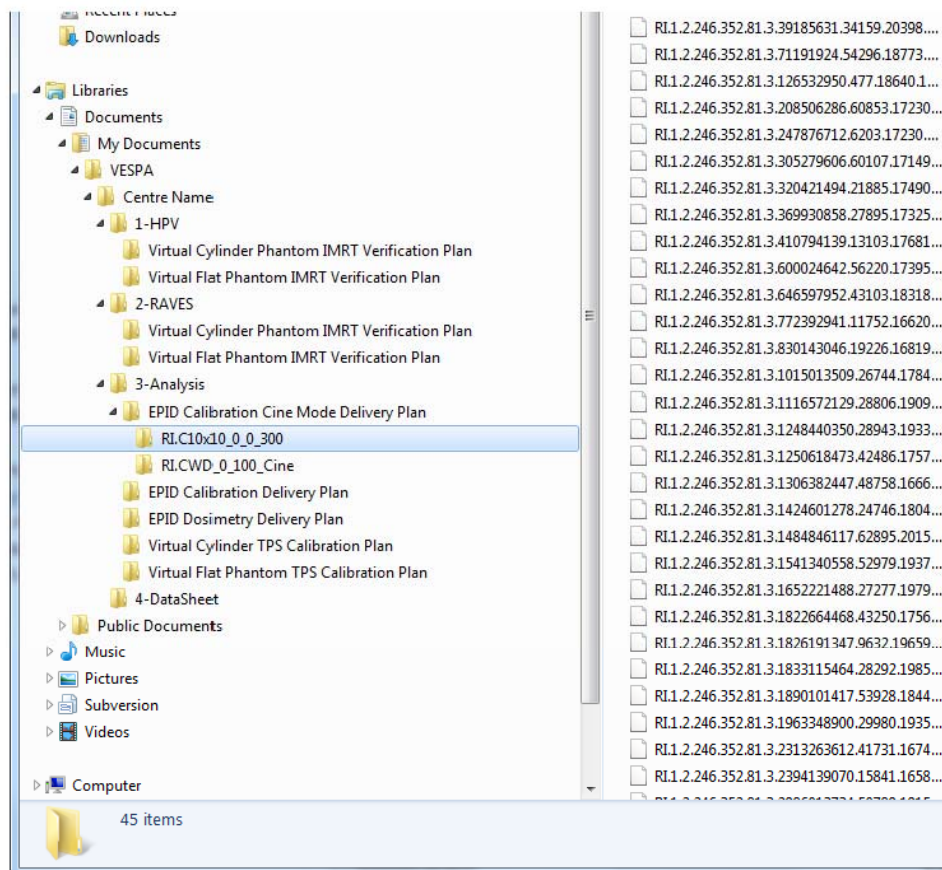
Virtual Flat Phantom VMAT Verification Plan Files



EPID Calibration Delivery Plan Files



EPID Dosimetry Delivery Plan Files



EPID Calibration Cine Mode (Varian Clinac Only) Delivery Plan Files

Appendix B - Data Uploading

- Make sure you have all the data files above.
- There should be no actual patient names in these files.
- Once you are ready contact P. Greer who will send a link for uploading to OwnCloud.

Appendix C – Importing DicomRT files

Importing the Calibration Patients

- There are two “Patient Sets” with different plans attached.
- The major one has four plans attached to the Virtual Flat Phantom

Plan Name	Plan Description
VFP_TPS_Cal	Virtual Flat Phantom TPS Calibration Plan
EPID_Cal_Del	EPID Calibration Delivery Plan
EPID_Cine_Del	EPID Calibration Cine Mode Delivery Plan
EPID_Dos_Del	EPID Dosimetry Delivery Plan

- The second is has two plans attached to the Virtual Cylinder Phantom

Plan Name	Plan Description
VCP_TPS_Cal	TPS calibration fields
VCP_TPS_Box	TPS four-field box

- The Machine Name and Tolerance Table should have been setup for your files so that they will import fully.
- Use Import Patient to load the Virtual Flat Phantom with these DICOM plans
- The DICOM plan files alone may also be able to be imported onto an existing Virtual Flat Phantom patient or existing Virtual Cylinder Patient if required.
- For the plans that require delivery set up EPID position, EPID imaging templates, Plan Approval, Scheduling etc. so that these Plans can be delivered. Make sure to attach Cine Mode where required.
- **Adapt the dose-rates of the plans where necessary, i.e. if you deliver IMRT at 400 MU/min then the EPID Calibration Delivery Plan needs to be performed at 400 MU/min as well. If you are doing VMAT at 600 MU/min then the calibration deliveries need to be at 600 MU/min.**

Appendix D – Collapsing VMAT plans to gantry zero

Eclipse

- When creating the verification plan select Set Gantry to zero. Normally warnings will be given about delivering this field however the dose should calculate.
- The dose from each Arc should be exported separately. This can be done by selecting “Place each field in a separate verification plan” or by exporting field doses.

Pinnacle

- A Pinnacle Script is required to do this. Contact the Central Site for more information if this is required.

Monaco

- To be advised

Document History

<u>Date</u>	<u>Version</u>	<u>Author</u>	<u>Changes</u>
2015		PG	Reordered the procedure for clarity Clarifications to text. Removed references to Dicom plans for Open field dose plans. Changed to 40x40 field size. Added more gantry sag fields
			Removed DVH text
			Changed dose file naming RD etc
			Changed field naming to just gantry angle. Added 6 and 10 MV specification Naming of fields Removed excessive TPS field size now consistent VFP and VCP Raves to RAV
2015-09-24	2.0	PG	Combined IMRT and VMAT with colour coding
2016-02-11	2.1	PG	Minor change to export of Trial plan doses beam by beam
2016-30-30	2.2	PG	Added folder structure for file upload zip file
2016-05-26	2.3	PG	Removed checklist to separate doc
2016-10-12	4.0	PG	Removed text about DICOM RT mode. Updated the import plans text. Removed colour coding and separated out IMRT and VMAT. Much more detail on type of imaging to use.
2016-10-24	4.1	PG	Changed Virtual Flat Phantom to single verification plan Some reordering. Changed image files in tables to RI. Added back export of RD field doses for trial plan just for IMRT Specified that can export single plan file for TPS_Cal plans
2017-04-26	4.11	PG	Minor correction to image names for cine calibration Minor formatting changes



Trans-Tasman Radiation Oncology Group

**Virtual EPID Standard Phantom Audit
(VESPA)**

EPID Guide

Version 4.01

26 September 2017

Team Contacts

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EPID dosimetry set-up for Varian aS500/aS1000 on C-Series linacs (Clinacs)

Acquiring images in clinical system

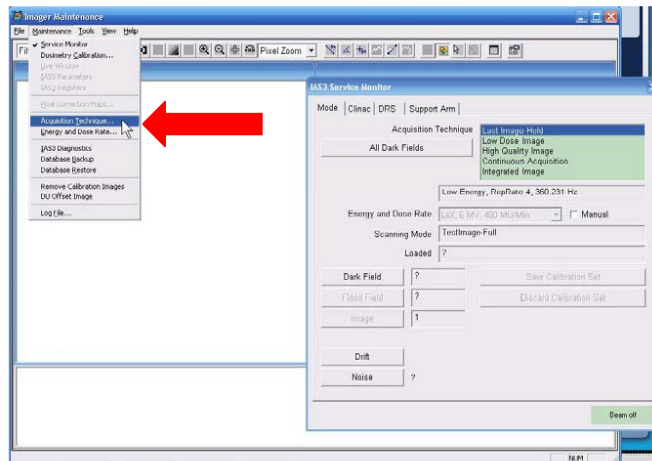
- This requires that the Integrated/IMRT Imaging Mode or Continuous/Cine Mode template be attached to the field before treatment.
- If you have Mosaic make sure to get images in the cache – see below.
- The EPID position must be entered into your R&V system to obtain (0,0,-5cm) isocentre position
- At treatment the EPID is deployed from the console area. Check that it has reached preset position. If the EPID deploys to a different (optimised) position then do not treat or image the fields. Go back and set the EPID position again in your R&V system.
- Save and export the acquired EPID images in DICOM format
- If using Aria these are most easily exported from Dicom Import/Export or ImageBrowser

Mosaic/Varian Clinac combination users

- Images from Varian EPID can be exported from Mosaic but insufficient is stored in the image headers. These can not be used for dosimetry.
- The images are saved in the cache while the patient is still open on the 4DITC.
D:\VMSOS\AppData\TritImaging\DSS\Primary\Cache\Session]
- If any difficulties are encountered please contact the Central Site for more information.

Setup of Cine-Mode imaging (VMAT only)

- Mode up the Linac with the energy and dose-rate using for VMAT otherwise AM-Maintenance imaging parameters will be empty (can mode up linac in Service Mode)
- Open AM-Maintenance software which is usually on the desktop of the 4DITC
- Go to the “Maintenance” menu and select “Acquisition Technique”:



- Then select “Continuous Acquisition” (see below).

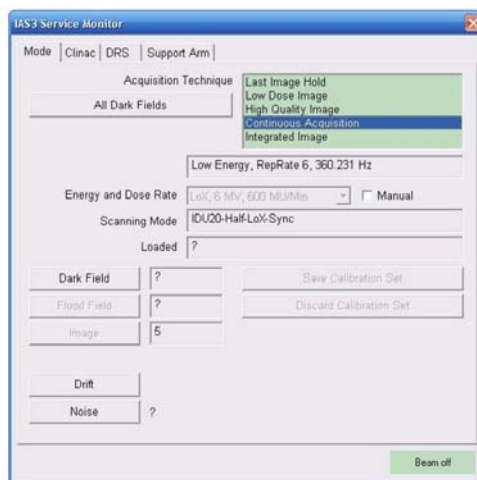
- Make sure the Energy is that used for VMAT e.g. LoX, 6MV
- Change the Resolution to **Full** if it is set to Half (aS1000)
- Write down the number of frame averages displayed.
- Change the frame averages to 5.
- Click OK
- Dark field and flood-field calibration may now be required if this mode is not already setup. Follow instructions below for DF/FF calibration below. Make sure you do this for Continuous Mode.
- Remember to set the EPID back to original settings if you use Continuous Mode for other imaging in your clinic.

System Overflow with Cine-Mode imaging

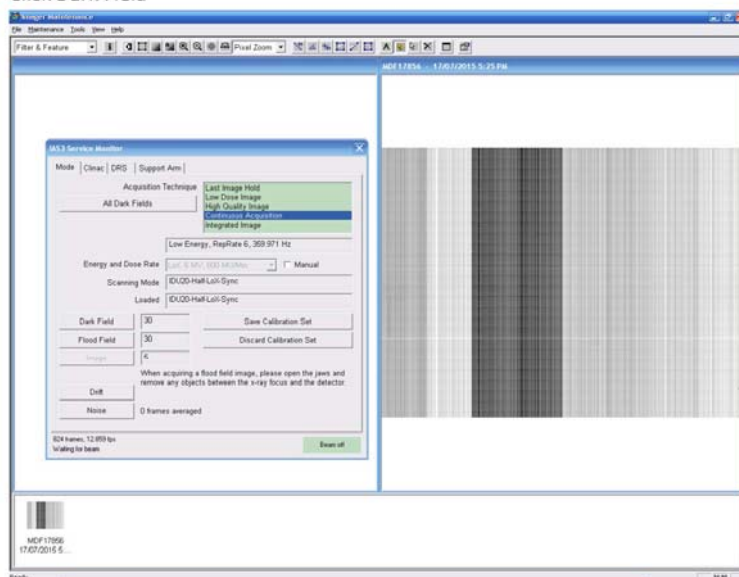
- Only a certain number of images can be held in the buffer at any one time. The frame-averages of 5 should be sufficient to keep within this limit. However if overflow does occur the following can be done then increase the number of frame-averages for cine mode.
- It may also be necessary to partially deliver the plan e.g. first arc, save the images, close the software, reopen and then deliver the second arc.

Dark Field and Flood-Field Calibration (C-Series Linac, AM-Maintenance Software)

- If using PortalDosimetry then note that this calibration can change EPID response and hence CU calibration
- Mode the Linac up with the energy and dose-rate used required (Service Mode is OK)
- Retract the couch so it is out of the way and will not collide with EPID when it is deployed.
- Open AM Maintenance which is usually on the desktop of the 4DITC
- Position the EPID to isocentre. To do this Click Service Monitor Support Arm tab.
- For isocentre enter 0 cm for all directions for Preset Position, Click Download
- Then hold down PV Motion Enable and Auto buttons on OBI and PV Positioning Console.
- Select the Mode tab on Service Monitor
- Select the Imaging Mode required (e.g. Integrated or Continuous Mode). Energy and Dose Rate should be picked up from the Linac



- Click Dark Field



- Gantry and Collimator to zero, jaws to Y=30, X=40 cm. This size will not irradiate the EPID electronics.
- Click Flood Field, should state waiting for beam.
- Dial up 300 MU and beam-on.
- When Flood-Field (FF) appears in window stop beam
- Check that Flood-field is not saturated using Profile Tool from corner to corner. Should be no flat signal areas. If images are saturated try starting again with EPID positioned 5 cm below isocentre.
- Click Save Calibration Set

- To test acquire a 100 MU Image (see instructions below) with the same field size. Select Integrated Image Mode (or Continuous Acquisition mode if testing this)
- Click Image, should state Waiting for beam below and the frames-per-second for this mode
- Beam-on.
- For Continuous Mode images pick an image in the middle of the set.
- This should be very flat with a low standard deviation (e.g < 0.5% for a region of interest covering most of the detector). Check that image is not saturated.
- File – Save All
- The images are stored in C:\Program Files\Varian\Oncology\Treatment\AM\images in a folder labelled with the date.

Flood-field location

- The FF image should be available in C:\Program Files\Varian\Oncology\Treatment\AM\images in a folder labelled with today's date.
- The latest FF being used by the system can also be found in C:\Program Files\Varian\Oncology\Treatment\AM\config\AM\IAS3\Data stored in .hnc format. These are stored as MFF-energy-dose-rate-IDU-resolution-mode eg. MFF-06-600-IDU20-Full-LoX-Sync is the Continuous Acquisition mode for VMAT and the MFF-06-XXX-IDU20-Full-LoX-Integrated is the Integrated Imaging mode for IMRT.

EPID dosimetry set-up for Varian XI aS1000 or aS1200 on Truebeam

Acquiring images in clinical system

- Imaging templates can be added on-the-fly if required but it is better to set these in Aria.
- If you have Mosaik see below for more information as IPS must be used.

IMRT audit with ARIA

- Use clinical dosimetry imaging for every image and export DICOM from ARIA.

IMRT audit with Mosaik (IPS)

- Use clinical dosimetry imaging with IPS running.
- Export last IPS image for integrated images in DICOM format.

VMAT audit with ARIA (Clinical and IPS)

- For integrated images Use clinical dosimetry imaging and export DICOM from ARIA.
- For VMAT delivery cine images use IPS running and export all image frames (there are lots).

VMAT audit with Mosaik (IPS)

- For integrated images Use IPS running and export last IPS image for integrated images in DICOM format.
- For VMAT delivery cine images use IPS running and export all image frames (there are lots).

Dark and Flood field calibration

- This is very straightforward in Service Mode.
- The lead site has Truebeam and can be contacted for any queries.

Use of IPS imaging on Truebeam (VMAT Aria/Mosaik and IMRT Mosaik)

- Truebeam has cine or movie type imaging but this is not useful for dosimetry. Do not use.
- Always **acquire a Dosimetry Mode image with Truebeam**.
- All integrated calibration images can be obtained by exporting the last “frame” from the IPS as this is the integrated image.
- Make sure you have the **Image Processing Service running** (see Figure 1). This will store the single (cumulative image) frames of an integrated image.

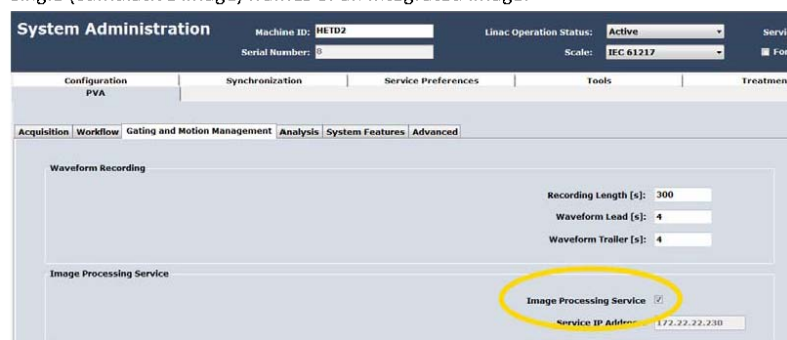


Figure 1 - Enabling the Image Processing Service

- (At your Truebeam, start major mode 'Developer' or 'Service'
- 1. Open IPS tab (see Figure 2)
- 2. Click **Refresh**
- 3. Select the acquisitions that correspond to the portal images you would like to convert.
Note: Type is "MV Triggered" for dosimetry images
- 4. From the export button **dropdown menu**, choose **DICOM** file format
- 5. Click on Export
- 6. Choose a location and click OK.
- 7. **Note:** For the VMAT delivery we need all the IPS frames. We will make cine-images from these.

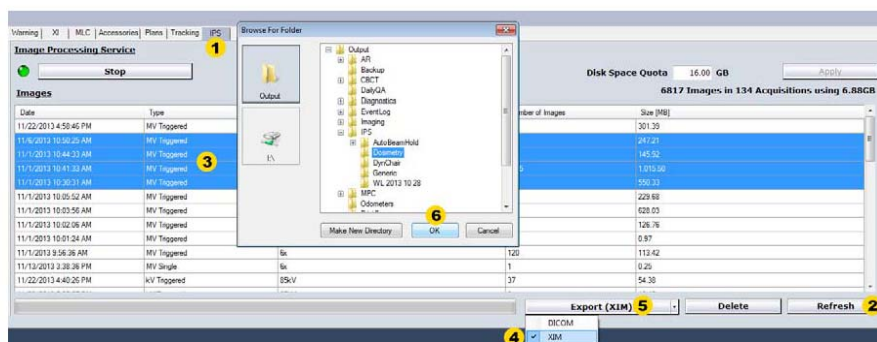


Figure 2 Image Processing Service

EPID dosimetry set-up for Elekta

Positioning EPID

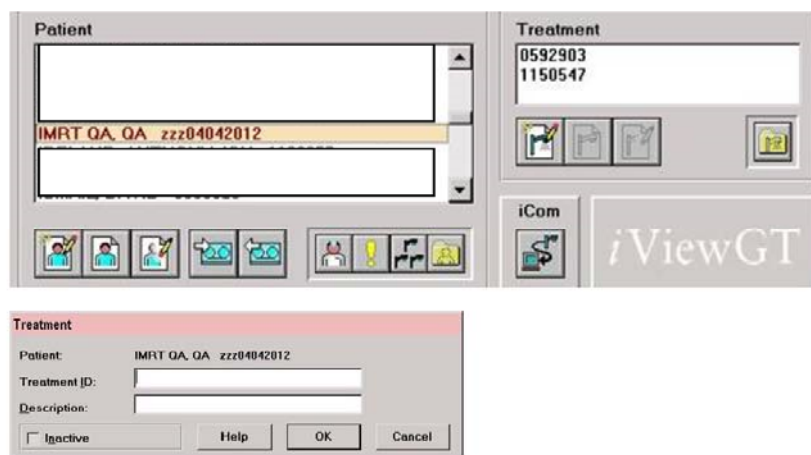
- Remove graticule from head
- Retract couch so is not in the beam
- Position EPID, EPID will stop at zero, do not keep pressing down or will not be at zero position
- Once positioned for the imaging session do not move the EPID.

Flood Field Calibrations

- It is **HIGHLY RECOMMENDED** that a Flood-Field (Gain) Calibration is Scheduled by the Center before the Vespa audit and that this calibration includes the Cine Mode Imaging)
- This is recommended to be performed by experienced users.

Acquire Integrated Images

- Note: The sri.ini file parameter IMRTDosimetricWeighting must be set = 1. This removes any normalization from the image and maintains the relative intensities of each segment. This should already be set-up correctly.
- This file is on the local C drive in C:\iView folder
- Calibration Fields can be delivered in Service Mode but IMRT fields should be acquired in Clinical Mode (receive external prescription). Do not acquire these images with the Clinical patient in iView, use a QA patient as described below.
- Before starting measurements a new treatment and new fields need to be created in iView. For example, create a patient called "IMRT QA" or "Vespa" and create a new treatment for each patient or test exposures.



- In iView, a field must be defined for each Calibration and Treatment field.
- Give each field a descriptive name (use Vespa Field Names from Tables).

Patient	Treatment	Field
IMRTQA zzz	0289310 0524245 0695473	10x10ref

- Acquire the Calibration fields using “single exposure”. With 0 IMRT segments:
- For the IMRT fields - At this point you must know how many segments are in the IMRT field and enter this value into “IMRT segments” per figure below (none of the other settings are important).

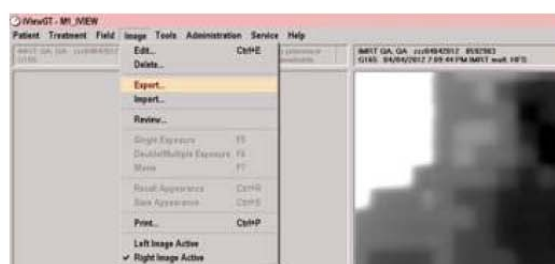
The 'Field' dialog box shows the following settings:

- Patient: IMRT QA, QA zzz04042012
- Treatment: 1150547
- Field ID: G140
- Description: (empty)
- Site: (empty)
- Orientation: Head First - Supine
- Sequence: 0
- Source to skin (cm): 0.0
- Imaging:
 - Gain (non-GT): Low (Normal)
 - Frame averaging: Maximum
 - iCom-Yx: None
- Setup:
 - Gantry angle (deg): 0
 - Dose rate (mu/min): (empty)
 - IMRT segments: 17 (highlighted)
 - Energy: Low
- Message log: (empty)
- Buttons: Help, OK, Cancel

- For the IMRT fields, select **multi-frame acquisition**.
- Acquire images in iView, beam on in QA mode with ‘port during’ exposure (Mosaiq). iView acquires an image for each IMRT segment, then automatically generates a composite image after all expected segments have been acquired – **On QA patient in iView**.

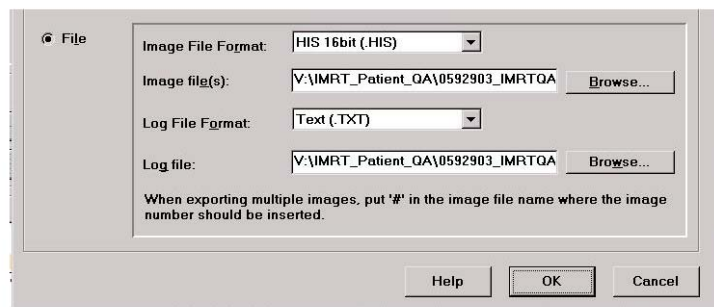
Export Images

After all fields are acquired, export images and associated log file(s) from iView per below:



- Images should be exported one at a time in .HIS format with associated log file.
- Select Field and also Select image in Box on far right.
- Image – Export
- Select This Image Only
- Choose HIS16bit image file format.** Do not export to DICOM server.

- Name the .HIS image according to the VESPA FileName in the tables. A quick way is to Browse, select previous image and then modify the name.
- Choose log file format (.TXT) and browse to the same folder. Name will be automatically generated based on the .his filename.
- A single log file will be generated with the pixel scaling information required to create integrated images (see note at beginning of document about sri.ini file settings).



Works in Progress - Acquire Cine-Mode Images (VMAT only) with Inclinometer

Inclinometer Setup

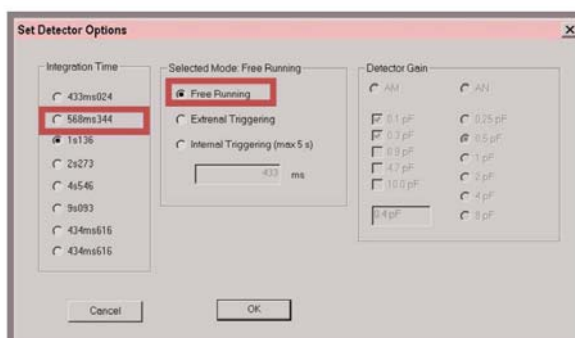
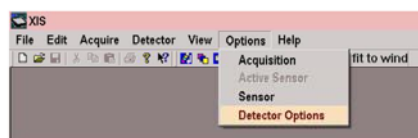
- [Kim instructions here](#)

XIS Setup

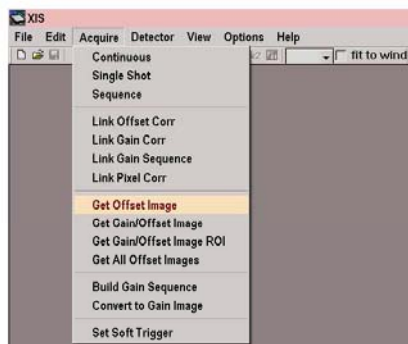
- Cine Mode images for Elekta must be acquired with the Perkin Elmer XIS software **and gantry angles must be independently recorded with an inclinometer.**
- Exit iView. Open the XIS software on the iView computer.
- Select No for Sensor Enumeration message



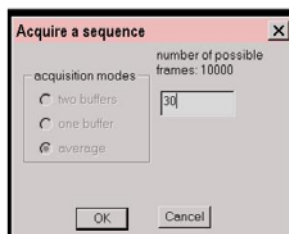
- Options – Detector options.



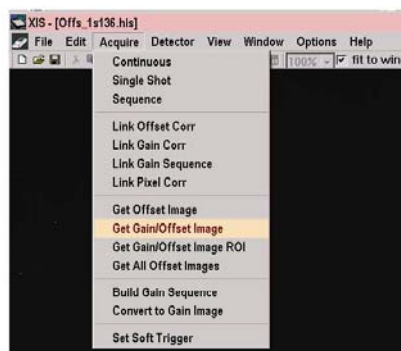
- There is a limitation on the maximum number of frames that can be acquired in the buffer usually around 190 or so. If the VMAT arc is not finished when the maximum frames are acquired then delivery data is lost and the Vespa will not work.
- We recommend to select 568 ms integration time to ensure that all frames are captured i.e. beam off occurs before all frames have been acquired.



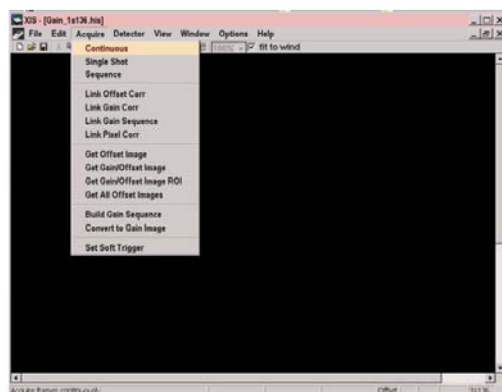
- Offset and Gain images are required for the XIS software. Select Acquire – Get Offset Image
- Enter 30 frames to be acquired for Offset sequence. Select 'ok'. This acquires an image with no radiation beam (dark field).



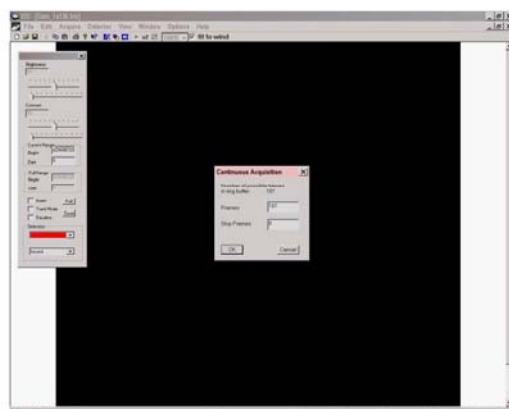
- Save the Offset image (use default file name.his) to a folder you have set up for this experiment. Leave the Offset image open (close any other images) while acquiring the Gain image.
- To acquire the Gain image, first set up a field on the LINAC to cover the entire detector (26 cm x 26 cm field will cover the Elekta EPID) with about 100 MU. Select Acquire – Get Gain/Offset Image.



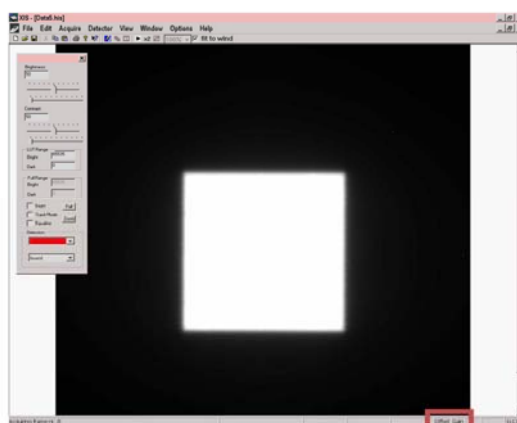
- Enter 30 frames to be acquired for the Gain sequence. Before starting acquisition, turn beam on and wait a couple of seconds to allow dose rate to stabilise at max rate, then start Gain Image acquisition.
- After 30 frames have been acquired (see frame count on bottom left of XIS application), save the Gain file to same location as the Offset file (use default file name.his). [Do not close XIS software or these will be lost]
- The Offset and Gain images are frame average images. The subsequent images acquired for VESPA will be CINE images using the 'Continuous' acquisition mode.



- Unlike iView, XIS images are not triggered by the LINAC beam. Therefore to capture a beam for a given MU, the acquisition must be started immediately prior to beam-on and stopped immediately after beam-off (with a few seconds margin to ensure the entire beam is collected). For each exposure, the minimum number of frames required will depend on the MU / dose rate $[\text{MU}/(\text{DR}/60)] \times \text{frame rate (fps)}$, plus some extra frames before and after beam-on to ensure the entire exposure is collected (e.g. 50 frames should be ok for 100 MU at conventional dose rates). Skip Frames = 0



- Below shows an example of a 10x10 image which is Offset and Gain corrected (indicated at bottom right of application).
- Check that the image is not saturated by viewing the pixel values with a left mouse click. Saturated images will have pixel values of 65535 in the field. If saturated then the frame rate needs to be reduced, but keep in mind this may cause buffer problems for VMAT acquisition because more frames will be required. [Do not Play image use step through frames to view frames.] Closing the acquired image should clear the buffer for the next image.

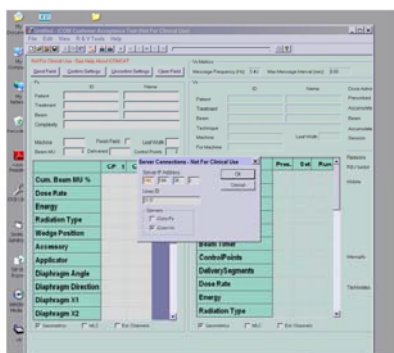


- Use the same 'continuous' acquisition methods described above to acquire VMAT images. CNE images are large files so best to avoid leaving them on your clinical system (100 frames ~ 200 MB).

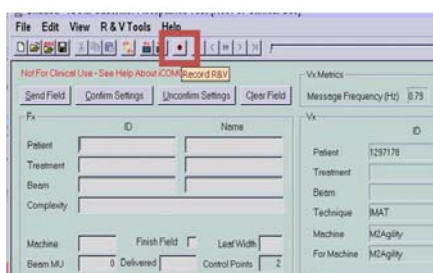
Acquiring gantry angles during VMAT from iCOMCAT

VESPA requires gantry angles for each EPID frame acquired during VMAT delivery. If an inclinometer is not available, an alternative way to acquire gantry angles on Elekta machines is to use Elekta's iCOMCAT software. Log files of LINAC parameters during delivery can be written to file. The gantry angles are written to file with a time stamp. This information can be synchronised to EPID frames off-line by the VESPA analysis after some post-processing. Instructions for using iCOMCAT are below:

- Immediately before VMAT image acquisition, launch the iCOMCAT application (icomcat.exe - usually on the XVI workstation). Do not launch the application until all other images have been acquired.



- Just before you beam on for the VMAT delivery, select the 'Record R&V' button.



- Immediately after VMAT delivery, stop R&V recording.
- Save to somewhere you can access for data transfer (File – Save R&V Recording).
- If a second arc is to be acquired, close iCOMCAT and restart it before repeating the above process for the second arc.
- Submit this file with the VMAT EPID data to VESPA.
- Sankar has Python code to extract time resolved gantry data from this file.

Document History

<u>Date</u>	<u>Version</u>	<u>Author</u>	<u>Changes</u>
09-07-15		PG	100 cm EPID position.
		PG	Reordered sections. Retitled
		PG	Changed his export to single image at a time.
		PG	Added Cine imaging
10-08-15		PG	Added Overflow to Varian Cine Modified Elekta IMRT imaging instructions Added preliminary Elekta VMAT instructions
26/09/17		PG	Minor editing of XIS instructions

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