Improvement of EPID-based techniques for dosimetry and investigation of linac mechanical performance in advanced radiotherapy

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DECLARATION

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7- McCowan PM, McCurdy BM, Greer PB, Rickey DW, Rowshanfarzad P. An investigation of geometry issues for EPID dosimetry during rotational IMRT, *Medical Physics*, 2010; 37(7): 3896 (proceedings of the 56th annual meeting of the Canadian organization of medical physicists and the Canadian college of physicists in medicine, 15-19 June 2010, Ottawa, Canada).


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ABSTRACT

Advances in radiotherapy have increased the complexity of treatment delivery techniques. The complexity of plans, with dynamic variation of field shape, gantry speed and dose rate require highly accurate techniques for quality assurance of the treatment machines and dosimetric verification of the treatment plans.

There has been a growing interest on the application of electronic portal imaging devices (EPIDs) for dosimetry applications and quality assurance testing of linear accelerators (linacs). The ultimate aim of this thesis is to develop methods to ensure more accurate treatment deliveries using EPID-based techniques.

The project is divided into two parts. The first part is based on improvement of the accuracy of EPID dosimetry with Varian systems by either accounting for, or reduction of, the effect of backscattered radiation from the treatment room walls and the EPID support arm.

The effect of backscatter from the treatment room walls was quantified for the first time using a number of measurement setups and comparisons with measurements in the presence of an independent portable wall. The Varian support arm backscatter was accounted for or reduced by three methods: (a) application of an experimentally derived backscatter kernel into an existing EPID dose prediction model, (b) insertion of lead sheets to reduce the non-uniform backscatter, and (c) insertion of a thicker piece of lead over the arm area and considering it as an arm component which could effectively reduce the backscatter effect. Application of the backscatter kernel measured with this lead shielded arm into the model was the most effective method to improve the accuracy of EPID dosimetry predictions.

In the second part of the project, EPID-based measurement methods were used and new algorithms were developed for faster, easier, more robust, more accurate quantitative techniques for characterization of the linac components than existing methods. The results could be used for improvement of EPID dosimetry measurements and/or be included in the linac quality assurance program. The study includes: determination of the mechanical isocentre position with a level of accuracy suitable for stereotactic treatments; determination of the sag in EPID, gantry, jaws and MLC systems during arc deliveries; determination of gantry angle during rotation; and finally, a comprehensive investigation of MLC leaf positioning and dynamic performance in static and arc delivery modes.

The proposed methods have been tested and are applicable for routine quality assurance of the linear accelerators used for advanced treatment techniques with all linacs, independent of their make and model.
CHAPTER 1

Overview
PART I: INTRODUCTION

According to the World Health Organization (WHO), cancer led to the loss of 7.9 million lives in 2007. This number is estimated to increase to 12 million in 2030; which makes cancer one of the major causes of death worldwide accounting for about 13% of all deaths. Another study in 2008 has shown that one-third of Australian men and a quarter of Australian women will develop cancer by the age of 75 (Koh et al., 2008). Between 1990 and 2000, there has been a 36% increase in new cancer cases while the population growth was only 12% (Koh et al., 2008).

Statistics have shown that cancers of lung, stomach, liver, colon and breast are the most common types of this fatal disease worldwide. However, around 30% of cancer deaths are expected to be preventable (WHO Report 2009).

In the United States about 50 to 60 percent of cancer patients receive radiation as a therapeutic or palliative treatment (for relief or prevention of specific symptoms) along with surgery and chemotherapy. The main goals of radiotherapy are improving the quality of the patients’ lives and increasing their life expectancy (Bomford and Kunkler 2003).

1.1 RADIOTHERAPY

Radiation therapy or radiotherapy is a well known and widely used method for the treatment of many types of cancer. It uses high energy ionizing radiation to destroy tumours by depositing the energy of the secondary particles and photons in cancer cells and causing damage to their DNA. This method is especially effective for the treatment of localized tumours where radiation can be targeted at the neoplastic cells (Bomford and Kunkler 2003).

Radiotherapy is divided into two main categories including external beam radiotherapy and brachytherapy. In external beam radiotherapy (also called teletherapy) the source of radiation is outside the patient’s body; while in brachytherapy the radioactive sources are either sent close to the tumour (sometimes implanted into the tumour tissue) or injected as a radio-labelled chemical compound to accumulate in the lesion (targeted therapy) (Suntharalingham et al. 2003).

The first teletherapy machines used high intensity radioactive sources (such as $^{60}$Co, $^{137}$Cs and $^{152}$Eu) or X-ray generator machines which have gradually been replaced by medical linear accelerators (linacs) (Johns and Cunningham 1983, Podgoršak 2005).

Using advanced linacs and modern radiotherapy techniques such as intensity modulated radiation therapy (IMRT), the prescribed dose can be delivered to the tumour while keeping damage to the normal surrounding tissues as low as possible. These machines and treatment techniques will be described in detail later in this chapter.
1.2 LINEAR ACCELERATORS (LINACS)

Throughout the past decades, medical linear accelerators have been commonly used in radiotherapy with the aim of producing ionizing radiation for cancer treatment. Linear accelerators use high frequency electromagnetic waves to accelerate charged particles (electrons) to high energies (mega-electron volts) through a linear waveguide tube (Podgoršak 2005).

In contrast to the research linacs used for high-energy physics studies, medical linacs are compact machines with the ability to irradiate tumours from different angles. In a medical linac, electrons are accelerated along a straight line through a waveguide and can reach kinetic energies from 4 to 25 mega-electron volts (MeV) (Karzmark et al. 1993).

High energy electron beams from a linac can be used for the treatment of tumours near the body surface (superficial treatment), whereas for the treatment of deeply seated tumours, the accelerated electrons must strike a target with high atomic number (usually tungsten). The resulting high energy X-rays can penetrate through the body and interact with cancer cells (Khan 2010). A remarkable advantage of using high energy photon beams is delivery of the maximum dose under the skin surface (Loverock 2007a).

The energies and types of beams produced by different linacs depend on their structure. Some linacs are only capable of production of relatively low energy X-rays (4 MV or 6 MV), whereas more modern types can provide both megavoltage electrons and X-rays- typically photons with energies of 4 and 10 MV, or 6 and 10 MV, or 6 and 18 MV in addition to 6, 9, 12, 16 and 22 MeV electrons (Podgoršak 2005). Therefore, radiation oncologists can use multiple electron and photon energies and dose rates (1 Gy/min to 10 Gy/min) to treat deeply seated tumours (Metcalfe 2007, Podgoršak 2005). The view of a linac in the treatment room is shown in Figure 1.1. The patient is positioned on the treatment couch which can rotate around its vertical axis. The radiation beam is collimated and directed toward the patient from the gantry head (details are given in section 1.3) which can rotate by an entire 360° arc around the patient to enable treatments from different directions. The collimator system can also rotate around its axis to set the beam shape. Two types of imagers are also incorporated into the structure of the linacs (kV and MV imager). They are both used for patient setup and accurate positioning (Figure 1.1).
Figure 1.1. A modern linear accelerator (Varian Trilogy, Varian Medical Systems, Palo Alto, CA): The gantry can rotate around its horizontal axis from -180° to 180° (in the figure the gantry is at zero angle), the collimator can rotate around its axis from -165° to 165°, the couch which supports the patient can rotate from -95° to 95°, The MV imager uses the linac treatment beam for imaging while the kV imager needs a kV beam generator. Both imagers have supporting arms to move to the required positions.

1.3 STRUCTURE OF MEDICAL LINEAR ACCELERATORS

The main components of a linear accelerator are schematically illustrated in Figure 1.2. These components generally include: an electron gun, radiofrequency (RF) system, auxiliary systems, beam transport system, target, beam collimation and monitoring systems. There may be differences in details among various manufacturers due to differences in design or beam energies. In this section, the functions of the main linac components are briefly explained with a focus on the collimation systems which are the subject of a major part of this thesis.
1) The electron gun

The first step to operate a linac is to produce electrons. The electron gun is a small electrostatic accelerator. The electron gun may be a diode consisting of a cathode and an anode, or a triode made of a cathode, an anode and a grid. The cathode is a spiral tungsten filament which can boil-off electrons at a high temperature. Electrons accelerate toward the anode and then escape through a port into an accelerator guide. The filament temperature is about 1100 to 1200 degrees Celsius during the operation (Metcalfe 2007, Podgoršak 2005).

2) The RF power generation system

The electrons produced in the injection system need to be accelerated in an accelerating waveguide. The power generation system includes:

- **RF power source** which could be a magnetron or klystron. Magnetrons are used in lower energy linacs (4 and 6 MV) but klystrons are usually used in high energy dual beam linacs (Loverock 2007a).

- **Pulsed modulator** which generates the required pulses to run the electron gun and the RF generator system

- **Control unit** to enable the timing of the pulsed modulator

- **Accelerating waveguide** which accelerates electrons to very high speeds (near the speed of light). They may be either a travelling or a standing type waveguide.
3) **Auxiliary systems**

There are several systems which are not involved with the electron acceleration but are necessary for operating the machine. They include the water cooling, vacuum pump, air pressure controller, radiation leakage shielding, and gas dielectric system for transmission of the microwaves from the RF power generation system to the waveguide (Podgoršak 2005). It must be noted that the large amount of lead used for the radiation shielding makes the gantry head extremely heavy (Figure 1.3).

![Figure 1.3. An uncovered treatment head rotated by 90°: A thick shell of high-density shielding material (lead) is visible.](image)

The linac head contains several components, which influence the production, shaping, localizing, and monitoring of the clinical photon and electron beams (Podgoršak 2005).

4) **Beam transport system**

After acceleration, the electron beam transport system guides the high energy electrons from the accelerating waveguide to the scattering foil (for electron therapy) or onto the target (for X-ray therapy) (Figure 1.2). The system contains magnetic steering, focusing coils and bending magnets. These magnets bend the electron beam by 90 or 270 degrees (Podgoršak 2005), although in some older linac designs (e.g. Varian 600C/D) no bending magnet was required and the beam was directed straight through the waveguide.

5) **X-ray target**

In order to produce X-rays, the accelerated electron beam is directed toward a heavy metallic target usually made of tungsten or a copper-tungsten alloy (Figure 1.2). About 90 percent of the beam energy is converted into heat and the rest of it results in the production of Bremsstrahlung X-rays. The water cooling system reduces the target temperature to achieve
higher efficiency. The target can be easily moved out of the beam path to change the beam configuration into an electron beam. The thickness of the target depends on the required X-ray energy: thicker targets are used for high energy beams and thinner targets are used for lower beam energies (Podgoršak 2005, Loverock 2007a, Metcalfe 2007). The reason is to provide the required thicknesses for the production of Bremsstrahlung photons such that the incident electrons cannot traverse the targets while the produced photons undergo minimum attenuation (Podgoršak 2006).

6) Flattening filter

The X-ray intensity produced in the target is mainly forward-peaked (Figure 1.4). The flattening filter is used to create an almost flat beam suitable for clinical application by larger attenuation of the centre of the beam (Figure 1.4(b)). Separate flattening filters are required for each of the photon energies. The filters are fixed on a rotary carousel and can be changed. The overall shape of a flattening filter is like a cone and it may be made of copper, tungsten, steel or lead/steel alloy depending on the required energy and the manufacturer’s choice (Metcalfe 2007, Loverock 2007a). In the electron mode the filter is moved out of the way.

![Figure 1.4. The effect of flattening filter on the beam shape: (a) unflattened and (b) flattened photon beam](image)

7) Electron scattering foil or magnetic scanning device

The primary electron pencil beam is very narrow (only a few millimetres wide); therefore, a scattering foil or a magnetic scanning device is used to sufficiently spread the beam for treatments with electron beams. The magnetic field scans the electron beam across the treatment area and produces a sharper beam energy spectrum than the scattering foil. In addition, the photon contamination -which is usually present when the scattering foil is used- does not exist when using the scanning device. However, the scattering foil is more convenient for beam quality control and dosimetry applications (Podgoršak 2005, Metcalfe 2007).
8) **Beam monitoring system**

The presence of a beam monitoring system is critical to ensure the delivery of the prescribed radiation dose. This system usually consists of two independent sealed parallel plate ionization chambers (the dual ionization chambers in Figure 1.2) made of Kapton (which is a polyimide film). They are sufficiently thin to avoid beam perturbation. The function of the ion chamber is to monitor the dose rate, the integrated dose, and the field symmetry. For patient safety, if the first chamber fails, the second chamber will act as the backup (Podgoršak 2005, Metcalfe 2007). In addition, a time interlock inside the system can stop the beam if both ion chambers fail (Metcalfe 2007).

9) **Beam collimation system**

The collimation system is designed to change the size and shape of the treatment beam to meet the individual radiotherapy plan requirements. There are three collimation devices to modify the beam intensity in advanced medical linear accelerators: (a) the fixed primary collimator, (b) the secondary adjustable beam defining collimators (jaws), and (c) the tertiary or multi-leaf collimators (MLC). The secondary and tertiary collimators will be explained in more detail based on Varian linacs, since a part of this thesis focuses on their mechanical performance.

**a) Primary or fixed collimator:** The primary or fixed collimator is placed between the target and the flattening filter (Figure 1.2). It usually has the shape of an open cone made of tungsten which gives the beam a circular cross section. The main role of the primary collimator is to regulate the scattered photons to the forward direction and minimize gantry head radiation leakage. The primary collimator dimensions are usually adjusted to provide a circular beam diameter of 50 cm at source-to-surface distance (SSD) of 100 cm if there are no other collimation systems in the beam path (Podgoršak 2005, Metcalfe 2007). The source is either the target or the scattering foil.

**b) Secondary collimators:** The secondary collimators in the linac head consist of two sets of adjustable flat-faced opposing blocks (X₁, X₂, Y₁ and Y₂ in Figure 1.5) to produce square or rectangular fields. These heavy blocks (also known as jaws) are usually made from lead or tungsten alloys and their thickness is about 8 cm and each weigh approximately 30 kg (Rowshanfarzad et al. 2012a). The two jaw pairs have different mechanisms for movement: the X-jaws slide by two parallel drive scrolls, while the Y-jaws use rack-and-track systems for movement (Figure 1.6) (Rowshanfarzad et al. 2012a). Each jaw is separately controlled by the linac controller system and can therefore move independently to define asymmetric fields (Khan 2003). The maximum jaw-defined field size is 40×40 cm² at SSD=100 cm.
Figure 1.5. Schematic illustration of the secondary collimators in a linac at zero collimator angle. Directions are specified as L: left, R: right, G: gun, and T: target sides. The figure shows the view from below the linac head (looking upward).

The secondary collimators can rotate around the beam central axis (Loverock 2007a, Metcalfe 2007). At zero collimator angle, the X-jaws and Y-jaws move in the cross-plane and in-plane directions, respectively. These directions are indicated in Figure 1.6 by the X (cross-plane) and Y (in-plane) arrows on the MV imager. The track for the Y jaws is slightly curved and has a small angle with the horizontal level at zero gantry angle (Rowshanfarzad et al. 2012a). The secondary collimators can block 99.6% of the 6 MV beams (Loverock 2007a, Metcalfe 2007).

(c) Multi-leaf collimators: The invention of multi-leaf collimators (MLCs) in the 1960s (Takahashi 1965) led to revolutionary changes in radiotherapy treatments. The MLCs are a series of interleaved collimators in the gantry head and are included in the structure of almost every modern linac. They have replaced the heavy lead blocks which were traditionally used for field shaping and protection of the critical organs (Podgoršak 2005, Loverock 2007a). Using the MLCs allows the production of irregularly shaped radiation fields to conform to the shape of the target and thus minimize the irradiation of healthy tissues. The shapes are provided by an array of narrow collimator leaf pairs, each controlled with its own miniature motor (Metcalfe 2007). Computer-controlled MLCs enable rapid changes in field shape when multiple fields are applied.
Most Varian linacs have either 80 or 120 leaves (40 or 60 leaf pairs) made of tungsten alloy with sufficient thickness to attenuate the X-rays to less than 2%. Some manufacturers have used curved edge leaves to reduce the penumbra, and have provided a tongue and groove joint between the leaf sides to minimize the leakage (Loverock 2007a). This has caused about 3% inter-leaf transmission as reported in the literature (Khan 2010, Boyer et al. 2001). For patient treatments with MLC-defined fields, the jaws can be driven to the field margins to minimize the leaf transmission and interleaf leakage. In Varian linacs, the jaws are positioned above the MLC leaves (Figure 1.6).

The MLC systems used for the experiments in this thesis were Millennium™ 120-leaf MLCs (Varian Medical Systems, Palo Alto, CA). They include two leaf banks (A and B) each with 60 machined tungsten alloy round-ended leaves mounted on a carriage (Figures 1.6 and 1.7). The 40 central leaves in each bank are 0.5 cm thick (at SSD=100 cm) and are called the inner leaves. The peripheral 20 leaves are 1.0 cm thick and are known as the outer leaves. Each leaf is equipped with a motor which is driven by the MLC controller and drives the leaf along the carriage. The maximum leaf speed is 30 mm/s and the leaf extension range is 14.5 cm (at SSD=100 cm). Further leaf motions require the movement of the carriage.
In Varian linacs, when the system is powered on, the leaves are calibrated using an infrared beam. The infrared source is located in the gantry head and sends the beam perpendicular to the leaves. Each leaf can be calibrated since the distance between the leaf carriage and the infrared source is known (Vieira et al. 2006).

**10) Optical field indicator**

An optical field indicator defines the position of the radiation field, field edges and the field centre to help the radiation therapists find which part of the patient’s body will be exposed. The light from a high intensity bulb is reflected from a thin mylar mirror positioned above the jaws. The light field is congruent with the radiation field (Khan 2003, Podgoršak 2005, Metcalfe 2007). The graticule at the centre of the optical field indicates the radiation centre. In addition, an optical distance indicator (ODI) allows the operator to see the distance between the X-ray target and a given surface.

**1.4 LINAC ISOCENTRE**

The idealized intersection point of the gantry axis of rotation with that of the collimator and treatment table is known as the mechanical isocentre (Lutz et al. 1988, Hartmann et al. 1995, González et al. 2004). In practice, due to the heavy weight and mechanical imperfections of the system, the isocentre is not a single point and its location changes with the rotation of the gantry, collimator or couch. This causes uncertainties in the determination of isocentre position and therefore, the isocentre is usually defined within a sphere of 1-2 mm diameter (Khan 2010). It is necessary to have tight isocentre tolerances to deliver geometrically accurate treatments.
(Khan 2003) and linac quality assurance programs include routine checks of the isocentre position (Khan 2010), since it is the primary reference location of external beam radiotherapy (Tsai et al. 1996). Sophisticated techniques are required to accurately characterize the linac isocenter. (González et al. 2004)

1.5 ELECTRONIC PORTAL IMAGING DEVICES (MV IMAGERS)

Electronic portal imaging devices (EPIDs) are a supplementary part of modern linacs and have been used as the main measurement tool in all parts of this project. Therefore, the structure and applications of different types of EPIDs are explained in this section.

Successful radiotherapy treatments strongly rely on accurate patient positioning, since high intensities of megavoltage energy beams are used in modern radiotherapy treatments. X-ray films were conventionally used to check the anatomical patient setup but they required processing which was a time-consuming procedure; therefore, the information was only available after the treatment (Munro and Bouius 1998, Herman et al. 2001). Nowadays, every modern linac is equipped with a two-dimensional megavoltage imaging device attached to the gantry by means of a supporting arm (Figure 1.1). The imager can provide real-time digital data for verification of the patient setup before and during the course of treatment. The images acquired by these devices are easy to process, enhance and archive. These imagers are known as electronic portal imaging devices (Munro 1999, Herman et al. 2001, Antonuk 2002, Loverock 2007b).

Different types of EPIDs have been developed but only three of them have become commercially available: camera based EPIDs, scanning liquid filled ionization chamber (SLIC) EPIDs, and amorphous silicon (a-Si) EPIDs (Antonuk 2002, van Elmpt et al. 2008, Loverock 2007b).

1.5.1 CAMERA-BASED EPIDS

Camera based EPIDs, also known as video-based (VEPID) or fluorescent screen-based EPIDs, were first described by Sven Benner and his colleagues at the University of Göteborg, Sweden (Benner et al. 1962) and became commercialized in the late 1980s (Antonuk 2002). The function of a camera-based EPID is schematically illustrated in Figure 1.8. The detector has a metal sheet (typically ~1-1.5 mm thick copper, steel or brass layer) attached to a thick fluorescent screen (Gd₂O₂S:Tb). High energy X-rays interact with the metal sheet after passing through the patient and produce high energy electrons. The electrons interact with the phosphor screen and produce a large number of optical photons (about ten thousand per interacting X-ray). The light fluence pattern is captured by a camera after reflection from a 45º mirror. The main duty of the mirror is to keep the camera away from harmful radiation. The
camera is connected to a computer system which converts the video signal into frames with digital format (Khan 2010). The recorded signal can be displayed and stored for further analysis (Boyer et al. 1992, Greene and Williams 1997, Munro 1999).

The metal sheet also absorbs low energy scattered electrons and photons from the gantry head, patient and patient support system. The scattered electrons degrade the image quality and decrease its contrast. The optimum thickness for the metallic plate is 1 g cm$^{-2}$. A phosphor screen with a thickness of about 400 mg cm$^{-2}$ can absorb most of the electrons produced in the metallic sheet (Antonuk 2002, Loverock 2007b). The main disadvantage of this system is its poor image quality, since only 0.01% to 0.1% of the emitted fluorescent light can be captured by the camera, and there is some light scattering inside the phosphor screen. In addition, the light scattering and reflection from the mirror reduce the image contrast (Wong and Yan 1994, Herman et al. 2001, Antonuk 2002, Jaffray 2005).

![Figure 1.8. Schematic illustration of the function of a camera-based EPID mounted on a linac](image)

### 1.5.2 SCANNING LIQUID FILLED IONIZATION CHAMBER EPIDs (SLIC EPIDs)

This device was developed at the Netherlands Cancer Institute (NKI) in Amsterdam by Meertens and van Herk in 1985 and has been commercialized since 1990 (Meertens et al. 1985, van Herk and Meertens 1988, Meertens et al. 1990). The size of the detector is about 32.5x32.5 cm$^2$ and contains 256x256 separate ionization chambers which provide a pixel pitch of about 1.27 mm. Each chamber has two electrode plates 0.8 mm apart and is filled with an organic fluid such as 2,2,4-trimethylpentane as the ionisation medium. A 1 mm thick plastoferrite layer covers the ionization chamber and acts as the converter of the incident high energy X-rays into electrons. The image is acquired by applying a sequential high voltage (250 V to 500 V) to the electrodes and measuring the signal for each. The time required for clinical image acquisition is between 0.6 s to 2.0 s (Mubata 2007, Antonuk 2002).

The compact size and light weight of matrix ionization chamber EPIDs were the main advantages of this system over the camera-based EPIDs, but they have a major limitation: the
measured signal depends on the rate of formation/recombination of the ion pairs in the ionizing fluid which is proportional to dose rate (Mohammadi and Bezak 2006). When the beam is turned on, the concentration of ion pairs increases over a period of time (~0.3 s) and after that no increase in signal occurs with further irradiation (equilibrium). The other restriction for the matrix ion chamber is that it requires higher doses to generate images compared with the other types of EPIDs (van Herk 1991, Parsaei et al. 1998) and needs a time gap between image acquisitions to allow for complete dissipation of charges (Curtin-Savard and Podgorsak 1997).

1.5.3 ACTIVE MATRIX FLAT PANEL IMAGER EPIDs

Active matrix flat panel imagers (AMFPI) are the latest and best known invention in the field of megavoltage imaging. This type of EPID was invented at the University of Michigan in 1987, developed in the 90s by Antonuk, Street and their colleagues and finally became commercially available in 2000. Based on the use of amorphous silicon in the detector assembly, active matrix flat panel imaging devices are also known as amorphous silicon (a-Si) EPIDs.

The quality of images acquired by these EPIDs is superior to camera-based and SLIC EPIDs. The efficiency of the AMFPI systems is much higher than camera-based EPIDs mainly due to their compact structure which places the detector layer close to the scintillator screen, and also the type of the detectors (Antonuk 2002). Therefore, AMFPIs have rapidly replaced camera-based and SLIC EPIDs. The main components of an active matrix flat panel imager include: an X-ray converter, a fluorescent screen, a photodiode array, an electronic data acquisition system and a computer system. Considering that a-Si EPIDs have been used in all parts of this study, the structure of this type of EPID is discussed in more detail.

(a) X-ray converter

The X-ray converter is a thin metallic plate which interacts with the incident high energy beam and produces Compton electrons (Antonuk et al. 1990a). The metallic converter layer is usually a ~1 mm thick copper or steel sheet which not only produces high energy electrons, but also provides a dose buildup medium* for the incident beam. In addition, it absorbs the low energy scattered radiation from the patient and patient support tools, and reduces their effect on the EPID image (Antonuk 2002).

(b) Fluorescent screen (phosphor)

The high energy secondary electrons are absorbed in a phosphor scintillator screen and produce optical photons. The phosphor screen may be made of terbium doped gadolinium

* When a broad photon beam interacts with a medium, the dose initially builds up to a maximum value where electronic equilibrium is attained, and then starts to decrease with increasing depth due to attenuation.
oxysulphide (Gd$_2$O$_2$S:Tb) or Thallium doped caesium iodide (CsI:Tl) (Mubata 2007). The phosphor thickness is ~134 mg cm$^{-2}$ in both Varian aS500/1000 and Elekta iViewGT EPIDs (Antonuk et al. 1990b, Munro and Bouius 1998, Antonuk 2002, Siebers et al. 2004).

(c) Photodiode array

A large two dimensional pixelated photodiode array seated in amorphous silicon detects ~90% of the optical photons (Antonuk et al. 1990a). The array consists of ~1 mm thick glass substrate that holds a number of electronic circuits. Each pixel has a capacitive element which is a highly sensitive photodiode, coupled to a conductive element that is a thin film transistor (TFT) (Figure 1.9). The photodiode senses and stores the charges produced as a result of interactions with light photons (electron-hole pairs). The thin film transistor which is usually made of hydrogenated amorphous silicon (a-Si:H), controls the signal readout and is known as the pixel switch (El-Mohri et al. 1999, Yorkston 2007, Antonuk 2002). It is worth mentioning that the a-Si:H TFTs have low leakage current which makes them popular for many applications (Sultana et al. 2008). The pixel switches in a row are controlled by a control line. During irradiation, the pixel switches remain non-conducting and during the readout the pixel switches are turned to conducting status. The array readout is controlled by applying reverse bias voltages to change the voltage of the control lines (El-Mohri et al. 1999, Antonuk 2002). It must be noted that the number of pixels may be different among the photodiode array designs. For instance, Varian aS500 and aS1000 EPIDs have arrays of 384×512 and 768×1024 pixels, respectively.

The production procedure for the electronic circuits involves many stages including plasma enhanced chemical vapour deposition (PECVD), etching and passivation which usually requires 5 to 10 sets of photolithographic masks on a thin (~1 mm) glass substrate (Antonuk 2002). The detector system is packed in an opaque housing to avoid the ambient light (Pang et al. 2001).

![Figure 1.9](image_url)

**Figure 1.9.** Schematic illustration of an a-Si megavoltage imaging system: (a) the detector array elements and electronics (adapted from Cowen et al. 2008); (b) details of the pixel structure and the peripheral electronics (adapted from Antonuk 2002). IC stands for Integrated Circuit.
It must be noted that the a-Si photodiode arrays and TFTs are highly resistant to radiation damage, even for radiation doses above $10^4$ Gy per year that may be received by a portal imager in a radiotherapy unit (Antonuk et al. 1990a, El-Mohri et al. 1999).

(d) **Electronic data acquisition and processing systems**

The electronic data acquisition system is connected to the array electronics. Data lines sample the signals stored in the photodiodes. The system receives the signals from each pixel (Antonuk 2002). The final images are processed by a computer system connected to the electronic acquisition system. The computer system minimizes the pixel-to-pixel variations and applies a gain correction for each pixel (Antonuk et al. 1996c). The acquisition system can be operated in either the integrated mode, by taking a single image averaged over several frames following an irradiation, or the continuous (cine) mode which refers to acquisition of a consecutive series of frames during radiation delivery (El-Mohri et al. 1999). The Varian EPID acquisition system could also be operated in “low dose” or “high quality” imaging modes where images are acquired by averaging 2 and 4 frames with the linac operating at 100 and 600 MU/min, respectively.

(e) **The EPID support arm**

EPIDs are attached to the linac gantry by means of a robotic support arm. The design, components, positioning and degrees of freedom for the support arm movements are different among manufacturers. Since a major part of this study is based on the support arms manufactured by Varian Medical Systems, the support arm designs of Varian EPIDs are described here.

- **The retractable arm (R-arm)** was the first design for the EPID support arm by Varian Medical Systems. This design consisted of two metallic bars that were attached by an axle which also held another metallic part that supported the EPID and moved it to the intended position. The photograph of a Varian R-arm is shown in Figure 1.10.
Figure 1.10. An R-arm holding a Varian aS500 EPID: (a) lateral view (b) view from beneath the imager

- **Exact arm (E-arm)** replaced the R-arm design many years ago. In E-arms, the two bars are integrated into one thicker component that holds the supporting part. In this design, the EPID is more firmly fixed with improved stability in positioning (Grattan and McGarry 2010). The photograph of a Varian E-arm is shown in Figure 1.11.

Figure 1.11. An E-arm holding a Varian aS500 EPID: (a) lateral view (b) view from beneath the imager

The E-arm is mainly constructed of steel sheets of 1-3 mm thicknesses and can move the imager vertically (variable target to imager distance) and horizontally (both in the in-plane [longitudinal] and cross-plane [lateral] directions). The E-arm and R-arm designs have similar degrees of freedom for their movements. The cross-plane motions are made by an electric motor using gearwheels and a metallic saw-tooth rail. There are more metallic components in the E-arm structure such as two steel bars fixed to a frame (Figure 1.11(b)). The rear housing of the EPID cassette is not uniform and has indentations to incorporate these components and the
imager cabling. Motions in the in-plane direction are more complicated since they are
accomplished through movements of the three-piece body of the robotic arm using three
junctions (Figure 1.11). The non-uniform structure of the arm and the presence of several
metallic parts in its structure have an impact on dosimetric applications of the EPID which is
explained in the next section.

1.6 OTHER APPLICATIONS OF EPIDS

As mentioned in the previous sections, EPIDs were originally designed to verify patient
positioning for radiotherapy treatments. However, some of their properties such as the ability
to provide real-time high resolution two-dimensional digital data (van Elempt et al. 2008), and
the fact that at present EPIDs are incorporated into the structure of all modern linacs have
attracted lots of attention toward using them for other purposes. The two most popular
applications of the EPIDs including linac quality assurance (QA) and dosimetry measurements
are briefly explained here.

1- Quality assurance of linacs

Amorphous silicon EPIDs are interesting tools for the quality assurance (QA) of linacs
considering their speed of imaging, linear response to dose and ability to provide images with
acceptable contrast and resolution, in addition to their easy image acquisition and storage
compared with conventional film-based methods (Chang et al. 2004, Warkentin et al. 2003,
Hansen et al. 2005). EPIDs have been confirmed as suitable devices for checking the beam
flatness and symmetry (Kirby and Williams 1993, Liu et al. 2002, Budgell et al. 2007). Amorphous silicon EPIDs have also been used for QA of the MLC leaf positions to verify the
accuracy, reproducibility and smoothness of leaf motions by either checking the leaf trajectories
(Samant et al. 2002, Baker et al. 2005, Sonke et al. 2004), or analysis of the measured dose or
fluence profiles (Chang et al. 2004, Chang et al. 2000, Warkentin et al., 2003, Mamalui-Hunter et

There have been reports on the application of different types of EPIDs to verify the linac
isocentre position, coincidence of the light field and radiation field (Ma et al. 1998, Prisciandaro
et al. 2003), collimator and couch axis of rotation (Ma et al. 1998), coincidence of MLC field
centre with its axis of rotation, MLC transmission, penumbra width measurements (Curtin-
Savard and Podgoršak 1997), leaf gap measurement, beam output (Vieira et al. 2006, Budgell et
al. 2007), field size or wedge factors (Budgell et al. 2007, Greer and Barnes 2007) and
assessment of the asymmetric jaw alignment (Clews and Greer 2009).

2- Dosimetry measurements
Amorphous silicon EPIDs are suitable candidates for dosimetry applications due to their stability (in both short and long term) (Louwe et al. 2004, Greer and Popescu 2003, Winkler et al. 2005, Nijsten et al. 2007, Vial et al. 2008) and linear response to dose. Two main dosimetry applications have been proposed for EPIDs: (a) pre-treatment verification of IMRT plans, and (b) exit (in-vivo) dosimetry.

The conventional method for dosimetric verification of the treatment plans is comparison of in-phantom dose measurements by ionization chambers or films with the dose predicted by the treatment planning software (Danciu et al. 2001, Alber et al. 2008). EPIDs do not need the time-consuming setup and processing procedure, have high resolution two-dimensional arrays and provide real-time images in digital format; therefore, they are much easier to use than films for verification of the treatment plans. Amorphous silicon EPIDs have proved to be effective tools for this application using calibration procedures to convert the EPID images into dose (Chang and Ling 2003, Talamonti et al. 2006, Warkentin et al. 2003, Wendling et al. 2006, McDermott et al. 2006, Hansen et al. 1996, Van Esch 2004).

Exit dose has conventionally been measured by fixing diodes or thermoluminescent dosimeters (TLDs) on the patient body, but correct placement of the detectors and data analysis is difficult and time-consuming. Moreover, doses could only be specified in few points by TLDs (Parsaei et al. 1998). EPIDs are much easier to use due to their positional accuracy and ability to provide an array of digital data (Chen et al. 2006).

Several research groups have studied the a-Si EPIDs in exit dosimetry through measurement and/or modelling. The major drawback to the application of EPIDs for dosimetry is that their response is not water-equivalent (El-Mohri and Antonuk 1999) due to the presence of the phosphor scintillator screen which has a high atomic number and makes the EPID highly sensitive to low energy radiation (Jaffray et al. 1994, Manfredotti et al. 1992). Accurate exit dosimetry measurements with a-Si EPIDs requires corrections for a number of effects such as off-axis differential energy response and scattered radiation from the patient and within the EPID. However, acceptable dosimetry results have been achieved after application of the required corrections and calibration of the EPID images to dose (Chen et al. 2006, Nijsten et al. 2007, Grein et al. 2002, Winkler et al. 2007, Piermattei et al. 2006). Successful modelling procedures have also been reported by many research groups (McCurdy and Pistorius 2000, McCurdy et al. 2001, Siebers et al. 2004, Kairn et al. 2008, Lin et al. 2009).

1.7 MODERN TECHNIQUES IN RADIOTHERAPY

There have been numerous advances in radiotherapy techniques over time, which have led to considerable improvements in the treatment outcomes and patient safety. Some of the specialized radiotherapy techniques including intensity modulated radiation therapy (IMRT),
intensity modulated arc therapy (IMAT) and stereotactic radiosurgery/radiotherapy which are relevant to this thesis are briefly explained below.

1) Intensity modulated radiation therapy (IMRT)

In conventional radiotherapy, most treatments were delivered with open uniform beams. The major disadvantage of this method was the exposure of healthy tissues to high radiation doses (Khan 2010). Two-dimensional radiographs were used for the determination of the treatment fields. Following the introduction of three-dimensional imaging (such as CT scanning), alongside developments in planning algorithms and increased computing power, and the invention of multi-leaf collimators for beam shaping, the quality of treatment planning was improved significantly (Caldwell and Mah 2005). Conformal radiation therapy was introduced which was based on the delivery of beams in a shape that conformed to the target volume. It gradually replaced conventional radiotherapy, with the aim of delivering the maximum dose to the tumour and the minimum dose to normal tissues (Khan 2010). Other three-dimensional imaging techniques (such as MRI, PET, SPECT and MRS) have also been used for treatment planning purposes (Caldwell and Mah 2005).

The treatment planning method in conformal radiation therapy was called forward planning and included the specification of the number of beams, and their shapes, directions, wedges and beam intensities (for static fields). Dose distributions were calculated by computers, and the shape of the radiation beams were formed by multi-leaf collimators to conform to the tumour shape as closely as possible. It must be noted that dose distribution was almost uniform in the target (Teh et al. 1999). In some cases wedges or tissue compensators were applied to modify the beam intensity and compensate for irregular body shapes. The wedges and tissue compensators were the first and simplest tools used for intensity modulation.

In modern radiotherapy, dynamic multi-leaf collimators (DMLCs) are used for intensity modulation through much more complicated processes. IMRT techniques are used to deliver radiotherapy beams with optimized complex dose plans produced by inverse treatment planning. The MLC systems provide the intensity modulation of the radiation beams during treatments (Khan 2010). Using the IMRT method, every radiation beam is divided into a number of beamlets with different intensities (Palma et al. 2010). Schematic illustration of an IMRT treatment is shown in Figure 1.12.
For inverse treatment planning, only the beam direction, beam energy and dose distribution in the tumour and surrounding tissues are given to the planning system. The planning algorithm suggests beam shapes and intensities to meet the dose limits. The optimal treatment field is determined using optimization techniques to optimize beam parameters such as beamlet weights (Teh et al. 1999, Amols et al. 2003). The reason for using inverse treatment planning is that in forward planning -as is used in conformal radiation therapy- there is no method to determine if there might be a better plan, and that the planner can not investigate all other options. In addition, due to the large number of variables in IMRT plans, a planner can not possibly suggest all the beamlet weights. The other important advantage of inverse planning is the ability for optimization which can make a big difference to the planning process (Webb 2007). However, delivery of the complicated IMRT plans requires longer treatment times such that the delivery of a fraction may take 15 to 30 minutes. (Palma et al. 2010).

During the IMRT treatments, radiation beams are emitted from different directions to deliver the prescribed dose to the target volume while avoiding the critical normal structures. Even different radiation doses can be delivered to different parts of the tumor volume. The MLC system enables the delivery of the non-uniform fluences according to the treatment plan (Khan 2010). Several techniques have been developed for IMRT beam delivery (Webb 2007). IMRT plans are commonly delivered using one of the following techniques:

- **Static Delivery (Step and shoot):** The first field is shaped by the MLC and after its delivery, the radiation is stopped and the second field is formed, delivered, and so on. Each field is composed of a set of sub-fields that have uniform intensities (Khan 2010, Van Esch et al. 2002, Aspradakis et al. 2005)
- **Dynamic Delivery (Sliding window)**: During the beam-on time, the gantry does not move but the window formed by the MLC leaf pairs transverses the field. The leaves move in the same direction each with a specified speed; therefore, the beam intensity is variable across the field (Kahn 2010, Van Esch et al. 2002, MacKenzie and Robinson 2002).

2) **Intensity-modulated arc therapy (IMAT)**

The main advantage of delivering radiotherapy beams in arcs is that the dose to the normal tissues around the target is spread out. This prevents the delivery of high doses to critical organs (Korreman et al. 2009, Palma et al. 2010). In Intensity modulated arc therapy (IMAT), the radiation beam is modulated during the treatment by changing the MLC leaves while the gantry rotates around the patient, and the beam is not turned off (Khan 2010, Yu 1995). Therefore, the treatment times are shorter (Palma et al. 2010). The difference between this method and IMRT is that in IMAT the MLCs move during gantry rotation and change the shape of the fields. Briefly, IMAT can be considered as a combination of dynamic MLC modulation and gantry rotations (Yu 1995, Yu et al. 1995). IMAT treatments usually require three to five arcs (Khan 2010, Yu 1995).

Recent advances to IMAT techniques have resulted in faster delivery techniques with smaller number of monitor units (MUs) such that the patient is treated by a single arc (sometimes two) with continuous changes in the field shape, dose rate and gantry speed (Korreman et al. 2009, Mans et al. 2010, Bedford and Warrington 2009, Matuszac et al. 2008, Rangaraj et al. 2010, Palma et al. 2010). Shorter treatment times lead to treating more patients per day, providing more patient comfort and smaller possibility of intra-fraction movements (Palma et al. 2010, Verbakel et al. 2009, Vanetti et al. 2009). In addition, tumour cells do not get a chance to repair their damaged DNA with short treatment times (Palma et al. 2010). The optimization technique used in this method (which is generally referred to as volumetric modulation arc therapy or VMAT) is named differently by the manufacturers: Elekta Inc. (Stockholm, Sweden) calls it VMAT™ while Varian Varian Medical Systems has named it as RapidArc™. Schematic illustration of a volumetric modulated arc therapy treatment is shown in Figure 1.13.

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* Linacs are calibrated to output a particular dose under particular conditions. For example, 1 MU could be the radiation which delivers 1 cGy using a 10×10 cm² field at the depth of maximum dose with SSD=100 cm.

3) Stereotactic radiosurgery

Stereotactic radiosurgery is a highly precise technique for the delivery of high doses of ionizing radiation (usually from a linear accelerator) to a localized small target volume of typically 1-3 cm diameter (Lutz et al. 1988, Podgorsak et al. 1989, Podgorsak et al. 1990, Ramaseshan and Heydarian. 2003, Verellen et al. 1999, Serago et al. 1992). This method was invented by Lars Leksell - a Swedish neurosurgeon, as a non-invasive method to obliterate the brain tumours located in positions that are difficult to access for surgery (Leksell 1951). If the whole stereotactic dose is delivered in one session, the treatment is called stereotactic radiosurgery (SRS) and if the stereotactic dose is delivered in multi-fraction sessions it is known as stereotactic radiotherapy (SRT).

The main advantage of stereotactic radiosurgery is minimum damage to the surrounding critical organs (usually brain tissues) by application of a collimated and confined beam directed toward a small target lesion. This treatment method has been explained in detail in chapter 6.
Part II- DESCRIPTION OF THE RESEARCH PROBLEMS AND LITERATURE REVIEW

This project is divided into two parts. The first part is based on the improvement of the accuracy of EPID dosimetry with Varian aS500/aS1000 systems by accounting for, or correction of, the effect of backscattered radiation.

In the second part of the project, EPID-based measurement methods have been used and new algorithms have been developed for faster, easier, more robust, quantitative techniques for characterization of the components of the linacs used in modern radiotherapy treatments. The results could be used for improvement of EPID dosimetry measurements and/or be included in the linac quality assurance program. The final goal however is to ensure more accurate treatment deliveries.

The research problems are explained in this part with a review of the relevant studies previously reported in the literature to present the existing level of knowledge in each part of the study.

1.8 EFFECT OF THE BACKSCATTERED RADIATION ON DOSIMETRY WITH VARIAN EPIDS

As already mentioned in section 1.6, there has been a growing interest on the application of EPIDs for dosimetry purposes. One of the problems associated with dosimetry using a-Si EPIDs is their sensitivity to low energy scattered radiation. This effect is caused by the presence of the high atomic number phosphor scintillator screen in the structure of a-Si EPIDs which makes them sensitive to low energy radiation due to the increased probability of the photoelectric effect (McCurdy et al. 2001, Siebers et al. 2004). For Varian EPIDs studied in this project, there are two main sources of low energy backscattered radiation which could affect the accuracy of dosimetry measurements:

1- The backscattered radiation from the EPID supporting arm
2- The backscattered radiation from the treatment room walls, ceiling and floor

More details are given in the following sections.

1.8.1 EFFECT OF THE SUPPORT ARM BACKSCATTER

Details of the support arm structure are given in section 1.5.3(e). The presence of several metallic parts in the structure of the support arm produces a non-uniform backscattered radiation to the detector array and introduces inaccuracies to dosimetry measurements (Ko et al. 2004, Moore and Siebers 2005, Wang et al. 2009, Greer et al. 2009, Gustafsson et al. 2009).
Different approaches have been proposed by researchers to model aspects of the backscatter component. One approach was to model the field size response of the EPID at central axis which is affected by backscatter, EPID scatter and optical glare. Monte Carlo methods were used to provide generic EPID dose kernels (Siebers et al. 2004) or imager-specific dose kernels (Wang et al. 2009). It has been suggested that a uniform water slab with sufficient thickness (1-1.6 cm depending on the imager (Wang et al. 2009) or 0.98 gr/cm² with a 21 mm thick air gap (Siebers et al. 2004)) could be used to include the effect of backscatter from the rear housing and the metallic parts beneath the EPID at the central axis, but the non-uniform distribution of backscatter was not modelled. Another study focused on the development of an energy fluence convolution model for a-Si EPID dose prediction and used an off-arm (backscatter-free) pixel sensitivity matrix to remove the effect of backscatter artifacts introduced by the large backscatter component of the flood-field correction image (Greer et al. 2009). Images then contained only a field-specific backscatter artifact which is generally smaller. A field size specific backscatter correction algorithm was introduced in a study based on backscatter measurements for different field sizes. Two correction methods including a matrix-based and an equation-based strategy were proposed. The results of both correction methods showed improvement for most of the 49 tested IMRT fields although there were a few exceptions (Berry et al. 2010).

The simplest method to minimize the effect of arm backscatter was used by researchers in Virginia Commonwealth University (Ko et al. 2004, Moore and Siebers 2005). They used Monte Carlo simulations to find the optimal backscattering material for an R-arm EPID using different thicknesses of water, copper and lead and found that placing 5 mm of lead between the arm and the detector could considerably reduce the non-uniformity in the backscatter signal for 6 and 18 MV beams (Ko et al. 2004). Later, in another study the imager was experimentally isolated from the mechanical support arm using the suggested 5 mm of lead behind the imager cassette and a maximum backscattered signal of less than 1% from the arm was reported for both energies (Moore and Siebers 2005). The insertion of lead layers seems an easy and feasible method to minimize the non-uniform backscatter from the arm; but no report has been released on the applicability of this method to the E-type robotic arms which has an entirely different design and construction. The backscatter from the E-Arm is known to be considerably greater than the backscatter from the R-arm with a maximum backscatter component of approximately 6.5% (Greer et al. 2009) compared to 4.5% for the R-arm (Lee et al. 2008). Moreover, the added weight to the arm of a large thickness of added lead is of concern; therefore it is important to determine the minimum backscatter thickness that still gives acceptable dosimetric and imaging performance, and to understand the effect on imager sag, which is particularly important for arc therapy verification (Mans et al. 2010, Grattan and McGarry 2010).
It must be noted that after the publication of the results of this project, a correction matrix for arm backscatter was developed by Cufflin et al. (2010) using Monte Carlo simulations. More recently, a model for dosimetry with a prototype backscatter-shielded Varian EPID has been reported (King et al. 2012). The model was optimized for dose reconstruction at different depths of water using images of the backscatter-shielded EPID.

1.8.2 EFFECT OF THE BACKSCATTERED RADIATION FROM THE BUNKER STRUCTURE

The presence of an effect on EPID dosimetry from the surrounding walls has not been quantified yet, and to date the author has not found any reports in the literature regarding this effect. It is therefore worthwhile to perform a systematic and quantitative study on the subject to ensure the accuracy of EPID dosimetry measurements. A typical treatment room structure is shown in Figure 1.14 which will later be discussed in Chapter 5.

![Figure 1.14. Schematic illustration of a linac delivering arc treatment while the EPID acquires images. The distance between the EPID detector and the bunker walls, ceiling and floor changes during rotation. The circular area on the floor shows the position of the treatment couch rotation system. The distances specified on the left are measured between the side walls/roof/ceiling and the linac isocenter.](image)

1.9 CHARACTERIZATION OF LINAC MECHANICAL PERFORMANCE

As pointed out in section 1.3, the linac includes several different components. The mechanical behaviours of these components during treatment delivery need to be thoroughly investigated and particularly characterized during gantry rotation, since the mechanical imperfections of the delivery system could affect the accuracy of patient treatments. In the second part of this project, several aspects of the linac mechanical performance have been studied, including:

1- The linac mechanical isocentre position for radiosurgery/radiotherapy treatments
2- The EPID and gantry sag during rotation which could affect the accuracy of EPID dosimetry and EPID-based QA measurements
3- The secondary and tertiary collimation systems sag as a function of gantry angle
4- Determination of the gantry angle during arc deliveries
5- The accuracy of MLC leaf positioning and its dynamic performance

1.9.1 THE LINAC ISOCENTRE POSITION FOR RADIOSURGERY/RADIOTherAPy

In high precision dose delivery techniques such as stereotactic radiosurgery (SRS) or stereotactic radiotherapy (SRT), the accuracy of the treatment procedure is vitally important due to the delivery of high radiation doses to a localized target (Rahimian et al. 2004, Ramaseshan and Heydarian 2003, Peace et al. 2008). Geometric errors in the system are the main causes of inaccuracies in stereotactic therapy techniques (Du et al. 2010), which may lead to serious consequences since the majority of these treatments are applied to intracranial tumours (Lutz et al. 1988). The most crucial geometric characteristic in SRS/SRT treatments is the exact position of the target relative to the linac isocentre during beam delivery (Treuer et al. 2000, Arjomandy and Altschuler 2000, Chojnowski and Gajewski 2010). Implementation of efficient treatments requires the development of comprehensive quality assurance protocols to ensure high levels of geometric accuracy. These should include methods to control the isocentre position which is affected by: gantry excursions during rotation as a result of its unbalanced weight (Skworcow et al. 2007, D’Souza 1999), irregularities caused by the precision bearing system used for movement control (Lutz et al. 1988, Skworcow et al. 2007, D’Souza 1999, Winkler et al. 2003, Friedman and Bova 1989, Drouet et al. 2002, Woo et al. 1992, Rosca et al. 2006, Fahring and Holdsworth 2000), and misalignment of the room lasers (Winkler et al. 2003, Torfeh et al. 2008, Grimm et al. 2011). Similarly, the rotation of the collimator and treatment table could affect the geometric accuracy. Up to ±1 mm deviation between the radiation and mechanical isocentre is agreed as the acceptable level for SRS/SRT treatments (Klein et al. 2009) and the relevant parts of the linac or positioning system need to be adjusted if this limit is exceeded. The isocentre verification process must be performed before every single SRS/SRT treatment (Chojnowski and Gajewski 2010, Dong et al. 1997, Hayashi et al. 2009, Takahashi et al. 2006, Galvin and Bednazz 2008).

At present, a variety of methods are used for verification of the linac mechanical isocentre position. A number of these techniques were conventionally based on the application of the Winston-Lutz (W-L) phantom (which includes a small ball bearing positioned at the nominal isocentre) and radiographic test films taken at a few gantry/couch angles (Rahimian et al. 2004, Lutz et al. 1988, Low et al. 1995, Serago et al. 1992, Tsai et al. 1996). The main problem with these methods was their dependence on film measurements which were time-consuming and
highly dependent on the observer. In addition, they could only provide qualitative results (Chojnowski and Gajewski 2010, Dong et al. 1997, Grebe et al. 2001, Ramani et al. 1995) with relatively large uncertainties up to ±0.5 mm (Winkler et al. 2003). Electronic portal imaging devices (EPIDs) have gradually replaced films and made isocentre verification much faster and easier. However, there are certain combinations of gantry and couch angles where EPID images cannot be acquired (e.g., gantry at 90° and couch at 90°) and conventional film-based techniques may have more flexibility in this respect (Dong et al. 1997).

Different phantoms (mostly W-L) have been used for EPID-based isocentre verification at a few discrete gantry/couch positions, and various algorithms have been applied for the analysis of the results with sub-millimetric accuracy (Peace et al. 2008, Du et al. 2010, Dong et al. 1997, Winey et al. 2011, Liu et al. 2004, Du and Yang 2009). In some studies, the beam was collimated by secondary or tertiary collimators and the algorithms were based on global thresholding methods (Chojnowski and Gajewski 2010, Torfeh et al. 2008, Sykes et al. 2008), but this technique has the disadvantage of sensitivity to noise (Lee et al. 1990). Other studies used edge detection filters (Sobel or Canny) combined with an operator to find the centres of the field and the target (Hough transform) (Du et al. 2010, Du and Yang 2009, Sykes et al. 2008). These methods are also sensitive to noise, object size and image artifacts, and require application of filters to reduce the uncertainty (Winey et al. 2011, Pithadiya et al. 2011, Baily et al. 1985, Maini and Aggarwal 2009, Hunt and Nolte 1988). Segmentation of the radiation field and application of a convolution kernel was also suggested for finding the radiation field and the ball bearing centre. In this method, some error is introduced due to image magnification (Winkler et al. 2003). It must be noted that all studies were based on EPID images acquired at discrete gantry angles which cannot completely describe the geometric status of the isocentre (Mao et al. 2009).

Since a variety of methods have been used for the determination of linac isocentre position and considering the importance of the subject, details of these methods and their advantages and disadvantages have been described in a review paper in chapter 6.

1.9.2 EPID AND GANTRY SAG DURING ROTATION

Due to the increasing complexity of radiotherapy techniques, there is a growing demand for more accurate methods of treatment plan dose verification. Film dosimetry has conventionally been used for quality assurance of treatment plans, but replacing them with EPIDs would be a much easier and more time-efficient alternative (Berger et al. 2006, Greer and Vial 2011, van Elmpt et al. 2008). As already mentioned in the previous sections, the gantry does not follow a perfect trajectory during arc deliveries, as a result of the presence of several heavy components in its head, and mechanical imperfections in the bearing system and junctions (Clarke and Budgell 2008, Du and Gao 2011, Drzymala et al. 1994). In addition, the EPID which is mounted
on the linac by a retractable robotic support arm rotates with the gantry during arc deliveries to acquire dosimetry images. But the support arm components are not rigidly fixed to keep the EPID in a perfectly stable position during gantry rotations (Iori et al. 2010). In addition to the introduction of inaccuracies to dosimetry measurements, EPID sag causes image artifacts such as spatial distortion and blurring in images (Midgley et al. 1998, Ali and Ahmad 2009, Sillanpaa et al. 2005). The effect of gravity on the gantry and the EPID movements during treatment must be quantified to develop correction methods to achieve accurate dosimetry results and high quality images (Clarke and Budgell 2008).

Several research groups have investigated the EPID/gantry sag during rotation. One method used for the determination of EPID sag in the imager plane perpendicular to the beam axis was to take images of a ball bearing positioned at or close to the linac isocentre at discrete angles (Du and Gao 2011, Glendinning and Bonnett 2000) or a large number of gantry angle settings (Pisani et al. 2000, Groh et al. 2002). The algorithms developed for marker detection were not described in these studies and the EPID sag was not measured along the beam axis.

Another method investigated changes in position of the centre of a square jaw-defined field. EPID images at discrete gantry angles were generally used (Samant et al. 2002, Ansbacher 2006) or in some cases cine EPID images were acquired during gantry rotations (Bakhtiar et al. 2011, Nicolini et al. 2008). The EPID radial flex was also measured by detection of changes in the size of the square field (Ansbacher 2006). A limitation of these methods is that both gantry and jaw sag affect the measured EPID sag results.

Changes in the EPID images of a grid phantom and a lead block of known dimensions were investigated as a measure of the imager sag along the beam axis (Clarke and Budgell 2008). Some other researchers used specially designed phantoms in conjunction with complicated analysis software (Mao et al. 2008, Mao et al. 2009, Mao et al. 2011, Mamalui-Hunter 2008, Cho et al. 2005) to characterise the source and imaging detector geometry from images acquired at discrete gantry angles. Furthermore, there is a known uncertainty in positioning of the markers in phantoms which affects the results, although a method to overcome this problem has been recently reported (Ford et al. 2011). The application of the Sobel edge detection filter for marker detection can introduce uncertainties (Winey et al. 2011, Baily et al. 1985). These measurements also require the availability of special phantoms.

It must be noted that although EPID sag during arc treatments should be corrected for accurate dosimetry results, the application of EPIDs for dosimetric verification of treatment plans also requires consideration of the gantry sag, since motion irregularities during gantry rotation are not taken into account in treatment plans and this could lead to some differences between the planning system predictions and EPID dosimetry results. Measurement of gantry sag during rotation using EPID has been described by Du (Du and Gao 2011, Du et al. 2012).
1.9.3 THE SECONDARY AND TERTIARY COLLIMATION SYSTEMS SAG DURING ARC DELIVERY

Rigorous quality assurance (QA) procedures for the MLCs must be included in comprehensive QA schedules to ensure the accuracy of dose delivery (Mubata et al. 1997, Budgell et al. 2000, Boyer et al. 2001, Samant et al. 2002, Ezzell et al. 2003). The QA of the collimation system is even more essential in advanced radiotherapy treatments such as arc-IMRT. Rotation of the gantry during arc deliveries can affect the position of heavy-weight components in the gantry head such as secondary collimators (jaws) and MLCs (LoSasso et al. 2008, Samant et al. 2002, Sharpe et al. 2006, Sykes et al. 2008, Ford and Lutz 2004). The MLC system performs a computerized and mechanical self-check; however, it is necessary to develop independent methods to monitor its operation (Mubata et al. 1997).

Although jaws are not commonly used to shape IMRT or IMAT treatment fields, there are some examples of junctioned fields, particularly head and neck fields where one side of the fields are defined by jaws. In addition, for large field sizes, displacement of the jaws could result in inferior target coverage (Van Esch et al. 2011).

The functioning of radiotherapy equipment must remain within the internationally accepted tolerances during their lifetime. This requires QA protocols to routinely check the geometric and mechanical performance of all linac components including jaws and MLCs (Kutcher et al. 1994, Nath et al. 1994, Boyer et al. 2001, Klein et al. 2009). The acceptance limit for deviations in jaw and MLC positioning is within 1 mm (Klein et al. 2009); however, Rangel and Dunscombe have shown that the systematic error in leaf positions should be limited to 0.3 mm for acceptable clinical outcomes (Rangel and Dunscombe 2009). The conventional methods for routine jaw/MLC QA tests which are commonly used in most radiotherapy centres are based on visual inspection of light field and/or film images to check the radiation and light field coincidence (Kirby 1995, Luchka et al. 1996, Mubata et al. 1997, Boyer and Shidong 1997, Boyer et al. 2001, Ezzell et al. 2003, Prisciandaro et al. 2003, Dunscombe et al. 1999, Graves et al. 2001). These methods are not quantitative (Samant et al. 2002) and are usually limited to measurements at a single or few gantry angles. There has been one further study where phosphor plates were used to measure the MLC offsets at a number of distinct collimator and gantry angles by producing MLC-defined star patterns for each measurement setup and using a software to detect the MLC offsets relative to the collimator centre of rotation in the X and Y directions (Rosca et al. 2006).

A number of methods have been introduced for the QA of individual MLC leaf positions using EPID images (James et al. 2000, Vieira et al. 2002, Yang and Xing 2004, Sonke et al. 2004, Baker et al. 2005, Mohammadi and Bezak 2007, Jørgensen et al. 2011). These methods were either limited to static gantry angles or did not discriminate between the gantry, EPID and MLC/jaw
sag where they were performed in the arc mode. However, leaf position is a function of both carriage position and leaf position within the carriage. Systematic discrepancies in MLC leaf positions may result from displacement of the whole leaf carriage and support assemblies (Ezzell et al. 2003). In fact dose errors are more related to gap errors rather than individual leaf position errors (LoSasso 2008, Oliver et al. 2010), such that an error of 0.5 mm in the gap size of 1.5 and 2.5 cm could introduce a dose error of about 2% in head and neck or prostate IMRT fields (LoSasso 2008) and an error of 1 mm in 1.9 and 3 cm gaps leads to around 3.5% and 5% dose errors, respectively (Oliver et al. 2010).

Therefore, investigation of the gravity effect on the opposed leaf banks and jaws during arc deliveries is an important topic which may not only improve machine QA routines, but also benefit patient specific QA checks using EPIDs for arc treatments.

In a previous study, Samant et al. (2002) measured the shift in MLC leaf banks relative to the central axis using EPID images at zero gantry angle assuming their camera based EPID had a rigid geometry (Samant et al. 2002). In practice, the EPID cannot be considered as a mechanically stable structure during gantry rotation in arc treatments (Gao et al. 2007, Ansbacher 2006). In addition, the detection precision was restricted to the size of EPID pixels (Yang and Xing 2004). Although the presence of jaw or MLC carriage sag during gantry rotation has been pointed out in previous studies, no other reports were found in the literature about their measurement.

### 1.9.4 DETERMINATION OF THE GANTRY ANGLE DURING ARC DELIVERIES

For verification of the predicted doses in IMAT treatments, it is necessary to provide gantry angle-resolved dosimetric information (Ling et al. 2008, Mans et al. 2010, Bhagwat et al. 2010, Teke et al. 2010). Misalignment of the linac angular settings could severely affect the dose distribution of an IMAT plan delivery and have serious clinical consequences due to the steep dose gradients and complex MLC shapes (Chang et al. 2007).

For routine QA of linacs, a level is positioned on a flat surface of the gantry head close to the graticule and the gantry is rotated until the bubble settles at the centre. Using this method, the gantry angle indicator can be checked only for cardinal angles and the flatness of the surface usually remains unchecked (Chang et al. 2001 and 2007). Another method suggested for QA of the angle indicator is to perform a star shot on film at distinct static gantry angles and determine the angle based on the film setup (Chang et al. 2007). This method is not suitable for testing in arc mode and introduces the difficulties of film measurements and processing. A ±1 degree limit has been recommended as the action level for the gantry angle readout system by the AAPM Task Group 40 (Kutcher et al. 1994).
In a study on linac gantry angles during arc treatments, cine images were acquired during IMAT deliveries using an electronic portal imaging device (EPID) with a specially designed phantom in the beam. The accuracy of gantry angles recorded in the header of these DICOM images were investigated by comparison to the angles derived by following the phantom features in each image (Ansbacher et al. 2010, McCowan et al. 2011). It was found that the same angle may be repeated in the headers of successive images due to the low frequency of angle readout update.

Other proposed methods for the determination of gantry angle were based on EPID images of phantoms containing a number of radio-opaque fiducial markers. Edge detection filters or thresholding methods were used to detect the marker edges and numerical optimization functions were applied to find the centre of each ball bearing. Gantry angles were derived from the relative positions of the markers in the images (Cho et al. 2005, Mamalui-Hunter et al. 2008, Mans et al. 2010). However, the information for different gantry speeds have shown a delay of about 0.4 s (~one frame) in the data gathered by the acquisition system (Mans et al. 2010).

1.9.5 THE ACCURACY OF MLC LEAF POSITIONING AND ITS DYNAMIC PERFORMANCE

It is essential for patient safety that the MLC performance be routinely monitored through strict quality assurance programs to ensure the accuracy and reproducibility of leaf motion in every fraction of the treatment plan (LoSasso et al. 1998 and 2001). In IMAT treatments the QA of the collimation system is even more crucial, since doses are delivered in more complex plans with the MLC shape, gantry speed and dose rate changing during treatment in single or multiple gantry rotations (Yu et al. 1995, Otto 2008). In fact, it is rather challenging to provide appropriate techniques for the QA of linacs used for such sophisticated techniques.

Errors in MLC positioning may be either due to the inaccurate positioning of individual leaves or the result of systematic shifts in the leaf banks (Rangel et al. 2009) or MLC carriage. The accuracy of individual leaf positions may be affected by several factors including mechanical imperfections, gradual degradation of the performance of each motor (LoSasso et al. 2001), cable communication malfunctions (Jorgensen et al. 2011), and loss of counts by the potentiometer encoders (LoSasso et al. 2001). Leaves may slow down, get stuck, be misaligned, skewed or be affected by inertia or gravity. In addition, the drive support assemblies could fail or have delay in communications with the MLC controller (Litzenberg et al. 2002, LoSasso et al. 2008). The MLC bank alignment during gantry rotation may be affected by gravity which causes drifts in the carriage drive and its supporting assemblies (LoSasso et al. 2001, Mubata et al. 1997), since they are heavy components and their weight could affect the window widths.

Dosimetric consequences of systematic errors in leaf banks are reported to be proportional to the extent of errors and are more pronounced for treatments with small window openings (Rangel et al. 2009, Luo et al. 2006). Every 1 mm shift of the leaf banks has been predicted to result in 2.7% and 5.6% change in the reference equivalent uniform dose for prostate and head and neck fields, respectively (Rangel and Dunscombe 2009). In another study, Monte Carlo simulations have predicted ~6% dose difference as a result of 1 mm MLC systematic error in prostate step and shoot IMRT treatment plans (Luo et al. 2006). An error of 1 mm in dynamic IMRT delivery with a window of 2 cm has been reported to cause 5% dose error (Richart et al. 2011). However, measurements of the effect of 0.5 mm systematic offset in the leaves have shown up to 12% and 6% dose difference for head and neck and prostate fields, respectively (Rangel et al. 2010).

Budgell et al. (2000) have shown that accurate dose delivery for IMRT fields requires better than 1 mm accuracy in leaf positioning. The AAPM task group report 142 recommends ±1 mm as the MLC positioning tolerance (Klein et al. 2009) while the ESTRO guidelines propose ±0.5 mm as the acceptance criterion (Alber et al. 2008).

Different devices have been used for the QA of MLC leaves for IMRT/IMAT delivery. The conventional method for two-dimensional (2D) checking of MLC leaf positions was to use film images of a dynamic MLC (DMLC) leaf pattern (Chui et al. 1996, Dirkx et al. 2000, Venencia et al. 2004, Chauvet et al. 2005, Ling et al. 2008, Sharma et al. 2011). Film images were either visually inspected or scanned, digitized and contrast enhanced (LoSasso et al. 2001). Visual inspection is a subjective and inaccurate method for checking the leaf pair alignment and uniformity of the gap widths. Using the scanned films is a better option but it is time-consuming and labour-intensive (Vieira et al. 2002 and 2006, Chang et al. 2004, Baker et al. 2005). Another disadvantage of films is their pixel-to-pixel noise (Low et al. 2001).

Electronic portal imaging devices (EPIDs) provide images in digital format which can be used to provide the image data required for quick analysis. They are easy to use and have a similar level of sensitivity as films for MLC QA applications while their imaging speed can be adjusted to catch up with the leaf motion (Chang et al. 2004). Therefore, EPIDs were considered as more efficient alternatives to films, allowing the test to be performed more frequently (LoSasso et al. 2001, Samant et al. 2002, Chang et al. 2004, Baker et al. 2005, Richart et al. 2011). EPIDs have been used for the QA of MLC performance in several studies (James et al. 2000, Vieira et al. 2002 and 2006, Zeidan et al. 2004, Sonke et al. 2004, Baker et al. 2005, LoSasso et al. 2008, Richart et
al. 2011, Jorgensen et al. 2011, Mei et al. 2011). There have also been reports on the application of 2D array detectors such as MapCheck, MatriXX and PTW-729 (Li et al. 2003, Moreno et al. 2011) for this purpose. The major concern about these devices is their low resolution.

Another method for the QA of MLC leaves would be to use the dynamic log files (or Dynalog files) created at the end of each IMRT delivery by the Varian MLC controller software. Dynalog files have been tested (Li et al. 2003) and used as reference in many studies on the leaf positioning accuracy (Litzenberg et al. 2002, Zygmanski et al. 2003, Venencia et al. 2004, Stell et al. 2004, Ling et al. 2008, Okumura et al. 2010). However, the validity of the data in Dynalog files strongly depends on the accuracy of leaf position readouts. Any possible drift in the encoder readings could be temporarily improved by the MLC re-initialization (LoSasso et al. 2008), but it must be routinely checked by independent imaging methods (Litzenberg et al. 2002, Mubata et al. 1997).

Accurate delivery of a dynamic IMRT/IMAT treatment requires not only accurate leaf positioning, but also correct leaf speeds (Richart et al. 2011). The leaf speed and its acceleration/deceleration should be investigated since they could affect the beam delivery and lead to artifacts in the beam intensity profile (Wijesooriya et al. 2005, Chui et al. 1996, Vieira et al. 2002). The MLC leaf positioning error is reported to be proportional to the leaf speed (Ling et al. 2008). Investigation of the stability of MLC has been performed by film imaging (Chui et al. 1996) or looking at the MLC log files at cardinal angles (Wijesooriya et al. 2005).
PART III. RESEARCH DESIGN

In this project, the research problems explained in section 1.9 are investigated using EPID images, whether the problem is directly related to the imager itself (e.g. the effect of backscattered radiation, or EPID sag measurements), or the problem is not related to the EPID, and the imager is only used as an independent measurement tool (e.g. investigation of MLC performance or verification of the linac isocentre).

a) Effect of arm backscatter

In the first part of this project, the effect of backscattered radiation from a Varian E-type support arm on EPID dosimetry is comprehensively investigated. The effect has been improved by three methods:

1) Development of an empirically derived backscatter kernel to describe the backscatter from a Varian E-type support arm. This kernel is then incorporated into an existing EPID dose deposition model to predict the effect of arm backscatter and achieve more accurate dosimetry results. The measurement-based kernel is also compared with a Monte Carlo calculated backscatter kernel. Details of this part of the project are given in chapter 2.

2) Using lead sheets between the EPID rear housing and the E-type support arm. A detailed experimental study is required on how the EPID backscatter artifacts depend on lead shielding thickness. The optimal lead thickness is determined and the effect of adding lead to the imaging system is investigated by determination of the field size dependence, field symmetry and the image quality. The stability of detector position during clinical arc treatments after the addition of lead sheets is also investigated, and finally the effect on patient skin dose is evaluated. This part of the study is explained in chapter 3.

3) Using a combination of the above methods by placing a piece of lead-shielding just above the arm area to reduce the effect of structural non-uniformities. The limiting problem with the weight of the large lead sheets is not present in this setup and therefore thicker layers of lead are used. The resulting backscatter kernel is measured and included in the existing dose prediction model. The effect on the accuracy of dosimetry with the lead-shielded arm and the modified model predictions is investigated by comparison to measurements as well as the previous model. This part of the work is discussed in chapter 4.

b) Effect of backscatter from the bunker structure

The effect of backscattered radiation from the treatment bunker walls, ceiling and floor is always assumed to be very small, but the possibility of introducing inaccuracies in EPID dosimetry results has not been investigated before, and its level of importance has not been quantitatively evaluated. The presence of such an effect could lead to errors not only in static
measurement conditions, but also during arc deliveries, since the distance between the EPID detector and the surrounding walls continuously changes during arcs; therefore, in this study it is investigated for both modes. Details of the investigation are given in chapter 5.

c) Determination of the linac mechanical isocentre position for radiosurgery/ radiotherapy treatments

The measurement method used in this project is to acquire cine EPID images of a Winston-Lutz phantom acquired during an entire 360° gantry rotation in stereotactic treatment setup to include every possible gantry angle in arc treatments. A robust algorithm is developed to automatically find the isocentre misalignment in each image (angle) with sub-pixel accuracy. The method is fast and highly accurate and is tested for routine pre-treatment quality assurance of the isocentre position during arc delivery with sufficient accuracy for SRS/SRT. The method is applicable to every linac make and model.

Due to the lack of classified information about the large number of different existing methods on this topic, chapter 6 includes two papers: first, a literature review article on the existing methods; and second, the paper which explains the method introduced as part of this project.

d) EPID and gantry sag during rotation

A simple, accurate, fast and automated method is proposed for detection and correction of the gantry wobble and three-dimensional EPID sag during arc deliveries. It is based on cine EPID images of a simple phantom acquired during an entire 360° gantry rotation. The gantry and detector sag are separately quantified at each gantry angle. The improvement in the accuracy of EPID dosimetry measurements using the derived corrections is investigated. Having the sag patterns for the gantry and EPID in clockwise and counter-clockwise directions is very helpful for the linac QA procedures. The method can be applied to every linac make and model. Chapter 7 contains detailed explanation of the EPID and gantry sag determination and correction methods used in this project.

e) The secondary and tertiary collimation systems sag during arc delivery

The sag of jaw and MLC systems are investigated for arc delivery treatments; therefore, the study is again based on cine EPID images of a simple phantom acquired during entire gantry rotations while remaining unaffected by the EPID and gantry sag. A fast, simple measurement method and a fully automatic algorithm are developed to quantitatively detect the positional displacements in the secondary and tertiary collimator systems with sub-pixel accuracy. The method is applicable to every linac make and model. Details of this part of the project are given in chapter 8.
f) Determination of the gantry angle during arc deliveries

A simple and easy-to-use phantom is proposed to be used in cine EPID images during a 360° arc and a fast accurate algorithm is developed to determine the gantry angle for each image. The measurement results are evaluated by comparison to the linac log files as reference. Furthermore, commercially available inclinometers which are supplied in conjunction with popular array dosimeters for pre-treatment verification of IMAT plans are compared with the proposed EPID-based method. For this purpose, simultaneous angle measurements by independent methods are compared with the linac potentiometer readouts at five gantry speeds. This part of the study is explained in chapter 9.

g) The accuracy of MLC leaf positioning and its dynamic performance

MLC characteristics during IMRT/IMAT deliveries are examined using an EPID-based method, and reliable QA techniques for MLC leaves are implemented. The method is based on some popular test patterns that have already been accepted and used worldwide. The elements which are quantitatively investigated include:

1. The accuracy and stability of each individual MLC leaf position and their resulting gap widths;
2. The skewness of leaf banks/carriage;
3. The speed of each leaf over the whole range of its positions;
4. The acceleration/deceleration of each individual leaf.

The method is based on the acquisition of EPID images in integrated and cine imaging modes and development of robust and highly accurate codes to automatically analyse the image data. The ultimate goal for this part of the project is to develop faster and more accurate QA techniques for MLC leaves in both dynamic and arc IMRT deliveries. Chapter 10 includes a comprehensive investigation of the MLC positioning and dynamic performance as studied in this project.
CHAPTER 2

Measurement and modelling of the effect of support arm backscatter on dosimetry with a Varian EPID

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Abstract

**Purpose:** Amorphous silicon EPIDs have been used for planar dose verification in IMRT treatments for many years. The support arm used to attach some types of EPIDs to linear accelerators can introduce inaccuracies to dosimetry measurements due to the presence of metallic parts in their structures. It is demonstrated that this uncertainty may be as large as ~6% of maximum image signal for large fields. In this study a method has been described to quantify, model and correct for the effect of backscattered radiation from the EPID support arm (E-Arm type, Varian Medical Systems).

**Methods:** Measurements of a support arm backscatter kernel were made using several 1×1 cm² 6 MV pencil beam irradiations at a sample of positions over the sensitive area of the EPID in standard clinical setup and repeated with the EPID removed from the support arm but at the same positions. A curve-fit to the subtraction of EPID response obtained on and off the arm was used to define the backscatter kernel. The measured kernel was compared with a backscatter kernel obtained by Monte Carlo simulations with EGS/BEAM code. A backscatter dose prediction using the measured backscatter kernel was added to an existing EPID dose prediction model. The improvement in the agreement of the modified model predictions with EPID measurements for a number of open fields and IMRT beams were investigated by comparison to the original model results.

**Results:** Considering all functions tested to find the best functional fit to the data points, a broad Gaussian curve proved to be the optimum fit to the backscatter data. The best fit through the Monte Carlo simulated backscatter kernel was also found to be a Gaussian curve. The maximum decrease in normalized root mean squared deviation (NRMSD) of the measured and modeled EPID image profiles for open fields was 13.7% for a 15×15 cm² field with no decrease observed for a 3×3 cm² (the smallest) field as it was not affected by the arm backscatter. Gamma evaluation (2%, 2 mm criteria) showed the improvement in agreement between the model and measurement results when the backscatter was incorporated. The average increase in Gamma pass rate was 2% for head and neck and 1.3% for prostate IMRT fields investigated in this study.

**Conclusions:** The application of the backscatter kernel determined in this study improved the accuracy of dosimetry using a Varian EPID with E-arm for open fields of different sizes, 8 head and neck and 7 prostate IMRT fields. Further improvement in
the agreement between the model predictions and EPID measurements requires more sophisticated modeling of the backscatter.

Key words: EPID dosimetry, backscatter, support arm, amorphous silicon, Monte Carlo

I. INTRODUCTION

Amorphous silicon (a-Si) electronic portal imaging devices (EPIDs) are extensively used for patient positioning during conformal and intensity modulated radiation therapy (IMRT) treatments.1-3 The detection areas of a-Si arrays are relatively large. They are resistant to radiation damage 4 and can provide on-line, two dimensional digital images with high resolution and signal to noise ratio.5-7

Many studies have shown that EPIDs have potential for dosimetry measurements.4,8-15 Dosimetry with EPIDs has the advantage of obtaining a 2D array of data without the need for time-consuming labor-intensive conventional methods such as dosimetry with TLDs or diodes. Setting the detectors up for measurements and analyzing the data requires a comparatively large amount of work and only yields data for a few specific points within the field.16,17

Particularly for amorphous silicon EPIDs, many studies have shown a linear relation between EPID response and dose.4,12,18,19 Models have been developed to predict a-Si EPID dose with the aim of dosimetric verification of dynamic IMRT fields for pretreatment verification.20-22

One of the factors that can affect the dose measured by a-Si EPIDs is the presence of the support arm. Amorphous silicon EPIDs are mounted on linear accelerators and moved into position using robotic arms. The metallic parts in the structure of the support arms produces backscattered radiation which can introduce considerable inaccuracies to the results of dosimetry measurements using EPIDs.7,23 In order to correct for this effect, Ko et al.7 have used Monte Carlo simulations and suggested that using 5 mm of lead for 6 and 18 MV beams could reduce the backscattered signal from the R-arm of a Varian aS500 EPID.7 Moore and Siebers experimentally tested these values by comparison of images obtained using an aS500 EPID in clinical configuration with those obtained after adding lead behind the imaging cassette.23 They found that adding the lead layers could also improve the image contrast and resolution. The major problem associated with the addition of lead to the EPID would be the heavy weight of the added layers. This method would also result in the introduction of a uniform backscatter to the whole image which would change the dosimetric properties of the EPID. Finally, Greer et al. developed an energy fluence convolution model for a-Si EPID dose prediction. They reported a 6.5% maximum increase
in Varian aS500 EPID response due to the E-arm backscatter. They used an off arm (backscatter-free) pixel sensitivity matrix to remove the effect of backscatter artifacts introduced by the large backscatter component of the flood-field correction image. In this paper a kernel has been experimentally determined to describe the E-arm backscatter of a Varian EPID and compared with the backscatter kernel calculated by Monte Carlo simulations. Finally, the measurement-based backscatter kernel has been incorporated into the EPID dose deposition model introduced by Greer et al. in order to predict the effect of arm backscatter and achieve more accurate dosimetry results.

II. METHODS AND MATERIALS

II.A. Materials

All measurements were performed using a Varian Clinac 2100 linear accelerator (Varian Medical Systems, Palo Alto, CA) using nominal 6 MV photon beams at a rate of 300 MU/min. The EPID used in this study was a Varian aS500 imager with a sensitive area of 40×30 cm² including a matrix of 512 pixels in the crossplane and 384 pixels in the inplane directions. The image detection unit was the IDU20 model. The aS500 has the same physical components as the aS1000. Each pixel has a square pitch of 0.784 mm. The robotic positioning arm was the E-arm type which is mainly constructed of steel sheets of 1-3 mm thicknesses. Details of the arm structure are illustrated in Fig.1. The arm can move the imager in both vertical and horizontal (including inplane or longitudinal and crossplane or lateral) directions. The lateral motions are made by an electromotor. The motor drives gearwheels on a steel saw-tooth rail which slides the EPID along two steel bars fixed to a metallic frame. The rear housing of the EPID cassette is therefore not uniform and has indentations to incorporate these components and also the imager cabling.

All images were acquired using the integrated imaging mode where the frame-averaged EPID signal is reported. This was multiplied by the number of frames to obtain integrated EPID images. Unless otherwise stated the EPID was positioned at a source-detector distance (SDD) of 105 cm. and the EPID was irradiated with a dose of 100 MU. Functional fitting was carried out using Origin software (v8.0725, Origin Lab Corporation, MA, USA) and MATLAB curve fitting toolbox (version: 7.8.0.347 (R2009a), The Mathworks Inc, MA, USA).
II.B. EPID dose prediction model

Details of EPID dose prediction model have been reported in a recent paper by Greer et al. The model was similar to the Pinnacle treatment planning system model for dose calculation. The TERMA (total energy released per unit mass) deposited in the EPID was determined from the energy fluence model using EPID interaction coefficients that were found experimentally. The predicted dose followed this equation:

\[
ER_{\text{predict}}(x,y) = MU \cdot CF \cdot T(x,y) \otimes K(x,y) \cdot O_c(F) \cdot P(x,y)
\]  

Where: 
- \(MU\) is the applied dose in terms of monitor units;
- \(CF\) is a calibration factor that gives a dose of 1 cGy/MU for a 10×10 cm² field size.
- \(T(x,y)\) is the TERMA in the EPID (cGy/ MU)
- \(K(x,y)\) is the EPID dose deposition kernel;
- \(O_c(F)\) is a field size dependent correction factor for the model results for jaw defined field sizes \(F\) to agree with measured EPID data, normalized to 1 for the 10×10 cm² field size;
- \(P(x,y)\) is a matrix of correction factors for the model results for off-axis ratio to agree with the EPID off-axis ratio for a large field irradiation, normalized at the central axis.

A dose deposition kernel is convolved with the TERMA to model the dose deposition in the EPID. This kernel was developed to include both radiative scatter and optical scatter processes within the EPID. It followed this equation:
\[ y = \exp(-25 \times r) + 8 \times 10^{-4} \exp(-1.5 \times r) + 1.75 \times 10^{-5} \exp(-0.22 \times r) \] (2)

The two correction factor matrices were experimentally determined using the ratio of EPID response and model predictions. The EPID was calibrated using a pixel sensitivity matrix (PSM) according to the method introduced by Greer in 2005.\textsuperscript{25}

II.C. Backscatter Kernel Measurements
The effect of arm backscatter was first investigated using images of a 40×30 cm\textsuperscript{2} field encompassing the entire EPID detector in the clinical configuration (mounted on the linear accelerator) and after removal of the EPID from the arm to eliminate the effect of backscattered radiation. The EPID was kept at the same position using lasers and marks on the EPID collision cap (the EPID cover), while the support arm was completely retracted and had no effect on the EPID response. Subtraction of the images showed a maximum increase of approximately 6\% in EPID response due to the presence of the arm (Fig. 2).

To determine the EPID backscatter kernel the EPID was positioned with the SDD at 120 cm. This distance was used due to the restriction on secondary collimator motion that limited which parts of the EPID could be irradiated at smaller distances. "Pencil beams" of 1×1 cm\textsuperscript{2} area were produced using the secondary collimator jaws of the linear accelerator and used to irradiate different parts of the detector array. The radially symmetric pattern given in Fig. 3 shows the position of the pencil beams on the EPID that was used to characterize the response of various areas of EPID that may be affected by backscattered radiation. The collimator was rotated to perform irradiations at different angles. Images were acquired with the EPID in the clinical configuration and off the arm as described above. Each image was repeated three times. In order to make sure that the radiation field was radially symmetric, a flood field image was taken with the EPID off the arm and the symmetry was found to be 100.03\%. The backscatter signal was determined by subtraction of the on arm and off arm EPID response. The backscatter kernels were derived for each position by finding the optimum functional fit through the backscatter signal.
FIG. 2. Three dimensional illustration of the E-arm backscatter effect on EPID response for a 40×30 cm² image obtained by calculation of the percentage difference between measured on and off arm EPID response to a fixed dose

FIG. 3. Top view of the EPID showing the radially symmetric pattern (57 points) used for irradiation of the EPID with pencil beams. The angle between successive radii was 22.5 degrees and the radial distance between the fields was 3 cm, except for the closest series to the center which were at 3.5 cm distance from the central axis. The position of the pattern is illustrated with respect to the support arm and gantry.

II.D. Backscatter Kernel Monte Carlo Simulations
Monte Carlo simulations were performed using EGS/BEAM user code²⁶ (NRC, Canada) and DOSRZnrc module to score dose in the phosphor layer in radially concentric bins. Dose kernels representing the dose deposited in the phosphor layer of the EPID were generated for the EPID on and off the support arm. The EPID cassette off arm was modeled in detail with specifications provided by the manufacturer. The EPID on arm simulation required that the support arm geometry be simplified to a single thickness of material at a specific distance behind the imager. This sort of simplification is typical for Monte Carlo simulations
and has been used in previous studies.\textsuperscript{7,21} Based on measurements combined with technical specifications supplied by the manufacturer, the support arm was approximated as a 3 mm layer of steel at a distance of 3.3 cm behind the imaging cassette back cover. This was based on the largest and closest component of the support arm. In reality the support arm is made of multiple components and materials with complex geometries, as indicated in Fig. 1. However, the replacement of the support arm in the simulation with a single representative layer allows the simulation results to be compared to the measurement results. EGS/BEAM user code DOSRZnrc was used to score dose in the phosphor layer from an ideal incident pencil beam of polyenergetic photons using the central-axis energy spectrum from Chytyk and McCurdy\textsuperscript{27}. DOSRZnrc is a cylindrical geometry user code, so the active EPID imaging area of 40×30 cm\textsuperscript{2} was approximated as a circle of radius 26 cm. Radial bin sizes used were 0.1-1.4 cm in steps of 0.1 cm, 1.4-2.0 cm in steps of 0.2 cm, 2.0-6.0 in steps of 1.0 cm, 6.0-16.0 in steps of 2, 20 and 26 cm, while simulation parameters included PCUT=0.01 MeV and ECUT=0.521 MeV. One hundred billion incident histories were used achieving a statistical uncertainty of approximately ±0.005\% of dose in the central scoring bin, with statistical uncertainty in any individual scoring bin beyond the central bin not exceeding ±0.05\%.

The off arm Monte Carlo backscatter kernel was subtracted from the on arm Monte Carlo backscatter kernel. Curve fitting software was used with multiple functions fitted to the resulting backscatter signal with a Gaussian curve found to be the optimal fit. The fit results were compared to those from the backscatter kernel measurement described above (Section II.C).

\textbf{II.E. Improved Model}

A separate backscatter component was separately added to the existing EPID dose prediction model. According to the results of measurements with pencil beams, it was assumed that backscatter would be present only when the support arm components were irradiated with the primary fluence.

A fluence mask image (a 384×512 matrix with component values of 1 for the area above the arm and 0 for the remaining area) was used to extract the fluence incident on the support arm. This fluence component was then convolved with the backscatter kernel to model the backscattered EPID dose component. As the backscatter dose component was modeled separately it was necessary to “calibrate” or weight the magnitude of this dose component relative to the forward EPID dose prediction. A weighting factor for the backscattered dose convolution result was defined so that the correct additional dose at the central axis due to backscattered radiation was modeled. To determine the weighting factor the additional dose at central axis due to backscatter was experimentally determined for secondary collimator
defined square fields of sizes 3×3, 6×6, 10×10, 15×15, 20×20 and 25×25 cm² by EPID irradiation on and off the support arm. These fields were then modeled with and without the backscatter dose component and the additional model response due to backscatter at central axis quantified. The weighting factor was then modified iteratively until the least squares difference between the model and experimental central axis backscatter was approximately minimized.

The backscatter kernel was added to the existing dose prediction model:

\[
Dose = MU \cdot CF \cdot \left[ T(x,y) \otimes K(x,y) + WF \cdot M(x,y) \cdot T(x,y) \otimes B(x,y) \right] \cdot O_c(F) \cdot P(x,y)
\]  

(3)

Where:
- \( M(x,y) \) is the mask matrix;
- \( B(x,y) \) is the backscatter kernel;
- \( WF \) is the backscatter weighting factor.

The \( O_c(F) \) values were re-derived. However, the \( P(x,y) \) correction is unchanged, i.e. changes in EPID profile modeling are only due to the introduction of the backscatter kernel.

II.E.1. Open Fields

The modified model in Eq. (3) was evaluated by comparison to EPID images of 3×3, 6×6, 10×10, 15×15, 20×20 and 25×25 cm² open fields defined by the secondary collimators with 100 MU irradiations. All measurements were made in a single session to minimize the linear accelerator output variations. For relative dose at the central axis the EPID response was recorded as the average of the central 9×9 pixels. Three measurements were averaged for each field size and all data were normalized to the smallest field.

For beam profile comparisons two methods were used: Gamma evaluation²⁸ was performed and the normalized root mean squared deviation (NRMSD) was calculated for the central 80% of the image profiles to quantify the agreement between the model and EPID measurements. The EPID images were acquired with 0.5 cm of additional solid water buildup placed on the EPID surface as the original model and dose deposition kernel was developed for this setup (the model was shown to perform nearly as well for no additional buildup).

II.E.2. IMRT Fields

The model results with and without backscatter were compared to EPID images acquired on the support arm for eight head and neck IMRT fields and seven prostate IMRT fields. These were produced using step and shoot delivery mode. It is of most interest to compare the new model that explicitly models backscatter to the current dosimetry methods that do not. Therefore, a slightly different EPID calibration was used for the original and new model. For
the original model the EPID was corrected with a pixel sensitivity matrix (PSM) which included support arm backscatter (Fig. 2). This is representative of most current EPID dosimetry practice where models are compared to FF corrected EPID images (the FF is recorded on the arm and hence contains support arm backscatter). For the new model that explicitly models the backscatter for each field, it is more appropriate that the PSM applied to the EPID image does not contain FF backscatter. We could have also compared the new model to EPID images corrected with a PSM containing backscatter by also including this known backscatter component in the model; however, the results would have been identical.

III. Results

III.A. The Backscatter Kernel

The effect of backscattered radiation from the arm on the EPID response to a 1×1 cm² pencil beam for a point above the arm is given in Fig. 4. The presented results are the average of three images.

![Figure 4](image)

**FIG. 4.** Effect of arm backscatter on EPID response measured for an area above the arm. The curves represent EPID response for *on arm* and *off arm* positions and their subtraction.

The figure shows considerable difference between the EPID response profile in the clinical setup and with the EPID removed from the arm. The central peak in the subtraction curve was attributed to uncertainty in jaw positioning and small displacements between the EPID positions on and off the arm, due to the small size of the detector pixels compared with the width of the aligning lasers in the treatment room. The effect of uncertainty in jaw positioning was found by taking five successive images for the same radiation field with the jaws repositioned using the linear accelerator control system before each image. By subtraction of any two of the images (both taken with the EPID on the arm) the effect of
uncertainty in the positioning of the secondary collimators was observed (Fig. 5). Minor changes in jaw position do not significantly affect the response outside of the primary field region but can explain the central peak observed above.

In order to investigate the differences caused by small misalignments in the EPID positioning, the jaws were kept fixed to a 1×1cm² field and the images acquired with the EPID on and off the support arm were subtracted. An artifact similar to Fig. 5 was observed as a result, which shows that even a small misalignment in the EPID position can lead to the appearance of a central peak.

Due to the presence of the two aforementioned effects, the large central peak was excluded for fitting the curve to the on-off arm profile data given in Fig.4. The peak was cut off at the position of the pixel where a sudden rise was observed in the value of on-off arm response. The ignored part of the curve corresponds to the size of the pencil beam on the EPID. The best fit to the data points in Fig. 4 was determined from among several functions in the available fitting software. The optimum fit to the data was given by a Gaussian curve as shown in Fig. 6.
FIG. 6. The fitted curve through (on arm - off arm) data points showing a broad arm backscatter kernel

The same analysis for a sample pencil beam incident on an area of EPID away from the arm is given in Fig. 7. The presented results are the average of three images.

FIG. 7. Effect of arm backscatter on EPID response measured for an area far from the arm. The curves represent EPID response for on arm and off arm positions and their subtraction.

No apparent functional dependence was found for on-off arm data (Fig. 8) which indicates that there is no noticeable change in signal due to the arm backscatter for a pencil beam at this location.
FIG. 8. No effect of arm backscatter on EPID response was detected for an area far from the arm.

A number of kernels were obtained for pencil beams at different positions of the detector array (Fig. 3). Each of them was applied to the model and compared with measurement results for a 20×20 cm² field. Aligned kernels are compared in Fig. 9 for several pencil beam locations.

FIG. 9. Several kernels found for different positions on the EPID (all Gaussian fits to the data points). Note that the kernels are aligned at the central axis for ease of comparison.

Finally the application of the backscatter kernel found for point #2 (which was above the arm, see Fig. 3) proved to be the optimum kernel that gave the closest prediction to measurement results. The selected backscatter kernel followed this equation:

\[ BSK = 6.912 \exp\left(\frac{-(r + 1.394)^2}{90.696}\right) \]  

(4)

Where \( r \) is the distance from the central axis of the kernel in cm.
This kernel was added to the primary model. Figure 10 compares the measured and modeled E-arm backscatter. Figure 10(a) illustrates the backscatter component measured for the whole EPID area using a large open field. It shows the relative difference between the images taken with the EPID on and off the arm. This is comparable to Fig. 10(b) which shows the model predicted backscatter component obtained by the relative difference between the model predictions with and without the backscatter kernel.

![Image](a) ![Image](b)

**FIG. 10.** Whole EPID showing E-arm backscatter component obtained by (a) measurement and (b) modified model prediction

### III.B. Monte Carlo Simulation

Among the several available functions, a Gaussian curve yielded the best fit through the backscatter data estimated by Monte Carlo simulation:

\[
B_{SK_{MonteCarlo}} = 1.921 \times 10^{-21} \exp(-0.014 \times r^2) \; ; \; R^2=0.985
\]  

(5)

In order to compare the Monte Carlo (Eq. 5) and experimentally derived (Eq. 4) kernels, both kernels had to be normalized (Fig. 11). The root mean squared deviation between the curves was 5.1% which may be attributed to the simple definition of arm structure for Monte Carlo simulation.
III.C. Model and EPID Backscatter for Different Field Sizes

Figure 12 shows the results of comparison of the model predicted backscatter magnitude at central axis with the EPID measured backscatter for square field sizes of 3×3, 6×6, 10×10, 15×15, 20×20 and 25×25 cm². All data are normalized to the smallest field size. The measured backscatter is presented by the ratio of the normalized on arm EPID response to the normalized off arm EPID response. The modeled backscatter is the ratio of the normalized modified model results (including backscatter, Eq. 3) to the normalized primary model results (with no backscatter, Eq. 1). The largest relative difference between the model and EPID measurements was 0.5% (for a 10×10 cm² field) and the root mean squared deviation of the two data sets in Fig. 12 was 0.2%.
FIG. 12. Comparison of normalized backscatter model predictions and EPID measurements for different field sizes on the central axis. The EPID response was recorded as the average of the central 9×9 pixels. Both data sets were normalized to a 3×3 cm² field.

III.D. Comparison of the Model and EPID Profiles

III.D.1. Open Fields

Results of comparison of the primary model (without backscatter kernel), modified model (including backscatter kernel) and EPID measured profiles for open fields of various sizes (3×3, 10×10, 15×15, 20×20, 25×25 cm²) are given in Fig. 13. The asymmetry observed in inplane profiles is attributed to the backscattered radiation from the support arm.
Comparison of the curves given in Fig. 13 shows that application of the backscatter kernel has led to improvements in the agreement between the model predictions and the EPID measured profiles especially for larger fields.

In Fig. 13(a), there are off-axis differences between the modified model and EPID measured dose profiles for a 25×25 cm² field. The reason is that the backscatter kernel overcorrects the dose on the central axis due to the application of a single kernel to the whole mask matrix; meanwhile, the model and measurement profiles are forced to match at the centre in order to align. This results in a lower off-axis response for the model relative to the EPID measured data. The small over-response of the original model off-axis can be similarly explained: this model does not account for the arm backscatter that exists mainly at the centre; therefore, forcing the profiles to match at the central axis increases the off-axis profile relative to the EPID measured data.

Gamma evaluation (2%, 2 mm criteria) was also performed for points above 10% of the maximum dose to find the agreement between the model and measurements. Figure 14 shows the Gamma evaluation for a 20×20 cm² field before and after the application of the arm backscatter kernel. The percentage of points passing the criteria increased from 86.0 to 95.8% with backscatter modeling, and the mean Gamma score decreased from 0.663 to 0.291. Results of Gamma evaluation for open fields defined by the secondary collimators are summarized in Table I which shows the improvement in the percentage of points with a Gamma index less than 1 for all field sizes.
The normalized root mean squared deviation (NRMSD) for the central 80% of the profile width is also reported in Table I for all tested fields as another numerical measure for the agreement between the modeled and measured data. For a 20×20 cm² field, using the modified model decreased the NRMSD by 12.2% (from 17.8% to 5.6%) which confirmed the Gamma evaluation results.

![Figure 14](image)

**FIG. 14.** Gamma evaluation for a 20×20 cm² field (a) before and (b) after the application of arm backscatter kernel

<table>
<thead>
<tr>
<th>Field Size (cm²)</th>
<th>Model-No Backscatter Kernel</th>
<th>Model With Backscatter Kernel</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Gamma&lt;1 (%)</td>
<td>Mean Gamma</td>
</tr>
<tr>
<td>3×3</td>
<td>75.2±0.2</td>
<td>0.814±0.005</td>
</tr>
<tr>
<td>10×10</td>
<td>91.4±0.1</td>
<td>0.473±0.003</td>
</tr>
<tr>
<td>15×15</td>
<td>88.7±0.3</td>
<td>0.594±0.011</td>
</tr>
<tr>
<td>20×20</td>
<td>86.0±0.2</td>
<td>0.663±0.023</td>
</tr>
<tr>
<td>25×25</td>
<td>79.9±0.3</td>
<td>0.758±0.011</td>
</tr>
</tbody>
</table>

### III.D.2. IMRT fields

The Gamma evaluation results for the models (with and without the arm backscatter kernel) compared with the EPID measurements for 8 head and neck and 7 prostate fields at
different gantry angles are given in Tables II and III. The Gamma pass rate increased for all fields with the modified backscatter model. The maximum increase in Gamma pass rate was from 91.2 to 96.6% for the head and neck fields and from 91.6 to 95.0% for the prostate fields. The average Gamma pass rate increased from 95.4 to 97.4% for head and neck and from 94.8 to 96.1% for prostate fields. The average mean Gamma for all fields decreased from 0.446 to 0.349 with backscatter modeling for the head and neck and from 0.446 to 0.420 for the prostate fields.

**TABLE II.** Gamma evaluation results (2%, 2mm criteria) for the measured head and neck IMRT fields and model predictions with and without backscatter kernel (two images for each). Percentage of points with Gamma index <1 and mean Gamma values are listed.

<table>
<thead>
<tr>
<th>Field No.</th>
<th>Gantry angle (degrees)</th>
<th>Model-No Backscatter Kernel</th>
<th>Model With Backscatter Kernel</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Gamma&lt;1 (%)</td>
<td>Mean Gamma</td>
<td>Gamma&lt;1 (%)</td>
</tr>
<tr>
<td>1</td>
<td>0</td>
<td>92.3±0.2</td>
<td>0.512±0.003</td>
</tr>
<tr>
<td>2</td>
<td>0</td>
<td>97.1±0.0</td>
<td>0.404±0.001</td>
</tr>
<tr>
<td>3</td>
<td>50</td>
<td>97.2±0.2</td>
<td>0.402±0.012</td>
</tr>
<tr>
<td>4</td>
<td>100</td>
<td>95.2±0.3</td>
<td>0.427±0.007</td>
</tr>
<tr>
<td>5</td>
<td>143</td>
<td>91.2±0.3</td>
<td>0.527±0.015</td>
</tr>
<tr>
<td>6</td>
<td>214</td>
<td>96.6±0.1</td>
<td>0.459±0.001</td>
</tr>
<tr>
<td>7</td>
<td>260</td>
<td>97.1±0.0</td>
<td>0.399±0.000</td>
</tr>
<tr>
<td>8</td>
<td>310</td>
<td>96.8±0.3</td>
<td>0.436±0.009</td>
</tr>
<tr>
<td>Average</td>
<td>95.4±0.8</td>
<td>0.446±0.018</td>
<td>97.4±0.3</td>
</tr>
</tbody>
</table>
TABLE III. Gamma evaluation results (2%, 2mm criteria) for the measured prostate IMRT fields and model predictions with and without backscatter kernel (two images for each). Percentage of points with Gamma index<1 and mean Gamma values are listed.

<table>
<thead>
<tr>
<th>Field No.</th>
<th>Gantry angle (degrees)</th>
<th>Model-No Backscatter Kernel</th>
<th>Model With Backscatter Kernel</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Gamma&lt;1 (%)</td>
<td>Mean Gamma</td>
</tr>
<tr>
<td>1</td>
<td>50</td>
<td>95.5±0.0</td>
<td>0.420±0.004</td>
</tr>
<tr>
<td>2</td>
<td>100</td>
<td>96.0±0.1</td>
<td>0.417±0.004</td>
</tr>
<tr>
<td>3</td>
<td>150</td>
<td>91.5±0.2</td>
<td>0.488±0.000</td>
</tr>
<tr>
<td>4</td>
<td>180</td>
<td>94.4±0.0</td>
<td>0.429±0.006</td>
</tr>
<tr>
<td>5</td>
<td>210</td>
<td>94.3±0.0</td>
<td>0.526±0.007</td>
</tr>
<tr>
<td>6</td>
<td>260</td>
<td>95.9±0.0</td>
<td>0.420±0.006</td>
</tr>
<tr>
<td>7</td>
<td>310</td>
<td>95.8±0.1</td>
<td>0.423±0.001</td>
</tr>
<tr>
<td>Average</td>
<td></td>
<td>94.8±0.6</td>
<td>0.446±0.016</td>
</tr>
</tbody>
</table>

IV. DISCUSSION

In this study a method was developed to reduce the effect of non-uniform backscattered radiation from the E-arm of a Varian EPID on dosimetry results. The arm backscatter introduced a broad asymmetric EPID dose component which confirmed some of the results reported in a previous study. The measured backscatter kernel was compared with the kernel obtained from Monte Carlo simulations. The differences may be attributed to the simplified geometry used for simulations.

For very large fields (25×25 cm²) the improvement was slightly less, probably due to the assumption of a single backscatter kernel for the support arm and application of a simple mask matrix. The simple mask matrix models the arm as a rectangular area of uniform backscatter. In Fig. 10 it is apparent that this does not perfectly model the complex backscatter distribution, and that further improvements in the model can be made in the future perhaps by accounting for backscatter from the metal bars.

Gamma evaluation of the images and NRMSD calculations on image profiles showed that the introduction of the backscatter kernel improved the agreement between the model and the EPID measured profiles in inplane direction for all open fields. The very small 3×3 cm² fields did not improve largely due to the fact that they were not affected by the arm backscatter, while for larger fields improvements were observed in the agreement between the model and measurements.

The agreement between the model predictions and EPID dose measurements also improved for all head and neck and prostate IMRT fields investigated in this study. The reason for
smaller improvement in prostate fields may be the small size of the radiation field and therefore not much affected by the arm backscatter. The small remaining differences between the model and EPID measurements can partly be attributed to the limitations of the Gamma function due to the steep change in dose values at the image borders.

It must be noted that the support E-arm has an asymmetric structure with several different components made of different materials, therefore, it is not possible to achieve a perfect dose prediction model using a simple backscatter kernel. However, introduction of the backscatter kernel to the existing dose prediction model in this study is one step forward to more accurate dosimetry measurements using EPIDs.

ACKNOWLEDGMENTS
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References:


Reduction of the effect of non-uniform support-arm backscatter for dosimetry with a Varian a-Si EPID
Abstract

Backscatter from the metallic components in the support arm is one of the sources of inaccuracy in dosimetry with Varian amorphous silicon electronic portal imaging devices (a-Si EPIDs). In this study, the non-uniform arm backscatter is blocked by adding lead sheets between the EPID and an E-type support arm. By comparison of the EPID response on and off the arm, with and without lead and considering the extra weight on the imager, 2 mm of lead was determined as the optimum thickness for both 6 and 18 MV beam energies. The arm backscatter at the central axis with the 2 mm lead in place decreased to 0.1% and 0.2% for the largest field size of 30×30 cm² using 6 and 18 MV beams, from 2.3% and 1.3% without lead. Changes in the source to detector distance (SDD) did not affect the backscatter component more than 1%. The symmetry of the in-plane profiles improved for all field sizes for both beam energies. The addition of lead decreased the contrast-to-noise ratio and resolution by 1.3% and 0.84% for images taken in 6 MV and by 0.5% and 0.38% for those in 18 MV beams. The displacement of the EPID central pixel was measured during a 360° gantry rotation with and without lead which was one pixel different. While the backscatter reduces with increasing lead thickness, a 2 mm lead sheet seems sufficient for acceptable dosimetry results without any major degradation to the routine performance of the imager. No increase in patient skin dose was detected.
1. Introduction

In recent years there has been a growing interest on the application of electronic portal imaging devices (EPIDs) for dosimetry applications (Kirby and Williams 1995, McCurdy et al 2001, McDermott et al 2006, van Elmpt et al 2008, Sabet et al 2010). The specifications of amorphous silicon EPIDs - as the most widely used type of EPIDs these days- have made them suitable candidates for dosimetry (Munro and Bouius 1998, McCurdy et al 2001, Greer and Popescu 2003, Louwe et al 2004, Winkler et al 2005). One of the problems associated with dosimetry using Varian a-Si EPIDs is the presence of the supporting arm. The Varian a-Si EPIDs (Varian Medical Systems, Palo Alto, CA, USA) are attached to the accelerators by a robotic arm, which moves the imager to the intended position. The older design of the support arms was known as retractable arm (R-arm) which has been replaced by the exact arm (E-arm) type. Both designs include several metallic components in their structure which produce a non-uniform backscattered radiation to the detector array and introduce inaccuracies to dosimetric measurements (Ko et al 2004, Moore and Siebers 2005, Wang et al 2009, Greer et al 2009, Gustafsson et al 2009, Rowshanfarzad et al 2010) Some approaches have been proposed by researchers to model aspects of the backscatter component. One approach has been to model the field size response of the EPID at central axis which is affected by backscatter, EPID scatter and optical glare. Monte Carlo methods were used to provide generic EPID dose kernels (Siebers et al 2004) or imager-specific dose kernels (Wang et al 2009). They have suggested that a uniform water slab with sufficient thickness (1-1.6 cm depending on the imager (Wang et al 2009) or 0.98 gr/cm² with a 21 mm thick air gap (Siebers et al 2004) could be used to include the effect of backscatter from the rear housing and the metallic parts beneath the EPID at the central axis. However, the non-uniform distribution of backscatter was not modelled. Another study has focused on the development of an energy fluence convolution model for a-Si EPID dose prediction and used an off-arm (backscatter-free) pixel sensitivity matrix to remove the effect of backscatter artifacts introduced by the large backscatter component of the flood-field correction image (Greer et al 2009). Images then contained only a field-specific backscatter artifact which is generally smaller. More recently a convolution method was developed that modelled the backscatter contribution to the image by convolving the fluence incident on the arm with an empirically determined backscatter kernel (Rowshanfarzad et al 2010). A field size specific backscatter correction algorithm was introduced recently in a study based on backscatter measurements for different field sizes. This method used reflection of the pixel values of the half of EPID near the arm on to the other half to find the correction matrices. This method improved the results for most of the 49 tested IMRT fields although there were a few exceptions (Berry et al 2010).
The simplest method to minimize the effect of arm backscatter has been used by researchers in Virginia Commonwealth University (Ko et al 2004, Moore and Siebers 2005). They have used Monte Carlo simulations to find the optimal backscattering material for an R-arm EPID using different thicknesses of water, copper and lead and found that placing 5 mm of lead between the arm and the detector could considerably reduce the non-uniformity in the backscatter signal for 6 and 18 MV beams (Ko et al 2004). Later, in another study the imager was experimentally isolated from the mechanical support arm using the suggested 5 mm of lead behind the imager cassette and a maximum backscattered signal of less than 1% from the arm was reported for both energies (Moore and Siebers 2005). The insertion of lead layers seems an easy and feasible method to minimize the non-uniform backscatter from the arm; but to date no report has been released on the applicability of this method to the E-type robotic arms which has an entirely different design and construction. The backscatter from the E-Arm is known to be considerably greater than the backscatter from the R-Arm with a maximum backscatter component of approximately 6.5% (Greer et al 2009) compared to 4.5% for the R-Arm (Lee et al 2008). Moreover the added weight to the arm of a large thickness of added lead is of concern; therefore it is important to determine the minimum backscatter thickness that still gives acceptable dosimetric and imaging performance, and to understand the effect on imager sag, which is particularly important for arc therapy verification (Mans et al 2010, Grattan and McGarry 2010).

In this study the effect of arm backscatter has been investigated for an aS500 EPID mounted on a linear accelerator with an E-arm type support using lead sheets between the EPID rear housing and the support arm. A detailed experimental study of how the EPID backscatter artifacts depend on lead shielding thickness is performed. The optimal lead thickness is determined and the effect of adding lead to the imaging system is investigated by determination of the field size dependence, field symmetry and the image quality. The stability of detector position during clinical arc treatments after the addition of lead sheets is also reported, and finally the effect on patient skin dose is investigated.

2. Materials and methods
A Varian Clinac 2100 linear accelerator (Varian Medical Systems, Palo Alto, CA) was used to provide 6 and 18 MV photon beams. The EPID used in this study was a Varian aS500 imager mounted on the linear accelerator with an active area of 40×30 cm². The panel has 512×384 pixels with pixel size of 0.784×0.784 mm². Details of the EPID geometry and material composition are previously described by several groups (Siebers et al 2004, Van Esch et al 2004, Chang and Ling 2003). The image detection unit was the IDU20 model. All images were acquired using the single (integrated) imaging mode where the signal is integrated over the entire beam
delivery duration. The integrated EPID signal was determined by multiplication of the averaged signal by the number of acquired frames. The gantry angle was set to 0° and the EPID was positioned at a source-to-detector distance (SDD) of 105 cm for all experiments and irradiated by 150 monitor units (MU) at a rate of 300 MU/min, unless otherwise stated. The centre of the EPID detector array was located on the central axis (CAX) of the beam for all experiments. The data were acquired with no additional build-up placed onto the detector surface. The measurements were repeated three times and their averages were used in order to reduce uncertainty. The conditions for the final experiment on the stability of the detector position were different and are described in section II.F.

The robotic support arms used for Varian a-Si EPIDs can move the imager laterally, longitudinally and vertically. The type of arm investigated in this study was the E-arm (figure 1) with a number of metallic components in its structure, including an electromotor, gearwheels, a metallic saw-tooth rail, two steel bars, connecting cables, etc.

The image quality was assessed quantitatively using the portal image processing system (PIPS-pro 3.2.4, Masthead Imaging Corp., BC, Canada) including QC-3V phantom and software. Image quality test measurements were repeated ten times. The effect of adding lead to the imager on patient skin dose was measured using a Markus plane parallel chamber (PTW, Freiburg, Germany) in conjunction with an NE 2570/1 Farmer electrometer (Nuclear Enterprises, UK). A 30×30×40 cm³ solid water slab phantom (Gammex RMI, Middleton, WI, USA) was used for this experiment.

The 1 mm thick lead sheets (40×30 cm²) used in this study were commercially available and purchased from AMAC Alloys Group Company, Victoria, Australia.

![Figure 1. Schematic illustration of the aS500 EPID and its support E-arm. The position of lead layers added for this study is specified between the EPID back housing and the arm.](image)

### 2.1. Determination of the lead thickness

The amount of lead required to eliminate the effect of backscatter from the support arm was determined by the insertion 30×40 cm² lead sheets of 0 to 5 mm thickness in 1 mm increments
to the back of the EPID detector area between the EPID back housing and the support arm (figure 1).

For this purpose it was necessary to compare the EPID images acquired in two conditions: first with the imager mounted on the support arm (normal operating conditions) with lead sheets added beneath the EPID and next with the EPID removed from the arm with similar amounts of added lead to the bottom. The latter condition was set up by detaching the imager from the support arm and setting it to the same position (as far as possible) on the treatment couch using the mechanical pointer and the room lasers. A foam layer was used to replace the couch top, and the EPID position was adjusted to avoid the backscatter from the carbon fibre rails of the treatment couch. Three images with the beam irradiating the entire EPID detector area were taken and averaged for each setup and lead thickness. The percentage relative difference between the averaged images was found using equation (1):

$$RD = \left( \frac{I_{on\ arm} - I_{off\ arm}}{I_{off\ arm}} \right) \times 100$$  \hspace{1cm} (1)

Where: $RD$ is the relative difference (%) and $I$ represents the average image matrix.

The optimum lead thickness was determined by comparison of the maximum relative difference between the corresponding images measured with the EPID on and off the support arm. The optimum lead thickness was specified considering the reduction in arm backscatter signal as well as the additional weight imposed to the imager. This thickness was used for all of the following measurements.

2.2. Arm backscatter with lead: field size dependence

The effect of the addition of lead on the reduction of the asymmetric arm backscatter was investigated by comparison of the EPID images in a range of jaw-defined field sizes (3×3, 6×6, 10×10, 15×15, 20×20, 25×25 and 30×30 cm²) acquired on the central axis with the EPID on and off the support arm, once without lead and once after the insertion of the lead sheets. The average EPID response over an area of 15×15 central pixels was recorded for each image and the average of three measurements was taken for each field size. The percentage relative difference between the EPID response on and off the arm was determined for each condition.

2.3. Field symmetry

The effect of the added lead layers on the radiation field symmetry was investigated by comparison of the in-plane profiles of EPID images taken with the EPID on the arm with and without the lead sheets for all of the field sizes mentioned in section II.B. Three images were averaged for each field size and the average of the five central columns was used to attain
smoother profiles. The images were acquired in flood-field acquisition mode to avoid the flood field calibration image which is normally used by the system software to correct the images, as it has been taken in on-arm position. The pixel sensitivity corrections were accomplished using the method introduced by Greer in 2005 (Greer 2005).

The symmetry of the profiles was determined from the definition by American Association of Physicists in Medicine (Nath et al 1994) as given in Eq (2).

\[
\text{Symmetry (\%)} = \left( \frac{R_{\text{left}}}{R_{\text{right}}} \right)_{\text{max}} \times 100
\]  

(2)

Where; \( \left( \frac{R_{\text{left}}}{R_{\text{right}}} \right)_{\text{max}} \) is the maximum ratio of the EPID response from two symmetric points on either side of the central 80% of the full width at half maximum (FWHM) of the profile.

2.4. Effect of SDD on the arm backscatter

Due to the variations in the position of the EPID support arm during vertical movements, the effect of changing the SDD on the amount of arm backscatter was investigated by taking open images of the entire EPID detector at SDD=105, 110, 120, 130, 140, 150 cm on and off the arm. The maximum relative difference (Eq. (1)) between the on and off-arm images was used for comparison.

2.5. Image quality

In order to check possible deterioration in the quality of images after the addition of lead layers, the resolution and contrast-to-noise level (CNR) of the EPID images taken with and without lead were measured using a QC-3V phantom and accompanying PIPS-pro software. Details of the phantom structure have been described in the literature (Rajapakshe et al 1996, Menon and Sloboda 2004, Poynter 1999) The EPID was set to SDD=150 cm with the QC-3V phantom on its surface. Images of the entire detector area were acquired using 100 MU irradiations and analysed using the f_{50} spatial frequency (line pair/mm). The experiment was repeated ten times to achieve reasonably consistent results.

2.6. Stability of the detector position

A major concern would be the weight of the added lead sheets which might result in sag during the EPID movements due to the effect of gravity. Therefore, it was crucial to investigate the stability of the detector position after the insertion of lead, since a large displacement could deteriorate the accuracy of dosimetry during IMRT and arc-IMRT treatments (McCurdy and Greer 2009). This was performed by detection of the central pixel position during the gantry
rotation. For this purpose, images were acquired at SDD=105 and 150 cm in continuous mode during 360° gantry rotations while delivering 360 MU with 6 MV photons once before and once after the addition of lead. Ten frames were averaged to produce each image. An in-house developed MATLAB code was used to detect the central pixel for all images acquired within each 360° rotation. The code takes an image and normalizes it to the centre of the image size. Then takes profiles in both X and Y directions through this point and finds the 50% value by linear interpolation of the pixel values on either side of the 50% value.

The changes in the position of the central pixel were compared for the EPIDs with and without lead at both SDDs using the root mean square deviation (RMSD) according to Eq. (3).

\[
RMSD(a, a') = \sqrt{\frac{1}{n} \sum_{i=1}^{n} (a_i - a'_i)^2}
\]

(3)

Where \(a\) and \(a'\) are the corresponding values in two series of measurements (with and without lead) and \(n\) is the number of measurements (in this case number of images taken during rotation).

2.7. The effect on patient skin dose

The effect of backscattered radiation from the lead layers on patient skin dose was investigated by comparison of the readings of a Markus plane parallel chamber at the exit surface of a 40 cm solid water phantom positioned isocentrically in the beam with and without lead sheets at the back of the imager. This is the most extreme condition which roughly simulates a prostate treatment and provides the least possible distance between the patient and the imager at 150 cm SDD. Both 6 and 18 MV beams were tested using 20×20 cm² fields.

3. Results

3.1. Lead thickness

The amount of lead required to effectively reduce the non-uniform backscattered radiation from the support arm was determined by comparison of the relative difference between on and off-arm measured images with increasing thicknesses of lead.

The results of relative difference data (%) for various lead thicknesses are shown in figures 2(a) and 2(b) for 6 and 18 MV beams, respectively. The average of five adjacent central in-plane profiles are presented to smooth the profiles.
Figure 2. Comparison of the in-plane profiles showing the percentage relative difference between the on and off-arm EPID images for different thicknesses of lead added to the EPID structure in: (a) 6 MV and (b) 18 MV beams.

The effect of the addition of different thicknesses of lead is more clearly shown in figure 3 by comparison of the maximum values of the profiles in figures 2 for 6 and 18 MV beam energies. The profiles are plotted through the area with the maximum arm backscatter effect.
Figure 3. Comparison of the maximum values in the in-plane profiles of figure 2 for 6 MV and 18 MV beams. A curve is fitted through each of the data series for better visualization. Extrapolation of the data predicts the effect for larger lead thicknesses.

According to the results presented in figure 2 and 3, adding 2 mm of lead results in a sharp decrease in the amount of arm backscatter. Adding more lead can further reduce the arm backscatter effect but at a much lower rate while adding a considerable amount of weight to the imager (up to 6.8 kg for 5 mm lead). Therefore, a 2 mm sheet was selected as the optimum thickness of lead to be added to the bottom of the imager to reduce the non-uniform backscattered radiation from the support arm.

It must be noted that in this study the maximum RDs have been considered to include the most extreme conditions, but in practice only few fields would irradiate right to the edge the EPID (the position of maximums for the setups with lead). If the EPID response at about 10 cm off-axis toward the gantry (the position of the maximum for no-lead conditions in figure 2) is considered as the reference, then the RDs between on and off-arm EPID responses with 2 mm added lead will be 1.3% and 1.2% for 6 and 18 MV beams, respectively.

3.2. Field size dependence

The effect of 2 mm added lead on the imager response for different field sizes is illustrated in figure 4 which shows that the effect of arm backscatter on EPID central axis response has almost disappeared. For the largest field, the relative difference between the detector responses on and off the arm has reduced from 2.3% to 0.1% using 6 MV and from 1.3% to 0.2% using 18 MV beams after the insertion of lead.
Figure 4. The effect of adding 2 mm of lead to the EPID on the reduction of central axis backscatter from the support arm in (a) 6 and (b) 18 MV beams.

It must be noted that the lead layers produce a uniform backscattered radiation to the whole image. This uniform backscatter is not included in figure 4 since the vertical axis scales the relative difference between on and off-arm images which removes the effect. The net backscatter from the lead was measured through comparison of the off-arm EPID response with and without lead. It proved to depend on the field size and increase the EPID signal by up to about 5% for the largest field in both energies (figure 5).
Figure 5. The effect of uniform backscatter from lead for different field sizes determined from the percentage relative difference between the off-arm EPID response with and without lead.

3.3. Field symmetry

The EPID in-plane profiles for different field sizes are presented in figure 6. It can be seen in the figure that for the images taken without lead layers, the asymmetry in the image profiles increases on the side of the imager closer to the gantry particularly for large field sizes. This is due to the increased contribution of the non-uniform backscatter from the arm. These images were corrected by a pixel sensitivity matrix that was backscatter-free. In conventional flood-field corrected images the opposite effect is observed with the profiles lower on the gantry side due to the backscatter in the flood-field image. With the addition of 2 mm lead sheets, better symmetry is observed for the image profiles taken in both energies. It must be noted that due to the uniform backscatter from the lead layers (as mentioned in section III.B) there was a uniform increase in the central axis EPID response for the images taken with lead; therefore, their profiles have been shifted to match with corresponding no-lead image profiles at the central axis for comparison.
Figure 6. The effect of added lead sheets on the symmetry of the in-plane profiles of open field images in (a) 6 and (b) 18 MV beams.

Quantitative results for field symmetry calculated according to equation (2) are given in figure 7.
3.4. Effect of SDD on arm backscatter

The effect of changing source-to-detector distance is shown in figure 8. The results indicate that the change in SDD does not have a major effect on the amount of backscattered radiation from the arm. The maximum difference in the arm backscatter component (between SDD=105 and 150 cm) was 0.95% and 0.57% for 6 and 18 MV energies, respectively.
3.5. Image quality

Results of image quality measurements using QC3V phantom for the EPID with and without 2 mm added lead sheets are given in table 1.

Table 1. Results of image quality measurements comparing images with and without 2 mm lead

<table>
<thead>
<tr>
<th></th>
<th>6 MV</th>
<th>18 MV</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>CNR±SD</td>
<td>f&lt;sub&gt;50&lt;/sub&gt; Resolution±SD (lp/mm)</td>
</tr>
<tr>
<td>No lead</td>
<td>1230±13</td>
<td>0.426±0.027</td>
</tr>
<tr>
<td>2 mm lead</td>
<td>1214±14</td>
<td>0.422±0.026</td>
</tr>
<tr>
<td>Relative difference (%)</td>
<td>-1.3±0.2</td>
<td>-0.84±0.0004</td>
</tr>
</tbody>
</table>

Only a small decrease in the CNR and f<sub>50</sub> resolution of the images taken with lead was detected for both energies which are not expected to have a serious effect on the detection levels required for imaging applications of the EPIDs. These differences were not detectable by naked eye and do not cause any clinically significant reduction in image quality according to the tolerance levels addressed by McGarry et al. 2007.

3.6. Stability of the detector position

The position of the EPID central pixel in X and Y directions was compared for the EPID with and without 2 mm lead sheets for a whole gantry rotation using about 42 images taken during the irradiations (figure 9). The RMSD of the central pixel positions in the X-Y plane was 0.6 pixels for SDD=105 cm and 0.7 pixels for SDD=150 cm (about 0.5 mm for both distances) with a
maximum detected difference (peak to peak) of 0.9 pixels for SDD=105 cm and 1 pixel for SDD=150 cm.

Figure 9. The central pixel displacement in the horizontal plane after the addition of lead compared with the standard clinical conditions during a 360° gantry rotation

3.7. The effect on patient skin dose
No increase was detected in the response of the plane parallel chamber as a result of the addition of 2 mm lead sheets using 6 and 18 MV beams.

4. Discussion
Although the method of adding lead layers between the EPID and the support arm has already been introduced for the Varian R-type arm (Ko et al 2004, Moore and Siebers 2005), a more comprehensive investigation on Varian E-type arm seemed necessary due to the worldwide popularity of the E-arm and significant differences of its design and structure with the superseded R-type arm. The present study has taken more features into account to investigate the applicability of this method including: the amount of extra weight on the imager, field size effect, effect of SDD on the backscattered radiation, EPID sag during arc treatments, field symmetry, and the effect on patient skin dose.

In a recent study, lead and solid water were examined as the backscattering material by comparison of the EPID responses for a number of different measurement setups and EPID configurations (standard clinical and direct) (Gustafsson et al 2009). It was shown that the standard a-Si EPIDs are very sensitive to backscatter radiation and it is preferable to use lead rather than solid water as the material behind the imager, due to the smaller increase induced by lead to the detector signal.
Results of measurements for the determination of the required lead thickness (figure 3) indicated that by adding 2 mm of lead to the back of the imager, the maximum arm backscatter decreased from 6.0% of the off-arm response to 2.4% for 6 MV, and from 3.5% of the off-arm signal to 2.2% in 18 MV beams. However, considering the clinically used area of the EPIDs (10 cm off-axis toward the gantry, figure 2), these values further reduced to 1.3% for 6 MV and 1.2% for 18 MV beams. The addition of more lead sheets could decrease the arm backscatter by a larger extent, but at a slower rate; it is predicted that about 10 mm of lead would be required to reduce the maximum backscatter from the arm to 1% in 18 MV beams and even more than this amount is required for 6 MV. On the other hand, the amount of added weight is another key factor for making a decision on the adequate thickness of lead. Even 5 mm of lead which has proved to reduce the arm backscatter artifact to 2% for 6 and 1.6% for 18 MV beams, results in an extra 6.8 kg weight to the imager which is 2.5 times heavier than the weight imposed by 2 mm of lead. Therefore, 2 mm was considered as the optimal amount of lead to be added to the back of the imager to reduce the effect of non-uniform backscatter from the arm with acceptable results for dosimetry.

Although the maximum backscatter for the R-arm configuration has been reported in another study as 2.5% increase in signal at 6 MV and 3.6% increase at 18 MV - which is much lower than the E-arm configuration- they have proposed to add 5 mm of lead to reduce the arm backscatter to below 1% of the EPID signal for both beam energies (Moore and Siebers 2005). This is 2.5 times more than what we have suggested in this study for reasonably acceptable results with the E-arm type, while E-arms produce a larger amount of arm backscatter due to their structure and the need for addition of thicker lead layers would be expected. The reason for the differences probably lies in the different acceptance levels considered for dosimetry results in the two studies. For the R-arm study, an acceptance level of maximum 1% increase in EPID response was considered, while for the E-arm it is practically impossible to achieve this level according to figure 3. Therefore, in this study the aim was to decrease the arm backscatter to the lowest practically achievable amount which proved to be less than 1.5% increase in EPID response for clinical field sizes. The idea was avoiding too much extra weight on the imager.

Investigation of the arm backscatter for different field sizes on the central axis showed that the effect is more significant for larger field sizes as was expected due to the larger area of the arm that is irradiated and can contribute backscatter. However, the measurements for different field sizes showed that the addition of 2 mm lead sheets could effectively decrease the non-uniform arm backscatter for both energies. Although the non-uniform arm backscatter was reduced using this method, an extra uniform backscatter radiation was added to the detector response due to the backscatter from the added lead layers. This is not a major source of error as previously reported by Gustafsson et al. (Gustafsson et al 2009) but can change the dosimetric
properties of the EPID and therefore needs to be considered for dosimetry measurements, but the correction is much easier for a uniform effect compared with a non-uniform one. The field symmetry improved after the addition of 2 mm of lead by a maximum of 4.1% (from 105.0% to 100.9%) and 2.4% (from 102.8% to 100.4%) in 6 and 18 MV beams, respectively. Regarding the standard radiotherapy limit for symmetry which is defined as: $|\text{Symmetry} - 100\%| < 3\%$, the presence of lead has made a considerable improvement for large field sizes. For clinically used areas of 10 cm off-axis, the improvement is 2.5% for 6 MV and 1.1% for 18 MV beams. The change in SDD did not affect the arm backscatter component more than 1%; therefore, the same amount of lead would be sufficient for all vertical distances. The addition of lead did not have a large adverse effect on the CNR and resolution of the images; hence the original function of the EPID as an imager is preserved. The backscatter from lead layers did not increase the dose at the exit surface of the phantom. This could be due to the distance between the phantom/patient and the lead sheets, since in clinical conditions the detector is positioned at SDD=150 cm which provides a relatively large air gap between the patient and EPID. On the other hand, the inherent copper buildup layer in the structure of the EPID may play a role in the absorption of low energy backscattered radiation. Finally, it was shown that the position of the central pixel in the detector array was not affected by the addition of lead to more than one pixel in the horizontal plane, and the EPID sag was not influenced by the extra weight, which assures the accuracy of this method for IMRT and arc-IMRT dosimetric applications.

5. Conclusions

In this study, a simple method has been used to reduce the amount of undesired non-uniform asymmetric backscattered radiation from the robotic support arm of a Varian a-Si EPID to improve the dosimetric properties of the EPID. Although the addition of 2 mm of lead has reduced the maximum amount of arm backscatter component from about 6.0% and 3.5% for 6 and 18 MV beams to less than 2.5% for both energies, it has not completely removed the non-uniform backscatter from the arm. According to the results of this study, elimination of the maximum arm backscatter to 1% using this method would require large extra weight. However, investigation of the EPID response properties have shown much improved dosimetric performance by adding 2 mm of lead to the imager.
Using this method, the asymmetric arm backscatter was changed to a uniform backscatter from the lead sheets, which makes it much easier to be included in the modelling of the EPID dose kernels.

The method developed in this study can be incorporated into the existing E-arm aS500 or aS1000 EPIDs with no need for any sophisticated mathematical processing or special devices.

Acknowledgments

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Improvement of Varian a-Si EPID dosimetry measurements using a lead-shielded support-arm

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Abstract

Dosimetry measurements with Varian amorphous silicon electronic portal imaging devices (a-Si EPIDs) are affected by the backscattered radiation from the EPID support arm. In this study, the non-uniform backscatter from an E-type support arm was reduced by fixing a (12.2×10.5×0.5 cm³) thick piece of lead on top of the arm, and the remaining backscatter was modelled and included in an existing dose prediction algorithm. The applied backscatter kernel was the average of kernels on different regions of the EPID over the arm. The lead-shielded arm reduced the non-uniform backscatter component by about 50% for field sizes ranging from 3×3 to 30×30 cm² and the field symmetry improved for medium to large fields up to 3%. Gamma evaluation of the measured and modelled doses (2%, 2mm criteria) showed that using the lead-shielded arm in the model increased the number of points with Gamma index<1 by 5.7% and decreased the mean Gamma by 0.201. Even using the lead alone (no modelling) could increase the number of points with Gamma index<1 by 4.7% and decrease the mean Gamma by 0.153. This is a simple and easy method to decrease the non-uniform arm backscatter and improve the accuracy of dosimetry measurements with the existing EPIDs used for clinical applications.

Key Words: Dosimetry; EPID; Backscatter; Support arm
INTRODUCTION

Amorphous silicon electronic portal imaging devices (a-Si EPIDs) have many desirable features for dosimetry applications, including a large pixel area that provide on-line high resolution digital data with a system that is set up quickly, easily and precisely.\(^1,2\) The ability of EPIDs to rotate around the patient with the gantry makes it a powerful tool for dosimetry applications, since the images acquired during rotations may enable real-time dosimetry.\(^3,4\) Therefore, the application of EPIDs for dosimetry purposes has remained a popular research topic for more than a decade. However, some corrections are required for accurate dosimetry results. For instance, Varian EPIDs are mounted on linear accelerators by support arms that have a number of metallic components in various parts of their structure which can produce non-uniform backscatter and introduce dosimetric artefacts, and thus affect dosimetry measurements. Currently, only one model of support arm is manufactured by Varian Medical Systems (Palo Alto, CA) which is called E-type arm (Fig. 1(a)) and has replaced the older design (R-type). Therefore, this study is focused on E-arm backscattered radiation. Different modelling methods have been proposed to improve dosimetry results made by Varian aS500 and aS1000 EPIDs.\(^5\text{-}13\) The EPID dose kernels developed in some of these studies were just limited to the central axis\(^5,8\) and the non-uniformity in the backscattered radiation was not modelled. Some field-size specific improvement methods were also developed using a backscatter-free pixel sensitivity matrix,\(^9\) or by reflection of the pixel values near the arm to the other half.\(^13\) A different approach was Monte Carlo modelling of an isolated EPID from an R-type arm by insertion of sheets of various materials between the arm and the imager.\(^6\) Results were empirically tested by insertion of large 40×30×0.5 cm\(^3\) (~6.8 kg) lead layers to reduce the non-uniform arm backscatter and replace it by a more uniform backscattered radiation from the lead.\(^7\)

In our previous studies, we have used two different approaches: In the first method, an existing dose prediction model was modified by adding a broad experimentally determined backscatter kernel. This kernel was selected among a number of kernels measured on different parts of the EPID, considering the maximum improvement made to the model. Convolution of the incident fluence with the backscatter kernel just for the area of the EPID above the arm improved the dose prediction model. However, this method was not perfect, mainly due to the application of a single kernel to model the backscatter from the entire arm (which has an asymmetric structure with components made of different materials).\(^11\)

In the second method the backscattered radiation from an E-type arm was reduced by insertion of different thicknesses of 40×30 cm\(^2\) lead sheets between the EPID rear housing and the arm. Considering the area of the EPID used for clinical applications and the amount of weight added to the imager, 2 mm was selected as the optimum thickness for lead sheets. Although this method considerably reduced the non-uniform backscatter, it was unable to completely remove...
it. Furthermore, the additional uniform backscatter from the lead sheets had to be included in the EPID dose prediction models. About 2.7 kg extra weight was added to the imager and retrofitting of the lead would be a bit challenging.

In the present study, a combination of the previous methods is used to improve the accuracy of dosimetry measurements. A piece of lead-shielding is placed just above the arm to reduce the effect of structural non-uniformities (Fig. 1). The limiting problem with the weight of the large lead sheets is not present in this setup and thicker layers of lead can be used. The resulting backscatter kernel is measured and included in an existing dose prediction model. The effect on the accuracy of dosimetry with the lead-shielded arm and the modified model predictions is investigated by comparison to measurements and the previous model.

**Fig. 1.** Schematic illustration of: (a) a Varian E-type EPID support arm showing the position of the small added lead layers, and (b) the back of the EPID with small added lead layers just covering the support arm area behind the imager (lead is considered as an arm component)

**MATERIALS AND METHODS**

All measurements were carried out using nominal 6 MV photon beams of a Varian Clinac 2100 linear accelerator (Varian Medical Systems, Palo Alto, CA). The Varian aS500 EPID imager used in this study was mounted on the linear accelerator with an E-type support arm. The EPID had an active area of 40×30 cm² including 512×384 pixels in the cross-plane and in-plane directions, respectively. The images were acquired using an IDU20 image detection unit in flood-field acquisition mode to avoid the calibration images normally used by the system software. The
EPID signal for each image was determined by multiplication of the signal averaged over the central 9×9 pixels by the number of acquired frames. The image profiles were plotted after correction of the images for differences in pixel sensitivity by application of a pixel sensitivity matrix using the method introduced by Greer. Measurements were made using 100 MU irradiations at a rate of 300 MU min⁻¹ at zero gantry angle. No additional build-up was used on top of the detector for these experiments. The uncertainty of measurements was reduced by taking the average of three measurements for each experiment. The error bars represent one standard deviation of the data.

For all off-arm measurements the normal couch top was replaced by a low density foam layer. The EPID was detached from the support arm and placed on the foam layer to avoid the backscatter from the structure of the treatment couch. Room lasers and marks on the EPID cover helped in placing the EPID at the same position as far as practically possible.

Varian a-Si EPIDs have the freedom of movement in the three main directions (lateral, longitudinal and vertical) with the aid of the robotic positioning arms. The arm used in this study (Fig. 1(a)) has a number of metallic components, mainly made of steel. Details of the arm structure are given in a previous publication.

The lead sheets were commercially available and purchased from AMAC Alloys Group Company, Victoria, Australia.

Image data processing was carried out using MATLAB software [version 7.10.0.499 (R2010a), The Mathworks Inc., MA]. Functional fitting was performed using MATLAB curve fitting toolbox [version 2.2 (R2010a)].

**Lead thickness**

In order to reduce the non-uniform backscattered radiation from the mechanical support structure, small sheets of lead (12.2×10.5×0.1 cm³) with a total thickness of 5 mm were fixed on the arm, behind the EPID housing. The area of the lead sheets was sufficient to cover the supporting part of the robotic arm. It is important to note that the lead piece was considered as a part of the arm structure, not the imager (Fig. 1).

The decision on the optimum lead thickness was made according to the previous studies with large lead sheets for both old and new arm designs. It has been suggested to use 5 mm of lead behind the EPID considering that the uniform backscatter from increasing thicknesses of lead reaches a plateau at 5 mm thickness. In addition, our previous study on the E-type arm showed that the decrease in non-uniform arm backscatter with the addition of more than 5 mm of lead is not expected to be large, at least for up to 10 mm added lead.
**Effect of the lead-shielded arm on field size response**

The effect of addition of lead on the backscattered radiation was investigated by comparison of the EPID images acquired in a number of different secondary collimator defined field sizes (3×3, 6×6, 10×10, 15×15, 20×20, 25×25 and 30×30 cm²) on the central axis with the EPID: (a) on the arm in standard clinical conditions (no added lead), (b) on the arm with the added lead, and finally (c) detached from the arm (off-arm) at the same position as the other two measurements. The average of three images was used to analyze the data for each setup. It must be noted that the lead was considered as a part of the arm; therefore, in all off-arm conditions there was no lead beneath the imager.

The percentage relative difference (RD) between the on and off-arm EPID response with and without lead was determined for each field size to represent the backscatter component (equation (1)).

\[
RD(\%) = \left( \frac{I_{(x,y)_{\text{arm}} on} - I_{(x,y)_{\text{off arm}}}}{I_{(x,y)_{\text{off arm}}}} \right) \times 100, \\
\]

Where: \( I \) is the matrix of the averaged images and \((x,y)\) addresses each of the matrix arrays.

Although our previous study had shown that the steel bars did not have a major effect in the backscattered radiation, the experiment was also performed with lead pieces covering both the arm and the steel bars to confirm the minimal contribution of the bars to this effect.

**The effect of lead on field symmetry**

The averages of the 5 central in-plane (gantry-couch) profiles of the above-measured EPID responses for various field sizes were used to investigate the field symmetry. The profiles were plotted after the application of the backscatter-free EPID pixel sensitivity matrix and compared with no further calibration or normalization procedure.

According to the American Association of Physicists in Medicine, the field symmetry is defined as equation (2):

\[
\text{Symmetry (\%) = } \left( \frac{R_{\text{left}}}{R_{\text{right}}} \right)_{\text{max}} \times 100, \\
\]

Where \( R_{\text{left}} \) and \( R_{\text{right}} \) are the EPID responses from two symmetric points on the two sides of the central 80% of the profile full width at half maximum.
**Backscatter kernel measurements**

The area over the arm was divided into nine regions (Fig. 1(b)) that were irradiated with secondary collimator defined 1×1 cm² fields (pencil beams). All measurements were made at SDD=120 cm due to the restriction in the collimators motions. Images were acquired from each region in two conditions: once with the EPID in normal clinical position (on the arm) with the lead sheets fixed over the arm, and once with the EPID detached from the arm (no lead at the back). The three images taken at each condition were averaged and the backscatter signal was determined by subtraction of the corresponding on and off-arm data in each region. The backscatter kernel was derived for each region by selection of the optimum functional fit through the backscatter signal (by comparison of the R² values given by MATLAB curve fitting toolbox). The nine backscatter kernels were averaged to provide the final backscatter kernel for the lead-shielded arm.

**Dose prediction model**

Details of the dose prediction model used in this study are described in previous papers. The EPID interaction coefficients were experimentally determined and used along with the fluence model to predict the total energy released per unit mass deposited in the EPID (TERMA). The dose was determined by convolution of the predicted TERMA with the EPID dose deposition kernel, considering the amount of radiation (MU), correction factors for field size and off-axis ratio, and a dose calibration factor. The measured backscatter kernel was added to the existing model applied to the fluence incident on the arm. The whole process of the addition of the backscatter kernel followed the method described in our previous report in detail. New weighting factor and field size correction factors had to be developed. Therefore, the on and off-arm measurements of the EPID response were required for 3×3, 6×6, 10×10, 15×15, 20×20, 25×25 and 30×30 cm² secondary collimator defined field sizes on the central axis.

**Evaluation of the model**

The accuracy of the model was investigated by comparison of the measured and predicted backscatter components at the central axis, and comparison of the measured and model predicted in-plane dose profiles with and without backscatter correction. Gamma evaluation (2%, 2 mm criteria) was performed for the points above 10% of the maximum dose to determine the agreement between the modelled and measured dose quantitatively.
RESULTS

Effect of the lead-shielded arm on field size response

The percentage relative differences between the EPID signal on the central axis in various on-arm conditions and off-arm setup using different field sizes are given in Fig. 2. This is the additional signal due to the non-uniform backscatter.

![Graph showing percentage relative difference in response with and without lead on the central axis for different field sizes.]

Fig. 2. The percentage relative difference between on and off-arm EPID response with and without lead on the central axis for different field sizes.

The addition of lead decreased the backscatter component particularly for larger field sizes. The largest decrease in the relative difference with the off-arm setup was from 3.5% to 1.3%. As expected, the addition of lead over the bars did not make a considerable difference (a maximum 0.2% increase in RD), and therefore all other measurements in this work were made with lead covering only the arm structure.

In addition, the maximum relative differences between the EPID response in off-arm setup and on-arm conditions with and without lead are shown in Fig. 3. Please note that the values in Fig. 2 are the RDs on the central axis position for various field sizes, while the data in Fig. 3 represent the maximum of RDs for each field size.
Fig. 3. The maximum relative differences between on and off-arm EPID response with and without lead for different field sizes

By placing the lead piece over the arm, the maximum relative difference between the EPID response on and off the arm has reduced for all field sizes, with the largest decrease from 6.8% to 3.7% in a 30×30 cm² field.

The effect of lead on field symmetry
The effect of the addition of lead on the field symmetry is illustrated in Fig. 4 showing the in-plane profiles of the images acquired with the EPID on the arm with and without lead, compared with the off-arm setup. The images were acquired in flood-field mode and corrected by a backscatter-free pixel sensitivity matrix. No further calibration or normalization procedure was applied.
Fig. 4. In-plane profiles of the EPID images taken with the imager off-arm, on-arm with lead and on-arm without lead, showing the effect of lead sheets on field symmetry

There is an increased signal in parts of the EPID closer to the gantry in medium to large field sizes, due to the backscattered radiation from the support arm. The backscatter signal from the lead-shielded arm was lower than the standard arm; therefore, the field symmetry improved as a result of the addition of lead for these field sizes. The symmetry measurement results are quantified for different field sizes according to equation (2) and are presented in Fig. 5.

Fig. 5. Quantitative results for field symmetry (%) in different field sizes

The addition of lead has resulted in up to about 3% improvement in field symmetry compared with the standard clinical conditions. The field symmetry for off-arm images is also shown for comparison.
**Backscatter kernel measurements**

After the reduction of the non-uniform backscatter from the arm, the remaining backscatter was modelled. For this purpose, the backscatter kernels were determined for different regions of the lead-shielded arm (Fig. 6). The structure of the support arm is inhomogeneous with different metallic parts of various sizes and thicknesses, cables, etc. Therefore, although Gaussian curves best fitted the data from all regions, the parameters of the curves were not the same. The backscatter signals were very low in two of the regions and no curve could be fitted through the data points. The average of the backscatter kernels found for different regions (Fig. 6) was used for modelling, as given in equation (3).

![Fig. 6](image)

**Fig. 6.** The backscatter kernels found for different regions of the lead-shielded arm and the averaged kernel which was used in the model

\[
\overline{BSK} = 2.912 \exp\left(-\frac{(r - 0.0437)^2}{21.206}\right),
\]

(3)

Where \(r\) is the distance from the central axis of the kernel in cm. This kernel was included in the dose prediction model to represent the remainder of the arm backscatter.

**Evaluation of the model**

The backscatter component derived from the model and the measurements at the central axis for various field sizes are shown in Fig. 7.
Fig. 7. Comparison of the measured and predicted backscatter components for different field sizes on the central axis

Each measured data point represents the relative lead-shielded on-arm to off-arm EPID measured dose, and each modelled data point shows the ratio of the model result for the model including backscatter kernel to the model with no backscatter correction. The largest relative difference between the model and the measurements was 0.1% for the $30 \times 30$ cm$^2$ field.

In Fig. 8, the in-plane EPID measured dose profiles are compared with the dose prediction model results with and without backscatter correction for different field sizes. The measured EPID images are normalized to the dose at the central axis predicted by the model which contains the backscatter correction. The model predictions with and without backscatter correction are absolute values and are not normalized. The application of the backscatter kernel has resulted in better agreement between the model predictions and the EPID measured profiles, particularly at larger fields.
Gamma evaluation (2%, 2mm) results for points above 10% of the maximum dose for a 20×20 cm² field are given as an example in Fig. 9 to quantify the two dimensional differences between the EPID measured and model predicted doses for: (a) with no lead on the arm and no backscatter correction applied to the model, (b) with the lead-shielded arm and no backscatter correction applied to the model, and (c) with the lead lead-shielded arm and application of the backscatter correction to the model.

The percentage of points with Gamma index less than 1 and the mean Gamma values comparing the modelled and measured doses for different field sizes are shown in Fig. 10 and 11 for the above conditions, respectively. The accuracy of the primary model (with no backscatter...
correction) is also evaluated by comparing it to the off-arm EPID measured dose (where no backscatter signal exists).

**Fig. 10.** The percentage of points with Gamma index less than 1 comparing the modelled and measured dose for different field sizes in various conditions. The lines are only added for more clarity.

**Fig. 11.** The mean Gamma values comparing the modelled and measured dose for different field sizes in various conditions. The lines are only added for more clarity. The error bars are smaller than the symbols used to plot the data.
Gamma evaluation of the measured and modelled doses (2%, 2mm criteria) averaged over all tested field sizes (3×3 to 30×30 cm²) showed that on average the improved model had 88.6% of points with Gamma index<1 and 0.490 mean Gamma, which is comparable to the EPID off the arm where no backscatter exists (90% of points with Gamma index<1 and 0.479 mean Gamma). Even using the lead alone (no modelling) could improve the accuracy of dosimetry results for the existing EPIDs currently used clinically (87.6% of points with Gamma index<1 and 0.537 mean Gamma compared with 82.9% and 0.691 for the conditions without lead and no backscatter model applied).

**DISCUSSION**

The purpose of this study is to suggest a method to increase the accuracy of dosimetry measurements using Varian a-Si EPIDs by reduction of the non-uniform backscattered radiation from the support arm, and model the remaining backscatter. The methods described in our previous studies¹¹,¹² made some improvements, but each had some drawbacks: The backscatter kernel used to modify the dose prediction model in our previous study was just based on the application of one of the kernels (obtained for one of the regions) to the whole area affected by the arm backscatter, while the structure of the arm was very inhomogeneous (section 3.3). This led to some inaccuracies in the predicted dose. Insertion of large lead sheets (40×30×0.2 cm³) was not able to perfectly remove the non-uniform backscatter, while adding a uniform backscatter from the lead sheets. This method not only required changes to the parameters of dose prediction kernel in the existing model, but also imposed about 2.7 kg extra weight on the imager. This may affect the long term accuracy of the EPID positioning.¹⁷

The present study combines the previous methods by incorporating a thicker layer of lead just on the area which affects the dosimetry measurements. Therefore, the non-uniform backscatter is reduced while adding much smaller weight to the structure (only about 0.7 kg). Fixing the small lead piece on the metallic arm was much easier than securely sticking a large lead sheet behind the imager or inside the housing (which is impractical for the existing imagers currently in clinical use).

Interestingly, the lead-shielded arm not only reduced the non-uniformity in backscatter from the arm components, but also reduced the total amount of backscattered signal, while adding extra material usually increases the backscattered radiation. This phenomenon could be explained by considering a number of effects:

1. Some low energy backscattered radiation from the steel parts is attenuated in the lead layer, thanks to the photoelectric effect.
(2) The lead layer also has a self-absorption property due to the photoelectric effect which damps the low energy scattered radiation originated from lead itself.\textsuperscript{10} That is the reason why the addition of more than 5 mm of lead does not increase the backscatter as shown in previous studies.\textsuperscript{6,15}

(3) Amorphous silicon EPIDs are known to be sensitive to low energy radiation due to the presence of the high Z phosphor scintillator layer which increases the probability of photoelectric effect.\textsuperscript{18} In a standard arm with no shielding (mainly made of steel), the dominant interaction between the high energy beams and the arm is Compton scattering which generally leads to the production of low energy backscattered radiation; while for a lead-shielded arm the dominant effect is pair production which releases higher energy backscatter. Therefore, the EPID is more sensitive to the backscattered radiation from a standard arm than a lead-shielded arm.

Measurements of the backscatter component for different field sizes showed that the non-uniform backscatter is reduced by more than 50% when lead is used over the arm (Fig. 2). Shielding the steel bars proved to slightly increase the backscatter (as expected from the results of the previous study) and was therefore abandoned. The field symmetry improved from 106% to 103% for the largest field which means that the symmetry has reached the acceptance criteria even for a 30×30 cm\textsuperscript{2} field, according to the internationally accepted levels.\textsuperscript{19}

Although using pieces of lead has reduced the non-uniform backscatter to a large extent, it is unable to perfectly remove the backscatter. More accurate dosimetry results require the modelling of the remaining backscatter and including it into the dose prediction model. Comparison of the backscatter kernels developed in this study with the kernels found in our previous study for parts of the EPID over the arm (with no lead on the arm) shows that the new curves are more consistent; therefore, their average can be considered as a reasonable representative for the backscatter from the whole arm. This method is more consistent than selection of one of the kernels which best improves the results. The present averaged backscatter kernel has shorter amplitude due to the lower backscatter signal. The dose profiles given in Fig. 8 show that the application of the backscatter kernel has improved the agreement between the measured and modelled profiles for the area above the lead-shielded arm. Another advantage of this model over the previous model (with backscatter kernel but no lead on the arm) is the better agreement on the side of the EPID far from the arm. Detailed Gamma evaluation results in Figs. 10 and 11 have compared the improvement in dose prediction for four setups.

(a) The EPID on the arm without lead and no backscatter correction (the standard clinical conditions)

(b) The EPID on lead-shielded arm with no backscatter correction
(c) The EPID on lead-shielded arm and applying backscatter correction
(d) The EPID off the support arm with no backscatter correction (ideal condition)

The largest improvement is observed with the addition of lead to the arm without the application of backscatter kernel ((b) vs. (a)). Application of the backscatter kernel further improved the EPID dose prediction. The results were close to the ideal condition where the EPID was off the arm and no backscatter is present ((c) vs. (d)), except for the largest and smallest fields. The reason for the difference at the 3×3 and 30×30 cm² fields is that the averaged kernel is over/under correcting the EPID response for backscatter. However, the 30×30 cm² fields are the most extreme conditions. It must be noted that even case (d) has not achieved 100% agreement with the measured dose, which could be due to the limited accuracy of the dose deposition kernel and/or other parameters of the dose prediction model used in this study.

**CONCLUSIONS**

In this study, the non-uniform backscatter from the support arm of a Varian a-Si EPID was reduced by fixing a small sheet of lead over the arm area to improve dosimetry measurements with these devices. Inclusion of the measured backscatter kernel for the new arm structure into an existing dose prediction algorithm improved the dosimetry results to a high level for routine field sizes. This method is simple and easy and does not require imposition of a large extra weight to the imager. According to the results of this study, using a 5 mm thick piece of lead on the arm alone can reduce the non-uniform backscatter to a reasonably low level which would be useful for improved dosimetry results by the large number of aS500 and aS1000 EPIDs currently used in clinics.

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CHAPTER 5

Impact of the backscattered radiation from the bunker structure on EPID dosimetry

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Abstract:
Amorphous silicon electronic portal imaging devices have been investigated and used for dosimetry in radiotherapy for several years. The presence of a phosphor scintillator layer in the structure of these EPIDs has made them sensitive to low energy scattered and backscattered radiation. In this study, the backscattered radiation from the walls, ceiling and floor of a linac bunker has been investigated as a possible source of inaccuracy in EPID dosimetry. EPID images acquired in integrated mode at discrete gantry angles and cine images taken during arcs were used with different field setups (18×18 and 10×10 cm² open square fields at 150 and 105 cm source-to-detector distances) to compare the EPID response at different gantry angles. A sliding gap and a dynamic head and neck IMRT field and a square field with a 15 cm thick cylindrical phantom in the beam were also investigated using integrated EPID images at several gantry angles. The contribution of linac output variations at different angles was evaluated using a 2D array of ion chambers. In addition, a portable brick wall was moved to different distances from the EPID to check the effect at a single angle. The results showed an agreement of within 0.1% between the arc mode and gantry-static mode measurements and the variation of EPID response during gantry rotation was about 1% in all measurement conditions.

Key words: linac, EPID, dosimetry, bunker, backscatter
I. INTRODUCTION

In modern radiotherapy techniques such as intensity modulated radiation therapy (IMRT) or intensity modulated arc therapy (IMAT), it is important to know the delivered dose with a high level of accuracy due to the steep dose gradients in the treatment plans;\(^1\)\(^{-4}\) therefore it is necessary to quantify all possible sources of error in dosimetry measurements.

Electronic portal imaging devices (EPIDs) have been studied and used for dosimetry applications for many years.\(^5\)\(^{-6}\) EPIDs already exist in the structure of modern linacs and are therefore easy to setup. They have a large area of pixels providing a high resolution two-dimensional array of real-time digital data.\(^7\) The response of amorphous silicon (a-Si) EPIDs is reproducible over short and long periods of time\(^8\)\(^{-10}\) and is linearly related to dose.\(^8\)\(^{-11}\)\(^{-12}\) They have been used for verification of IMRT treatment plans\(^13\)\(^{-14}\) and for \textit{in-vivo} dosimetry measurements.\(^15\)\(^{-16}\) However, like any other dosimetry system, EPIDs have their own drawbacks. For instance, the presence of a gadolinium oxysulphide scintillator layer in the structure of a-Si EPIDs has made them sensitive to low energy scattered\(^12\) or backscattered radiation.\(^17\) The effect of non-uniform backscatter from the EPID support arm in Varian linacs has already been investigated in several studies.\(^18\)\(^{-21}\) This led to the idea that there may also be an effect on EPID dosimetry measurements caused by the backscattered radiation from the treatment room structural components.

In the present study, the possibility of inaccuracies in EPID dosimetry as a result of the backscattered radiation from the treatment bunker walls is investigated and its level of importance is relatively evaluated. The presence of such an effect could lead to errors not only in static measurement conditions, but also in dosimetry during arc deliveries, since the distance between the EPID detector and the surrounding walls continuously changes during arcs; therefore, it has been tested for both modes.

II. MATERIALS AND METHODS

All irradiations were performed using 6 MV photon beams of a Varian Trilogy linear accelerator (Varian Medical Systems, Palo Alto, CA). EPID images were acquired in DICOM format using a Varian Portal Vision aS1000 EPID attached to the linac by an E-type supporting arm. The EPID had an active area of 40×30 cm\(^2\) containing 1024×768 pixels.

The bunker walls were constructed of ~2 m thick conventional concrete (~2.4 gr.cm\(^{-3}\)) to provide adequate radiation shielding for 6 and 18 MV radiotherapy beams. The distance between the linac isocentre and the left wall, right wall, floor and ceiling of the bunker were 370, 385, 130 and 145 cm, respectively.

The couch bearing system (for rotation) is installed in a cylindrical cavity (130 cm diameter and 30 cm depth) beneath a circular timber-top. The structure of the floor is not homogeneous in
this part, due to the presence of the steel bearing system of the couch and thick steel sub-
frames. The distance from the roof specified above is measured from the dropped ceiling of the
bunker, which is made of mineral fiber (a low density material) mounted on a grid-work of
metallic frames. The plenum space between the dropped ceiling and the structural concrete
ceiling (~30 cm) provides room to conceal piping, wiring and ductwork.

In order to investigate the effect of backscattered radiation from the bunker construction
components during arc deliveries, EPID images were acquired using 1200 MU irradiations at a
nominal rate of 600 MU/min in continuous (cine) image acquisition mode at a rate of 7.5 frames
per second (6 frames per image). The EPID was positioned at 105 and 150 cm source-to-
detector distances (SDD) during 360° gantry rotations, which yielded one image per ~2.5°
rotation. Images were taken for 10×10 and 18×18 cm² jaw-defined field sizes.

For static mode investigations, integrated EPID images were acquired at SDD=105 and 150 cm
for eight gantry angles in 45° increments using 100 MU at a rate of 300 MU/min. The effect of
backscatter was tested for open 10×10 and 18×18 cm² jaw-defined fields, as well as a 2.5 cm
wide sliding gap and a head and neck IMRT field to provide data for more realistic clinical
treatment conditions.

The linac output variation with gantry angle was also tested, since it might be the reason for
some of the differences observed in the EPID signal during gantry rotation. This was performed
using a MatriXX evolution two-dimensional array of ionization chambers (IBA Dosimetry, Germany)
fixed to the gantry head at 100 cm distance from the source using the head mount supplied by
the manufacturer. The head mount was firmly attached to the gantry head to minimize the
possibility of small movements during rotation and to hold the detector array perpendicular to
the beam at all gantry angles. The detector array consists of 1020 vented 0.08 cm³ ionization
chambers arranged in a 32×32 grid in a 24.4×24.4 cm² area. Measurements with the MatriXX
detector were made using 100 MU irradiations at a rate of 300 MU/min for 18×18 cm² jaw-
defined fields at eight gantry angles in 45° increments. The results were converted into DICOM
format for processing and the central 9×9 cm² of each image was used as the region of interest.

More investigation on the effect of wall backscatter was carried out by independently
measuring the effect. Eight brick blocks (20×60×7.5 cm³ dimensions, density ~1 gr.cm⁻³) were
walled up on a trolley, providing a 80×60×15 cm³ brick layer which was easily moved to
different distances (50, 75, 100, 150, 200 and 300 cm) from the back of the EPID. The EPID was
positioned at SDD=100 cm with the gantry set at 270° and integrated images were acquired
with 100 MU irradiations at 300 MU/min.

Although the MatriXX detector system has a backscatter layer equivalent to 3.5 cm of water in
its structure (22) and it is therefore unlikely to be affected by backscattered photons, a similar
experiment was performed with the wall moved behind the MatriXX.
In addition, a homogenous cylindrical poly-methyl-methacrylate (PMMA) phantom with a diameter of 15 cm was set up isocentrically in the beam along the gantry rotation axis. It was positioned off the end of the table to avoid the table artifacts. Changes in EPID response was investigated for transit dosimetry conditions using 15×15 cm² jaw-defined fields.

In this study, three measurement series were acquired for each setup. Evaluation of the imager response was based on the pixel values from the central 50% of the fields, in order to provide adequately large areas to capture the backscatter signal, and to limit the region of interest to within the field and thus eliminate the effects of any possible sagging of the EPID²³ (or MatriXX) and collimator jaws²⁴ during rotation. The field edges were determined by developing a code which used the image pixel values, picking the first and the last points with grey scale levels larger than 50% of the maximum signal in the cross-plane and in-plane directions. The region of interest was then limited to the central 50% of the field area.

Data analysis was performed using MATLAB programming language and software (The Mathworks Inc., Natick, MA).

III. Results

A. Open square fields

The range of variations in EPID response (difference between the maximum values ±1SD for each data series) for open 18×18 cm² fields at SDD=105 cm and 150 cm and 10×10 cm² fields at SDD=150 cm in both arc and static modes are given in Table 1.

TABLE 1. Range of relative EPID response variations (±1SD) using open square fields in different measurement setups at various gantry angles during cine EPID imaging in a 360° arc and integrated EPID images at distinct gantry angles

<table>
<thead>
<tr>
<th></th>
<th>SDD=150 cm, 18×18 cm² Field</th>
<th>SDD=105 cm, 18×18 cm² Field</th>
<th>SDD=150 cm, 10×10 cm² Field</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cine Mode, Arc Delivery</td>
<td>(0.71±0.02)%</td>
<td>(0.47±0.03)%</td>
<td>(0.52±0.06)%</td>
</tr>
<tr>
<td>Integrated Mode, Gantry Static</td>
<td>(0.77±0.04)%</td>
<td>(0.54±0.04)%</td>
<td>(0.52±0.05)%</td>
</tr>
</tbody>
</table>

Variations in EPID signal at SDD=150 and 105 cm using 18×18 cm² fields are shown in Figure 1 using integrated images acquired at discrete gantry angles in 45° intervals. In addition, the signal from cine EPID images acquired in arc mode at SDD=150 cm using 18×18 cm² fields are plotted for comparison.

One possible source of changes in EPID response at different gantry angles could be the small variation in the linac output which could be caused by small movements of the gantry head.
components which cause changes in the beam path. Results of the MatriXX two-dimensional dosimeter measurements at eight gantry angles in 45° increments are also given in Figure 1. The average results of three measurements series for each condition are used for Figure 1 and the data points in each series are normalized to the EPID response at zero gantry angle. Curves are fitted through each series of data points for easier visual comparison.

![Graph showing EPID response variations with gantry angle](image)

**FIG. 1.** Relative EPID responses from 18×18 cm² fields using: the integrated images acquired at discrete gantry angles in 45° intervals at SDD=150 cm and 105 cm, the cine images acquired during 360° arcs at SDD=150 cm, compared to the relative variations of linac output (measured by the MatriXX) at discrete gantry angles.

Figure 1 shows that the variations in EPID response with gantry angle follow different patterns from the MatriXX measurements (which represent linac output variations). The range of output variation with gantry angle was ~0.4%.

**B. Portable brick wall**

Changes in the EPID and MatriXX response due to the backscatter from a brick wall moved to different distances from the detector are shown in Figure 2. The percentage relative difference of each measurement with the reference condition of no brick wall (just the bunker wall) is plotted as a function of the distance between the wall and the detectors.
Figure 2 clearly shows the effect of low energy backscattered radiation from the brick wall on the EPID signal. The decrease in signal follows a double exponential curve. The MatriXX response remained unaffected by the presence of the wall, as expected. Therefore, any variation in MatriXX measurements at different gantry angles could be attributed to variations in the linac output.

C. Sliding gap and IMRT fields

Although measurements with square fields could be used to reveal the backscatter effect, it was necessary to investigate it in more realistic clinical conditions where the shape and size of the aperture change during the beam delivery. Changes in EPID response as a result of gantry rotation was tested for a simple 2.5 cm wide sliding gap moving across a 18×18 cm² field and also for a clinical head and neck dynamic IMRT field. Measurements were made at static gantry angles in 45° intervals and all data points were normalized to the EPID response at zero gantry angle. Results are shown in Figure 3.
FIG 3. Variation of EPID response using images acquired at discrete gantry angles in 45° intervals at SDD=150 cm for a 2.5 cm wide sliding gap and a head and neck dynamic IMRT field.

Variations in EPID response for the sliding gap and head and neck IMRT field at different gantry angles had a range of (0.85±0.05)% and (0.43±0.02)%, respectively.

D. Transit measurements

The presence of a phantom could affect the beam characteristics and modify the quality of backscattered radiation from the bunker. This effect was tested by placing a homogeneous cylindrical phantom isocentrically in the beam. The EPID responses in a 15×15 cm² field at discrete gantry angles in 45° intervals are compared with non-transit conditions (no phantom in the beam) in Figure 4.

The range of the relative EPID response variations (±1SD) was (1.14±0.11)% when measurements were performed in presence of the phantom and (0.61±0.03)% when there was no phantom in the beam.
FIG. 4. Variation of EPID response in transit conditions: Relative EPID responses from the images taken at discrete gantry angles in 45° intervals in presence of a cylindrical phantom in the beam are compared with no-phantom conditions.

IV. DISCUSSION

Considering the importance of highly accurate dosimetry measurements for advanced radiotherapy treatments, and due to the increasing application of EPID dosimetry techniques, it is important to understand the EPID dosimetry system characteristics and identify all possible sources of error in these measurements and quantify their effects. It has already been shown that EPIDs are sensitive to low energy radiation and therefore the backscattered radiation from the treatment room structural components may have an effect on EPID dosimetry measurement results. The present study was conducted to examine the existence of such an outcome and the extent of its influence.

Results of this study showed that the EPID response varied at different gantry angles and distances from the walls. According to Figure 1, the results observed for variation of the EPID response could not simply be attributed to variations in the linac output at different gantry angles. In fact, the output variations moderated the backscatter effect to some extent. However, since the final effect for EPID dosimetry in patients is a combination of the output variations and the backscatter from the walls, separate evaluation of these effects was not required.

Although sufficient data was provided to prove the existence of the effect of low energy backscattered radiation from the bunker structure on EPID signal, results of the measurements with the portable brick wall provided clear evidence for this effect. Due to the lower density and thickness of the blocks (compared to the bunker walls), the method was unable to exactly
produce the same amount of increase in the EPID signal, but was capable of revealing the presence of the effect.

The photon energy spectrum of a 6 MV beam peaks at about 1 MV and has lower intensity at higher energies.\(^{(26)}\) At this energy range, Compton scattering is the predominant interaction of the beam with both conventional concrete and steel, which were used as the main building material and for system installations, and have effective atomic numbers of 12.5 and 26.0, respectively. Since the EPID detector has a 40×30 cm\(^2\) active array, the portion of backscattered radiation from the bunker building structures which are at angles between \(~175^\circ\) to \(180^\circ\) relative to the incident rays, can mainly be detected. According to the well-known formula for Compton scattered photon energies, the energy of the majority of backscattered photons from the walls would be in the range of 200 to 250 keV. Amorphous silicon EPIDs are known to be more sensitive to this range of energy due to the presence of a high atomic number phosphor scintillator layer in their structure.\(^{(27,28)}\) This leads to changes in EPID response at different gantry angles and different distances from the walls as shown in Figure 1.

According to Table 1, increasing the size of the radiation field from 10×10 cm\(^2\) to 18×18 cm\(^2\) leads to an increase of about 0.2% in the backscatter effect from the bunker construction components. This phenomenon is attributed mainly to the larger interaction area of the wall and partly to the decrease in the mean energy of the beam for larger fields due to the contribution of more low energy photons from the head components.\(^{(29)}\) It must be noted that the range of EPID response variations was less than 1% even for the large 18×18 cm\(^2\) fields at 150 cm SDD. However, the effect of backscatter from the bunker walls was not limited to open square fields and was also present in dynamic dose deliveries as shown in Figure 3 with a range of less than 1% variation in EPID response.

Another point to consider is the possibility of changes in detector position (EPID sag) in the beam direction during arc deliveries. This has already been discussed in detail in a previous study\(^{(23)}\) and based on those results, the average EPID sag effect on EPID response would not be larger than 0.1%.

The backscatter effect was also detected in transit dosimetry conditions. The presence of phantom leads to changes in beam characteristics due to the removal of low energy head-scattered photons. However, it also results in the production of phantom-scattered radiation which have lower energies compared to no-phantom conditions. These photons have a higher probability of Compton scattering with the walls since the probability of Compton effect is inversely related to the beam energy. As a consequence, the presence of a phantom leads to the production of a larger amount of backscattered radiation from the wall with lower energies. As the EPID is more sensitive to low energy radiation, the relative EPID response in presence of the phantom has a larger range of variations (Figure 4).
Variations to the reported values in this study are expected depending on the size and construction of the bunker, but the order of magnitude is not expected to change much as ceiling and floor are often at comparable distances.

Addition of a lead sheet to the back of a Varian EPIDs has already been suggested in previous studies to reduce the non-uniform effect of arm-backscatter on EPID response \(^{18,19,30}\). This method would also be able to effectively remove the low energy wall-backscatter-induced dosimetric inaccuracies. Other manufacturers that do not have the problem of arm backscatter may use a lower density metal sheet to impose lighter weight on their systems.

**V. CONCLUSIONS**

The impact of the backscattered radiation from the walls, ceiling and floor of the bunker was expected to be very small but it was worthwhile to perform a systematic and quantitative study on the subject. This study showed that the effect can be ignored altogether for pre-treatment verifications with the imager panel at the isocentre (SDD= 100 or 105 cm), but the effect gradually increases with increasing SDD and even more so when the larger SDD is combined with transit dosimetry. Fortunately, even in this ‘worst case scenario’ the effect still remains limited to 1% at its maximum.

**ACKNOWLEDGEMENTS**

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Isocentre verification for linac-based stereotactic radiation therapy: review of principles and techniques

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Verification of the linac isocentre for stereotactic diosurgery using cine-EPID imaging and arc delivery

Pejman Rowshanfarzad, Mahsheed Sabet, Daryl J O'Connor, Peter B Greer
Isocentre verification for linac-based stereotactic radiation therapy: Review of principles and techniques

Abstract
There have been several manual, semi-automatic and fully automatic methods proposed for verification of the position of mechanical isocentre as part of comprehensive quality assurance programs required for linear accelerator-based stereotactic radiosurgery/radiotherapy (SRS/SRT) treatments. In this paper, a systematic review has been carried out to discuss the present methods for isocentre verification and compare their characteristics to help physicists in making a decision on selection of their quality assurance routine.

Keywords: isocentre verification, stereotactic radiosurgery, quality assurance, sub-millimetre accuracy
I. INTRODUCTION

A. Stereotactic radiosurgery/radiotherapy

The word stereotaxic or stereotactic is composed of the Greek word “stereos” meaning three dimensional and the Latin word “tactus” which means to touch. Stereotactic techniques are quite well known in various branches of neurosurgery.

Stereotactic radiosurgery is a highly precise technique for the delivery of high gradient conformal doses of ionizing radiation (usually from a linear accelerator) to a localized small target volume with typically 1-3 cm diameter.\(^1\)\(^-\)\(^6\) This method was invented by Lars Leksell - a Swedish neurosurgeon, as a non-invasive method to obliterate the brain tumours located in positions that are difficult to access for surgery.\(^7\) If the whole stereotactic dose is delivered in one session, the treatment is called stereotactic radiosurgery (SRS) and if the stereotactic dose is delivered in multi-fraction sessions it is known as stereotactic radiotherapy (SRT).

Stereotactic treatments may be delivered using three main methods including: Gamma-knife, X-ray knife and CyberKnife. In a Gamma-knife unit, 201 Cobalt-60 sources are arranged in a helmet pointing toward a target in the brain. The X-ray knife technique is linac-based and uses multiple non-coplanar arcs to aim at the centre of the target positioned at the linac isocentre. The treatment beams may be shaped by micro multi-leaf collimators or solid circular cones. In both Gamma-knife and X-ray knife methods, a stereotactic frame is used to fix the skull. Modern CyberKnife techniques are frameless and utilize image guidance systems to localize the tumour (cranial or extra-cranial) and monitor its motion using X-ray images of the bony anatomy. The gantry -which is a compact 6 MV linear accelerator mounted on a robotic arm- and the treatment table are moved to compensate for target motion.\(^8\)

The X-ray knife method for stereotactic treatments is more widely used, since it is cheaper and the linac can also be used for other common radiotherapy treatments (such as intensity modulated radiation therapy).\(^8\) Therefore, the present work is focused on these linac-based stereotactic systems.

The main advantage of stereotactic radiosurgery is minimum damage to the surrounding critical organs (usually brain tissue) by application of a collimated and confined beam directed toward a small preselected target lesion. This provides sharp dose fall-off out of the target volume.\(^2\)\(^,\)\(^4\)\(^,\)\(^5\)\(^,\)\(^9\)\(^-\)\(^14\) As the whole treatment doses (e.g. about 20 Gy to a 5 mm area) are delivered in a single session during stereotactic radiosurgery treatments, it is important to have high spatial accuracy.\(^5\)\(^,\)\(^13\)

Stereotactic radiotherapy requires repositioning of the patient in various fractions as closely as possible. This may introduce uncertainties to the treatment delivery. On the other hand, the time interval between fractions in this technique enables normal tissues to repair and thus improves the treatment outcome as a consequence of radiobiological effects.\(^15\)\(^-\)\(^17\)
In order to deliver successful stereotactic radiosurgery/radiotherapy treatments, it is essential to know the linac isocentre position - which is taken as the beam primary reference location\cite{18,19} with sub-millimetric accuracy during several successive non-coplanar arcs, as well as its precise mechanical pattern during the gantry, collimator and treatment table rotations.\cite{11,13,20-22}

**B. Positional Accuracy**

For dose delivery techniques such as SRS or SRT, it is vitally important to ensure the accuracy of the treatment procedure.\cite{4,10,23} There are several sources of uncertainty in stereotactic treatment systems such as errors in patient positioning, target localization, and dose delivery system.\cite{20,24} It is practically impossible to achieve perfect alignment mainly due to the presence of several geometric errors in the system.\cite{25} One of the critical geometric errors in SRS/SRT treatments is uncertainty in localizing the radiation field centre\cite{26,27} which directly affects the dosimetric accuracy\cite{28} and results in incorrect tumour targeting that may lead to the delivery of inadequate dose to the lesion and/or serious damage to the healthy adjacent tissues.\cite{25,29} Therefore, it is necessary to develop methods to reduce the probability of such errors by extensive and efficient quality assurance programs to ensure high level geometric accuracy of the treatment.\cite{13,30-32} This requires development of strict acceptance levels and safety margins by independent institutions.\cite{32-34}

The most influential geometric characteristic of the SRS/SRT treatments is the exact position of the target relative to the linac mechanical isocentre during beam delivery.\cite{27,35,36} In ideal conditions, the mechanical isocentre is defined as the point of intersection of gantry, collimator and treatment table rotation axes,\cite{1,18,37,38} but with the rotation of gantry, treatment table and collimator, the isocentre also moves in the space due to the mechanical limitations of the linac components.\cite{39-41} These limitations include gantry excursions during rotation due to its unbalanced weight which leads to bending or twisting of the gantry arm\cite{13,30} and irregularities that mainly originate from the precision bearing system of movement control.\cite{1,13,30,42-47} The geometric position of the isocentre during rotations is usually assumed to be inside a virtual spherical volume. Minimizing the isocentre movement could improve the accuracy of stereotactic treatments\cite{18} and this issue has been considered with special attention. The AAPM Task Group Report 142 (2009) recommends that up to ±1 mm deviation between the radiation and mechanical isocentre is acceptable for SRS/SRT treatments.\cite{48} In cases where uncertainties in target position were less than 1 mm, minor effects on dose distribution have been observed which were reported as clinically unimportant\cite{2,49}; but discrepancies larger than 1 mm are not acceptable as they may lead to severe side effects and require adjustment of the relevant parts of the linac.\cite{1,50} It has been reported that 2 mm positioning error in spine SRS could lead to more than 5% loss of tumour coverage and more than 25% increase in dose delivery to the
healthy tissues. \(^{(51)}\) While with an accuracy of 1 mm for SRS treatments, the error in dose delivery was reduced to less than 2\%. \(^{(20)}\) It must be noted that errors in SRS treatment delivery are non-recoverable, since the treatment is delivered in one session, while there is a chance of compensation in multi-fraction treatments such as SRT. The isocentre verification process must be performed before each SRS/SRT treatment. \(^{(36,52-55)}\)

At present a variety of methods are used for verification of the linac mechanical isocentre position. In this paper, these techniques are individually reviewed and discussed to make it easier for users to identify the points of strength and weakness in each and decide which method to choose for their isocentre quality assurance.

### C. Review of isocentre verification methods

The systematic search strategy through the indexed database “Pubmed”, “Scirus”, “Scopus” “GoogleScholar” and “ScienceDirect” was carried out using search terms including “isocentre verification”, “stereotactic quality assurance”, “stereotactic radiosurgery”, “isocentre offset”, “stereotactic quality control”, “mechanical isocentre”. The cited references of these papers that were published in English before March 2011 have also been included.

### II. METHODS USED FOR MECHANICAL ISOCENTRE VERIFICATION

#### A. Mechanical Pointer

The conventional method for isocentre verification in radiotherapy centres is to measure the distance between the tip of a mechanical pointer mounted on the gantry head and a fixed point mounted on the treatment table. \(^{(1,22,30,39,56,57)}\) This method is manual, laborious and time-consuming, and the results depend on the human observer. It is also limited by the size of the tip of pointer which is not exactly a single point. \(^{(31,45)}\) Although in one study the point on the treatment table was defined by a sharp needle attached to a micrometer for better accuracy and measurements were repeated at distinct angles with the gantry rotated clockwise and counter clockwise. The reproducibility of the method was 0.2 mm. \(^{(19)}\)

It must be noted that mechanical pointers are susceptible to damage and can be easily dislocated; therefore they are not quite suitable to be routinely used. \(^{(49)}\)

#### B. Winston-Lutz test

This technique was introduced by Lutz, Winston and Maleki at Harvard Medical School in 1988. \(^{(1)}\) The Winston-Lutz (W-L) phantom is a small metallic ball (made of steel, titanium or tungsten) that represents the planned target and is fixed on the treatment table by a locking mechanism. The phantom position is adjustable in three directions by means of a micrometer tool. \(^{(1,31,39)}\) The collimator used for SRS/SRT is attached to the gantry head and the ball is placed
as closely as possible to the isocentre by aligning the marks on the phantom with the treatment room lasers. The collimated beam is used to expose the radiographic test film mounted perpendicular to the beam direction on a stand behind the ball. The difference between the centre of the sphere shadow and the field centre reveals the isocentre movement\(^{(1)}\) (which must be within ±1 mm for stereotactic treatments). Measurements should be repeated at cardinal angles (0, 90, 180 and 270\(^{\circ}\))\(^{(1,36)}\), which requires changing the film for each setup.\(^{(1,36)}\) The offset is read on each film using transparent template guidance scales\(^{(22,39,58)}\) or scanning the film and software analysis.\(^{(14,36)}\) This method could be used to check the gantry, treatment table and collimator in various angles.\(^{(36,42,43)}\)

The Winston-Lutz test was relatively simple\(^{(2)}\) and became quite popular,\(^{(34,36)}\) but it was based on films; therefore, it inherited all film-related problems. The general disadvantages in using films include the cost of films, chemicals and processor maintenance, and occupation of archiving space. In addition, it is not possible to modify the properties of film images to improve contrast.\(^{(23)}\) Dust or marks on the film can lead to spikes in images.\(^{(40)}\) The film results for Winston-Lutz test are not quantitative\(^{(26,52,59)}\) and are based on manual evaluations\(^{(24,34)}\) and visual inspection which makes them highly dependent on the observer.\(^{(11,36,52,60)}\) The uncertainty introduced by the operator judgment has been reported 0.3 to 0.4 mm,\(^{(36,61,62)}\) although this could be reduced by scanning the films and using software analysis.\(^{(14,36,58)}\) In addition, film adjustment and processing are time consuming\(^{(11,36,52)}\) which makes it difficult to perform the test before each treatment. Furthermore, the measurements made at discrete angles cannot perfectly represent the geometric status of the isocentre.\(^{(63)}\) The uncertainty in film measurements were reported ±0.3 mm by Lutz et al.\(^{(1)}\) 0.2±0.1 mm by Friedman et al.\(^{(43)}\) and ±0.5 mm by Winkler et al.\(^{(42)}\) which were relatively large regarding the ±1 mm acceptance criteria. Even in one study with Gafchromic films, up to ±1 mm uncertainty was detected in isocentre positioning.\(^{(64)}\)

A mathematical method was developed by Low et al.\(^{(39)}\) which used the film-measured isocentre positional errors for eight gantry angle and couch settings to find the suitable offset for the phantom stand to minimize the distance between the linac isocentre and the target.\(^{(39)}\) A similar aim was followed by Grimm et al.\(^{(14)}\) who developed an algorithm to reconstruct the W-L phantom ball locus in three dimensions from two dimensional film images taken at certain couch and gantry angles and combined them with the images of lasers taken by digital cameras. The room lasers were repositioned for each angle to align with the isocentre using a special device that quickly gave the required displacements. The accuracy of the method was reported better than 0.25 mm for most cases.\(^{(14)}\)

In another study on SRS/SRT treatment systems, seven film irradiations at different gantry and couch angles were used to determine the deviation of the beam and the target centres. A
mathematical model was applied to analyse the data which led to seven equations with five parameters. Solving the equations by two computer codes showed the direction and values of discrepancies, which was then used for corrections. The accuracy of the procedure was tested by detection of the manually displaced target positions. The proposed method was too complicated and time-consuming. The application of W-L test has been extended to quality assurance procedures for proton therapy, where the acceptable range of isocentre position is even more limited and has been considered within ±0.5 mm accuracy by Ciangaru et al. In their study, Gafchromic EBT films for 16 gantry angles were digitized; the image noise was reduced by subtraction of a blank film from the exposed film pixel by pixel and smoothed using an averaging MATLAB filter. But this method inherited all of the film-related problems.

C. EPID-based isocentre verification methods

With the introduction of electronic portal imaging devices (EPIDs), they gradually replaced films and made the isocentre verification test procedure much easier. EPIDs have many advantages over films such as the ability to re-use, providing digital images, easy and quick data transfer and archive. The digital format of EPID images enables contrast enhancement which is very helpful in image analysis. It is important to note that EPIDs are currently included in the structure of modern linear accelerators and require no time consuming setup. The limitation of using EPIDs would be few combinations of gantry and couch angles where EPID images can not be acquired.

In the majority of QA procedures, the mechanical accuracy of the system is evaluated by using phantoms containing small, well-defined markers at the centre. The methods and algorithm designs for EPID-based studies using W-L phantom and some of the most popular phantoms are presented in this section.

C.1. W-L phantom

Many research groups have used EPIDs instead of films for the evaluation of discrepancies between the centre of a W-L phantom (target) and the linac mechanical isocentre. In one study, global thresholding technique was used to detect the centre of circularly collimated radiation fields, while the centre of the metallic sphere was determined by an algorithm applying bi-linear interpolation and thus eliminating the observer error. The overall accuracy of the method was within ±0.2 mm. The processing took 3 minutes which is much less than film-based systems, and the method was tested for few angles.

Global thresholding technique was also used in another study where the algorithm detected the centre of the phantom and compared it with the centre of square micro-MLC defined fields
extracted from the penumbra profiles. The process took 15 minutes to complete and was tested for 5 different gantry and couch setups. The results were within 0.3 mm of film measurements.\(^{(36)}\) It must be noted that global thresholding technique is sensitive to noise.\(^{(66)}\) In a similar study by Torfeh \textit{et al.}\(^{(21)}\) thresholding method was used for finding the secondary collimator defined radiation field centre, while the centre of the sphere was determined by convolution of a Gaussian kernel with the EPID image. The method was applied to some distinct angles and tested by detection of an arbitrary phantom offset.\(^{(21)}\) The overall accuracy was not reported and the sag in secondary collimator and EPID (which introduce inaccuracies) were not considered in the processing.

A different approach was used by Winey \textit{et al.}\(^{(58)}\) applying a double convolution method to the EPID image to find the centre of the field and the ball separately with sub-pixel accuracy. Their work was based on a previously developed technique by Guizar-Sicairos \textit{et al.}\(^{(67)}\) The method was applied 7 times at distinct angles and compared to measurements made by: (a) human observer using templates designed as a graphical user interface, and (b) an edge detection and centre of mass algorithm. The proposed method was faster than both due to the higher speed of calculations in Fourier domain. It was also tested by applying known displacements to the phantom and detecting them using the algorithm, and the accuracy was 0.1 mm.\(^{(58)}\)

Calvo-Ortega \textit{et al.}\(^{(62)}\) reported the results of a package designed for the EPID-based W-L test. The field centre was calculated from the penumbra of square profiles and the centre of the ball was found using a summation filter. The method was tested by application of manual shifts to the phantom and detection of the displacements with the algorithm. It was also compared with film measurement results at five angles. The accuracy was reported up to 0.2 mm.\(^{(62)}\)

In another study on W-L test, the radiation field was defined by circular and rectangular collimators fixed to the gantry head. Images of the target at the isocentre were acquired (on EPID or EBT2 films) at discrete angles and analysed: Sobel filter was first applied to each image to detect the edges of the field and the ball; then, Hough transform -which is quite popular in digital image processing- was implemented to localize both the radiation and the ball centres. The signal to noise ratio had a major effect on the accuracy of this method since Sobel and Hough operators are highly sensitive to noise.\(^{(58,68-71)}\) Sobel filter is also dependent on the object size\(^{(69)}\) and its output is almost always more than one value\(^{(68)}\) which results in a thick detected edge and increases the uncertainty of results.\(^{(58,72)}\) It is worth mentioning that Hough transform prolongs the processing\(^{(73,74)}\) as it requires one second per image to process.\(^{(26,58)}\) However, the absolute error for this method was reported as 0.02 mm. Du \textit{et al.}\(^{(25)}\) further investigated isocentre verification for SRS quality assurance by EPID-based W-L test in a number of MLC-defined square fields at cardinal angles, using mechanical graticule.\(^{(25)}\) Similar to the previous work, Sobel filter and Hough transform operator were used to localize the centres.
of the radiation field and the ball bearing. A two-component model was developed to overcome the issue of overlapping centres of the ball and graticule. A 0.47 mm systematic error was considered for the MLC leaf positioning. The problem with this method was sensitivity to variations in image intensity.\(^{(25)}\)

A different image analysis method was introduced by Winkler et al.\(^{(42)}\) for EPID-based W-L test. The method was applied to jaw-defined and circular collimator-defined (used for SRS/SRT) fields. EPID images were acquired at distinct angles and analysed in the following steps: (a) segmenting the radiation field and locating its centre, (b) segmenting the image of the tungsten sphere and locating its centre, and finally (c) determining the deviation of the two centres. A convolution kernel was used to improve the sharpness of the ball image to enable more accurate edge detection by decreasing the noise. The segmentation required image magnification which introduced up to 0.02 mm error. However, the overall accuracy of the method was estimated by comparison to visual and radiographic film tests and was reported as 0.1 mm.\(^{(42)}\)

C.2 Other phantoms

On Elekta Synergy linacs, the routine method for isocentre verification is to use a ball bearing phantom provided by the manufacturer, which has 8 mm diameter and is imaged at cardinal angles with the phantom aligned to the room lasers. The system software uses the edges of the jaw-defined field and the centre of ball bearing in each image. The field edges detected in images acquired at opposing angles were used to remove the jaw sag. This phantom was applied in one study as reference to benchmark the application of a QUASAR Penta-Guide phantom where the central air cavity replaced the ball bearing.\(^{(32)}\) Canny filter was used along with a binary mask to detect the field edges. The lines defining the field edges were determined and Hough transform was applied to the image to identify the four corners of the square field. The field centre was specified by finding the mean of the points at four corners. The centre of the air cavity was determined from the EPID image by using a low-pass filter to the centre of the image (to reduce the noise) and applying thresholding technique to identify the air cavity edges. The centre of the circle fitted to the results was the centre of the air cavity. Comparison of the method with the standard ball bearing phantom results showed slight differences and the systematic error was less than 0.2 mm.\(^{(32)}\) The main concern about this method would be the application of Canny filter which results in broken edges.\(^{(75)}\) In addition, Hough transform method is sensitive to noise.\(^{(71)}\)

QUASAR Penta-Guide phantom has also been used with a simple voxel-based contouring algorithm, considering the centre of the contour as the isocentre. The results were compared with the isocentre indicated by the room lasers. The contouring algorithm involved a large
amount of noise which introduced about 1 mm error to the outcome of the algorithm. The process took less than 6 minutes.\textsuperscript{(31)}

A graphite cylindrical phantom provided by Varian Medical Systems containing 16 tungsten carbide ball bearing markers was used in a different method by Mamalui-Hunter et al.\textsuperscript{(28)} for isocentre verification. The geometric location of markers were used to assess the linac mechanical isocentre in addition to the gantry and couch angles, source-to-detector distance, couch vertical position and gantry sag in each image. This method required the removal of non-uniform background using a numerical optimization function. The ball bearings were detected in the images using a Sobel filter-based method. It was only reported for some distinct angles and the precision was better than 0.1 mm. This method has all the issues mentioned above for the application of Sobel edge detection filter.\textsuperscript{(28)}

A more or less similar idea was used for a cubic phantom containing 13 steel ball bearings for the determination of a number of linac characteristics including the isocentre position from EPID images.\textsuperscript{(76)} The isocentre was found using 36 images (one per 10 degree gantry rotation) and fitting a curve through the estimated location of the radiation beam source. The location of the source was determined by giving weight to the signal intensity of each pixel. The centre of the source trajectory was considered as the isocentre. The accuracy of the method was limited by the source penumbra, pixel noise and uncertainty in measurements of the marker locations which was partly corrected by using a global optimization algorithm. However, the accuracy of setting the ball bearings in the planned positions in phantom was reported 0.5 mm, which introduced some errors to the results. The accuracy of the method was reported up to 1.6 mm.\textsuperscript{(76)}

Another phantom made of a radio-opaque material containing nine spherical indicators was used by Sharpe et al.\textsuperscript{(41)} The central sphere was placed at the nominal isocentre set by the intersection of the lasers in the treatment room. EPID images were acquired with the radiation collimated through a jaw-defined field at cardinal angles. The central axis of the field was determined from the field edges and compared with the target centre of mass. The latter was averaged for opposing angles to remove the jaw flex. The results were noisy and the process took 12 minutes.\textsuperscript{(41)}

D. Miscellaneous methods

Direct monitoring: In a proton therapy facility, isocentre deviations were directly measured by a CCD camera that monitored a scintillator screen exposed to a field containing a steel sphere at the isocentre.\textsuperscript{(77)} The ball shadow was detected inside the light spot generated by the collimator on the screen. An algorithm was used to first make circular fits through data points to find the contours of the beam and the ball shadows. Their centres were determined from the contoured
edges. The accuracy of the method was examined by applying known offsets to the sphere position and detecting them by the algorithm, which was reported up to 0.25 mm and took 10 minutes to complete.\(^{(77)}\)

**Computed radiography:** In a study based on Winston-Lutz method, computed radiography images of a lead ball bearing were acquired at 5 combinations of couch and gantry angles in micro-MLC-defined fields. The shifts in isocentre were manually determined on the magnified images.\(^{(23)}\) The accuracy of the method could be affected by the uncertainty in the position of micro-MLC leaves.

**Optical methods:** Gibbs et al.\(^{(78)}\) investigated the mechanical stability of the isocentre using a diode laser mounted in the radiosurgery collimator. The laser was detected by a photodiode detector that had a position-sensitive surface and rotated with the gantry to stay perpendicular to the laser beam at all times.\(^{(78)}\)

An optical tracking system has been used by Skworcow et al.\(^{(30)}\) along with Gafchromic films to find the mechanical isocentre. A mathematical optimization method was used to detect the isocentre position as well as the intersection locus of collimator axes at 24 gantry angles. The deviation of the two results was recorded as the isocentre displacement. This method was tested by applying manual offsets to the isocentre position and finding it with the proposed system, which showed up to 0.7 mm uncertainty.\(^{(30)}\)

Brezovich et al.\(^{(49)}\) developed a method to investigate the wobble in the isocentre as a result of treatment table rotations. The gantry light field was limited by a circular aperture and a steel ball was set at the nominal isocentre. Light-sensitive film images were taken with the table rotated to different angles and the thickness of the ring shaped image was used as a measure of the isocentre movements. The method was limited to non-dynamic treatments and required 40 minutes to perform.\(^{(49)}\)

**Polymer gel:** Another different approach for the evaluation of the isocentre in Leksell SRS/SRT system involved the application of polymer gels used for dosimetry.\(^{(12)}\) A head phantom including a polymer gel vessel at the isocentre area was irradiated using a collimated treatment field. The centre of the 50% dose level was considered as the isocentre and compared with the treatment planning system. It must be noted that the application of gel introduced errors due to uncertainties in dose determination (up to 3% for the high dose area). The noise in magnetic resonance evaluation images could affect the data particularly in the point of normalization which would deteriorate the whole profile.\(^{(12)}\) Furthermore, this method requires a long time for preparation, processing and analysis of data which makes it impractical for routine clinical application.

**Digital micrometer:** In one study, the deviation of the treatment table axis from the isocentre was measured along with the angle between the table axis and vertical direction using a two
dimensional digital micrometer.\textsuperscript{(50)} Four angles were examined and the mean axis position was calculated with a standard deviation which was attributed to the uncertainty in the alignment of lasers and the table wobble during rotation. Loading the table did not show a considerable effect on the table wobble.\textsuperscript{(50)}

III. CONCLUSIONS

The issue of isocentre verification with a high degree of spatial accuracy is extremely important in SRS/SRT treatments and requires highly efficient quality assurance scheduling. In this paper, various techniques and algorithms suggested by different groups of researchers leading to improved accuracy for verification of the mechanical linac isocentre have been summarized and discussed to make it easier for the physicists to decide which method to choose for their routine quality assurance procedure.

The problems with conventional methods of isocentre verification including the use of mechanical pointer and film-based Winston-Lutz method are described in detail in the introduction section and briefly listed in Table 1. As a result, there has been a growing interest on developing techniques to automatically investigate the stability of the isocentre position with sub-millimetre accuracy.

Several algorithms have been proposed for automatic isocentre verification that are mainly based on using EPID images of a phantom for the assessments, although there are few other less popular techniques using different methods or detectors. All EPID-based techniques have the advantages of being quick and automatic, providing digital images with no major setup required for the imaging device and producing quantitative results for the isocentre offset. The results can be used to adjust the lasers or mechanical features involved in the gantry or couch rotations. The advantages and drawbacks of each EPID-based algorithm are summarized in Table 2.
<table>
<thead>
<tr>
<th>EPID-based algorithms</th>
<th>Phantom</th>
<th>Advantages</th>
<th>Disadvantages</th>
</tr>
</thead>
<tbody>
<tr>
<td>Global thresholding technique(20,35,51)</td>
<td>W-L</td>
<td>Simple. Up to 0.3 mm accuracy.</td>
<td>Sensitive to noise. Needs further processing to achieve sub-pixel accuracy.</td>
</tr>
<tr>
<td>Double convolution method(57)</td>
<td>W-L</td>
<td>Fast. 0.1 mm accuracy.</td>
<td>Complicated method</td>
</tr>
<tr>
<td>Sobel edge detection filter + Hough transform(24,25)</td>
<td>W-L</td>
<td>0.02 mm accuracy</td>
<td>Relatively slow. Sensitive to noise and object size. Increased uncertainty due to detection of thick edges.</td>
</tr>
<tr>
<td>Segmentation + convolution kernel(41)</td>
<td>W-L</td>
<td>Simple. 0.1 mm accuracy</td>
<td>Image magnification introduces errors. Low resolution in radiation field segmentation.</td>
</tr>
<tr>
<td>Canny edge detection filter + Hough transform(31)</td>
<td>QUASAR</td>
<td>0.2 mm accuracy</td>
<td>Relatively slow. Broken edges. Sensitive to noise.</td>
</tr>
<tr>
<td>Contouring algorithm(30)</td>
<td>QUASAR</td>
<td>Sub-pixel accuracy</td>
<td>Slow. Large uncertainties.</td>
</tr>
<tr>
<td>Sobel edge detection filter + numerical optimization(27)</td>
<td>Varian</td>
<td>0.1 mm accuracy</td>
<td>Sensitive to noise and object size. Increased uncertainty due to detection of thick edges.</td>
</tr>
<tr>
<td>Edge detection filter + target centre of mass(40)</td>
<td>In-house designed</td>
<td>Sub-millimetre accuracy</td>
<td>Long processing times. Noisy results.</td>
</tr>
<tr>
<td>Signal intensity weighting + global optimization(75)</td>
<td>In-house designed</td>
<td>None</td>
<td>Complicated. Uncertainties due to source penumbra and pixel noise. 1.6 mm accuracy.</td>
</tr>
</tbody>
</table>

In conclusion, the basic methods for automatic investigation of the position of linac mechanical isocentre at few gantry and couch angles have been established. Nevertheless, there is still room for improvement of the algorithms. Future research can lead to the development of faster and more accurate methods for the determination of gantry or couch wobbles during SRS/SRT treatments.

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Verification of the linac isocentre for stereotactic radiosurgery using cine-EPID imaging and arc delivery

Abstract

Purpose: Verification of the mechanical isocentre position is required as part of comprehensive quality assurance programs for stereotactic radiosurgery/radiotherapy (SRS/SRT) treatments. Several techniques have been proposed for this purpose, but each of them has certain drawbacks. In this paper a new efficient and more comprehensive method using cine-EPID images has been introduced for automatic verification of the isocentre with sufficient accuracy for stereotactic applications.

Methods: Using a circular collimator fixed to the gantry head to define the field, EPID images of a Winston-Lutz phantom were acquired in cine-imaging mode during 360° gantry rotations. A robust MATLAB code was developed to analyse the data by finding the centre of the field and the centre of the ball bearing shadow in each image with sub-pixel accuracy. The distance between these two centres was determined for every image. The method was evaluated by comparison to results of a mechanical pointer, and also by detection of a manual shift applied to the phantom position. The repeatability and reproducibility of the method were tested and it was also applied to detect couch and collimator wobble during rotation.

Results: The accuracy of the algorithm was 0.03±0.02 mm. The repeatability was less than 3 µm and the reproducibility was less than 86 µm. The time elapsed for the analysis of more than 100 cine images of Varian aS1000 and aS500 EPIDs were ~65 and 20 seconds, respectively. Processing of images taken in integrated mode took 0.1 seconds. The output of the analysis software is printable and shows the isocentre shifts as a function of angle in both in-plane and cross-plane directions. It gives warning messages where the shifts exceed the criteria for SRS/SRT and provides useful data for the necessary adjustments in the system including bearing system and/or room lasers.

Conclusions: The comprehensive method introduced in this study uses cine-images, is highly accurate, fast and independent of the observer. It tests all gantry angles and is suitable for pretreatment QA of the isocentre for stereotactic treatments.

Keywords: isocentre verification, stereotactic radiosurgery, quality assurance, sub-millimetre accuracy
I. INTRODUCTION

In high precision dose delivery techniques such as stereotactic radiosurgery (SRS) or stereotactic radiotherapy (SRT), the accuracy of the treatment procedure is vitally important due to the delivery of high radiation doses to a localized target.\textsuperscript{1-3} Geometric errors in the system are the main causes of inaccuracies in stereotactic therapy techniques,\textsuperscript{4} which may lead to serious consequences since the majority of these treatments are applied to intracranial tumours.\textsuperscript{5} The most crucial geometric characteristic in SRS/SRT treatments is the exact position of the target relative to the linac isocentre during beam delivery.\textsuperscript{6-8} Implementation of efficient treatments requires the development of comprehensive quality assurance protocols to ensure high level geometric accuracy. These should include methods to control the isocentre position which is affected by: gantry excursions during rotation as a result of its unbalanced weight,\textsuperscript{9,10} irregularities caused by the precision bearing system used for movement control,\textsuperscript{5,9-16} and misalignment of the room lasers.\textsuperscript{11,17,18} Similarly, the rotation of the collimator and treatment table could affect the geometric accuracy. Up to ±1 mm deviation between the radiation and mechanical isocentre is agreed as the acceptable level for SRS/SRT treatments\textsuperscript{19} and the relevant parts of the linac or positioning system need to be adjusted if this limit is exceeded. The isocentre verification process must be performed before every single SRS/SRT treatment.\textsuperscript{8,20-23}

At present, a variety of methods are used for verification of the linac mechanical isocentre position. A number of these techniques were conventionally based on the application of Winston-Lutz (W-L) phantom and radiographic test films taken at a few gantry/couch angles.\textsuperscript{1,5,24-26} The main problem with these methods was their dependence on film measurements which were time consuming and highly dependent on the observer. In addition they could only provide qualitative results\textsuperscript{8,20,27,28} with relatively large uncertainties up to ±0.5 mm.\textsuperscript{11} Electronic portal imaging devices (EPIDs) have gradually replaced films and made isocentre verification much faster and easier. EPIDs require no major setup and automatically take images in digital format. Their images are quantitative and can be modified for contrast enhancement. However, there are few combinations of gantry and couch angles where EPID images can not be acquired (e.g. gantry at 90° and couch at 90°) and conventional film-based techniques may have more flexibility in this respect.\textsuperscript{20}

Different phantoms (mostly W-L) have been used for EPID-based isocentre verification at few gantry/couch setups, and various algorithms have been applied for the analysis of the results with sub-millimetric accuracy.\textsuperscript{3,4,20,29-31} In some studies, the beam was collimated by secondary or tertiary collimators and the algorithms were based on global thresholding method,\textsuperscript{8,17,32} but this technique has the disadvantage of sensitivity to noise.\textsuperscript{33} Other studies used edge detection filters (Sobel or Canny) combined with an operator to find the centres of the field and the target (Hough transform).\textsuperscript{4,31,32} These methods are also sensitive to noise, object size and image...
artifacts, and require application of filters to reduce the uncertainty.\textsuperscript{29,34-37} Segmentation of the radiation field and application of a convolution kernel was also suggested for finding the radiation field and the ball bearing centre. In this method some error is introduced due to image magnification.\textsuperscript{11} It must be noted that all studies were based on EPID images acquired at discrete gantry angles which can not completely describe the geometric status of the isocentre.\textsuperscript{38}

In this paper, a new, robust, fast, automatic and highly accurate cine-EPID-based method is introduced and tested for routine pre-treatment quality assurance of the isocentre position during arc delivery with sufficient accuracy for SRS/SRT.

II. METHODS AND MATERIALS

II.A. Experimental Setup

In this study, a W-L BrainLAB isocentre verification phantom (BrainLAB AG, Feldkirchen, Germany) including a 5 mm diameter tungsten ball embedded in a Polymethylmethacrylate (PMMA) holder and attached to a metallic supporting stand, was fixed to the treatment table by a mounting device. The position of the ball was adjustable in lateral, vertical and longitudinal directions using the mounting device knobs. The centre of the target was set at the nominal linac isocentre by aligning the marks on the PMMA holder with the room lasers while the collimator and couch were both set at zero angle. It must be noted that the room lasers have a width of \(~1\) mm and are set and routinely checked by comparison to the front pointer. A 30 mm diameter circular collimator was attached to the gantry head through the accessory tray slot. Details of the experimental setup are illustrated in Fig. 1. The resulting small field provided steep dose gradients in the penumbra region and therefore increased the dose fall-off outside the target. The collimator was firmly bolted and fixed to the gantry head so that it could not introduce any detectable sag with gantry rotation. The secondary collimators were set at $6 \times 6 \text{ cm}^2$ to avoid the radiation leakage around the drill hole collimator.

Irradiations were performed using 6 MV beams of a Varian Trilogy linear accelerator (Varian Medical Systems, Palo Alto, CA) equipped with an aS1000 EPID with a matrix size of 1024\times768 pixels and four other Varian Clinacs equipped with aS500 EPIDs which have lower resolution (384\times512 pixels). For the investigation of gantry flex, images were acquired using 360 MU irradiations at a nominal dose rate of 600 MU/min in continuous (cine) image acquisition mode with the EPID at 150 cm source-to-detector distance and 360° gantry rotations in both clockwise and counter-clockwise directions, which yielded one image per \(~3°\) rotation. The EPID was retracted at zero gantry angle before each set of measurements to ensure that the detector was in the desired location. In order to find the wobble in the isocentre position due to the treatment table and collimator rotations, images were acquired in integrated mode using
100 MU irradiations at a nominal rate of 300 MU/min with the table rotated from -90° to 90° in 20° intervals or the collimator rotated 360° in 45° intervals. The images were analysed using an in-house developed software written in MATLAB programming language.

II.B. The isocentre detection algorithm

The algorithm developed in this study dealt with the radiation field and the sphere shadow separately, and calculated the distance between their centres. In other words, it detected the distance between the nominal and actual isocentres. The procedures followed by the algorithm are explained below:

a) Each image was scanned in both cross-plane and in-plane EPID coordinates \((X_E \text{ and } Y_E)\) to pick the first and the last pixels that had a value equal to or larger than 50% of the maximum gray level. This determines an approximate field edge location. In Fig. 2, the points (a) and (b) represent the selected pixels in \(Y_E\) direction.

b) The half-way point between these two pixels in each direction was chosen as the position of the profile to be plotted in the other direction: the average in \(Y_E\) direction gave the position of the profile in \(X_E\) direction and vice versa (Fig. 2). The average of five adjacent profiles was used in each direction for better statistics.

c) On the \(X_E\) and \(Y_E\) profiles, the points at 50% of the maximum were determined (\(X_L\) and \(X_R\) in Fig. 3). This used cubic interpolation of four adjacent data points on each side of the profile as shown by the solid connecting lines in Fig. 3.
d) The average of these two locations gives the centre of the field for each direction. In Fig.
3 $X_c$ illustrates the field centre in the X direction.

e) A similar method was used for the centre of the ball shadow, but this time the "50% of
the peaks" were replaced by 50% of the depth of the valleys (shadows). In addition, only
the central sub-area of each image that contained the shadow was used (Fig. 3).

f) The distance between the centres were determined in $X_E$ and $Y_E$ directions.

g) The above processes were executed as a core function for each individual image in a
series of cine acquisitions, and the data extracted from every single image were stored
in two separate vectors, one for each direction.

h) Every data point was compared with a virtual reference point defined as the average of
the data (distance between the centres) in each direction. Such a definition for the
reference point has already been used in many of the previous studies in the
literature^{11,20,30,39}

i) The data were back-projected to the isocentre level and each image was labelled with
the corresponding gantry angle.

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**FIG. 2.** A sample EPID image of the W-L phantom: the first and last points with pixel value equal to or
larger than 50% of the maximum gray level in $Y_E$ direction are specified as points (a) and (b). The
average of these points has been used to identify where to plot the profile in $X_E$ direction.
II.C. Application of the algorithm

The analysed data were presented in a single printable page. The data analysis sheet contained plots which showed variations of isocenter position in cross-plane and in-plane directions for all gantry angles (-180 to 180) along with the maximum deviations and the absolute value of displacements in each direction (+ or -) which is also known as the offset range or end-to-end misalignment. The angles where the isocentre displacements exceeded the acceptable limits for SRS/SRT (±1 mm) were specified in separate compass diagrams for each direction in addition to a warning message. A sample output sheet is shown in Fig. 5.

The time required for the processing of more than 100 images acquired during a whole gantry rotation using a PC with 2.99 GHz CPU and 1.93 GB RAM was less than 65 seconds for aS1000 EPIDs and less than 20 seconds for aS500 EPIDs. The difference is due to their different pixel resolutions.

The method was applied 46 times on 5 Varian linacs clinically used in the radiation oncology department (one Trilogy with aS1000 EPID and 4 Clinacs with aS500 EPIDs) over a period of 8 months. Similar algorithms were applied to quantify the isocentre displacements as a result of treatment table and collimator rotations in horizontal plane.
II.D. Evaluation of the method
The method proposed in this research for 360° gantry rotations was tested by comparison to an independent method. Therefore, it was benchmarked against the widely used technique of isocentre verification with a mechanical pointer at cardinal angles, and was also compared with integrated images acquired in gantry static mode at cardinal angles. The method was also tested by application of a deliberate shift of 2 mm to the phantom position (set by a ruler) and finding the distance using the proposed algorithm. This test was carried out 5 times and the phantom was repositioned every time.

II.E. Repeatability
In order to find the inherent precision of the measurement method, the analysis tool was tested for repeatability by acquisition of successive integrated images in the same conditions without moving the W-L phantom. Five images were acquired to provide sufficient data points with statistical significance.

II.F. Reproducibility
The reproducibility of the method was investigated for both static mode and continuous acquisition during a 360° arc to find the observer-dependent precision of measurements including the phantom setup and its alignment to lasers. In both of the test series, the images (integrated for the static mode and cine images for the arc mode) were taken five times with the phantom repositioned for each measurement.

III. RESULTS
III.A. The algorithm outputs
Results of isocentre verification for one of the linacs during a 360° gantry rotation in clockwise and counter clockwise directions in both in-plane and cross-plane directions are given in Fig. 4 (table and collimator were fixed at zero). Some small differences existed between the clockwise and counter-clockwise measurements which led to root mean square deviations of 0.12 and 0.04 mm in X and Y directions (same as X_E and Y_E), respectively. Therefore, the average of the two series was used for the comparisons with standard methods. The measurement results for static mode and the mechanical pointer at cardinal angles -as an independent method- are also shown in Fig. 4. Differences between the data points at similar angles measured by each method are given for both directions which are within 0.2 mm for all coordinates.
FIG. 4. Results of the isocentre motions as a function of gantry angle using the algorithm for continuous imaging mode in clockwise and counter-clockwise directions compared with gantry static images and the mechanical pointer measurements in (a) X direction, and (b) Y direction.

A sample of the software output sheets for one of the linacs which had data points with deviations beyond the acceptance limits at some angles is shown in Fig. 5. The data points are plotted as a function of gantry angle in the graphs of the left column and those points that were out of the acceptance criteria are marked in red. In addition, some useful data are reported in the middle column including the maximum isocentre offset, the difference between the maximum deviations in + and – directions (recorded as the offset range), and the angle at which the maximum offset occurs. These provide important and practically useful information for linac
adjustments. All gantry angles with the isocentre out of range are better visualized in graphs of the right column for each coordinate.

FIG. 5. A sample output sheet for isocentre verification algorithm showing the deviations in each direction, marking the angles exceeding the acceptance limits for SRS/SRT, sending warning messages specifying the faulty angles and variation ranges in each direction. The date of the experiment is extracted from the image header.

III.B. Statistical Evaluations
The results were highly repeatable with standard deviations of 2 and 3 µm in X and Y directions. The reproducibility results had standard deviations of 86 and 42 µm in static mode and maximum standard deviations of 45 and 26 µm in arc mode in X and Y directions, respectively (Fig. 6).
FIG. 6. Results of the reproducibility test for cine images acquired during a 360° arc in both directions

Another useful statistical analysis of the results was performed by presenting the overall radial displacement of the isocentre from the reference point for a complete gantry rotation \( r = \sqrt{x^2 + y^2} \) in polar diagrams. Figure 7(a) shows the polar distribution of isocentre deviations for three experiments carried out in a single day. Results for the same type of investigation on measurements made during three successive weeks are given in Fig. 7(b). Interestingly, large differences can be seen between the distribution of discrepancies in the diagrams.
III.C. Detection of manually applied offsets

The effectiveness of the method was evaluated by using the algorithm for determination of an intentional offset of 2 mm using a ruler. The result was $2.03 \pm 0.02$ mm which is within the $\pm 0.1$ mm measurement error estimated for reading the ruler scale.

III.D. Treatment table and collimator wobble

The average of three experiments for the assessment of the shift in the treatment table rotation axis (which affects the isocentre position) for one of the tested linacs is shown in Fig. 8. It must be noted that the offset in isocentre position for conditions where both the gantry and couch were at zero angle was factored out from the series of measurements. This point was

FIG. 7. Analysis of radial isocentre deviations from the reference locus for three experiments carried out on: (a) the same day, and (b) three successive weeks. The red lines indicate the acceptance criteria.
considered as reference for the treatment table wobble measurements. The deviations were less than 0.8 mm in both directions in the horizontal plane.

![Graph showing the effect of couch rotation on the position of isocentre relative to the reference point (gantry and couch at zero angle) for one of the tested linacs.](image)

**FIG. 8.** The effect of couch rotation on the position of isocentre relative to the reference point (gantry and couch at zero angle) for one of the tested linacs. Most error bars are smaller than the symbols used for plotting, with maximum standard deviations of 19 and 17 microns in X and Y directions, respectively.

Although rotation of the circular collimator is not used in SRS/SRT treatments, the effect was checked to extend the method to quality assurance procedures for linacs used in intensity modulated radiation therapy (IMRT) or intensity modulated stereotactic radiotherapy (IMSR) treatments. The results for one of the tested linacs were less than 0.15 mm in both directions in the horizontal plane as shown in Fig. 9.

![Graph showing the effect of collimator rotation on isocentre displacement for one of the tested linacs.](image)

**FIG. 9.** The effect of collimator rotation on isocentre displacement for one of the tested linacs. The error bars are smaller than the symbols used for plotting, with maximum standard deviations of 6 and 9 microns in X and Y directions, respectively.
IV. DISCUSSION

The issue of isocentre verification with a high degree of spatial accuracy is extremely important in SRS/SRT treatments and requires highly efficient quality assurance technique. There has been a growing interest on developing techniques to automatically investigate the stability of the isocentre position with sub-millimetre accuracy. Several algorithms have been proposed that are mainly based on using EPID images of a phantom for the assessments. The EPID-based algorithm introduced in the present study, has the advantage of investigation of the isocentre position during an entire 360° rotation of the gantry using images acquired in cine-imaging mode. This is a considerable improvement over current methods as it encompasses all angles that might be used during SRS/SRT treatments. The importance of this issue can be seen in Fig. 5 where the deviations exceeded the acceptance limit in positions other than cardinal angles and could be missed if measurements were limited to these angles.

The algorithm is simple, robust and efficient, and does not require specification of the EPID resolution. It doesn’t involve image processing techniques and is therefore, not affected by the errors introduced in these processes. However, all EPID-based methods (including the present work) have a common limitation that they are unable to be used at certain combinations of gantry and couch angles. As a result, it is not possible to take the couch vertical shift into account.20

The method introduced in this study has an accuracy of 0.03±0.02 mm, reproducibility (observer-dependent precision) of up to 86 µm and repeatability (inherent precision) of up to 3 µm. Differences found between the intentional offsets and the algorithm results were within the errors in reading the mechanical scale (ruler). The polar diagrams of radial offset distributions during whole arcs showed that there are some differences between the measurements made over different courses of time, which is the result of mechanical limitations of the system and confirms that the test must be performed before every single treatment. The results of the tests can also be used for alignment of the room laser system to prevent systematic errors and is crucial for a successful treatment.

Similar algorithms were used for determination of the effect of collimator and couch rotations on the excursion of the isocentre. The offsets due to treatment table rotation depends on the design and structure of table and were relatively large for the tested linacs which was attributed to the bearing system used for couch rotation. This confirmed the report by D’souza et al. (1999).10

The time required for the experiment setup was less than 3 minutes. Image acquisition in arc mode for a 360° gantry angle took 72 seconds. The analysis of more than 100 images acquired during the gantry rotation by this algorithm takes only ~65 seconds for aS1000 and ~20
seconds for aS500 EPIDs which is a considerable improvement compared with previous algorithms. For couch rotations, 10 images were acquired within 15 minutes (to rotate the couch) that were all processed in 1.2 seconds.

In the algorithm proposed in this study, all parameters are directly determined from the images with no need for filters, convolution or magnification. The only part of the procedure that is set by the operator is the phantom setup which is quite easy; therefore, human-induced uncertainties are minimized. The output sheet provides valuable information that can be used in the repair and maintenance procedures. This was the case for two of the linacs used in this research. In one instance, the bearing in the gantry head was fixed which reduced the offset range by 1.0 mm in X direction and 0.1 mm in Y direction according to the results of the algorithm before and after the maintenance procedure. In the other case, adjustment of the room lasers led to 0.8 mm reduction in the offset range in X direction, according to the pre and post-correction test results.

The present algorithm can be extended to stereotactic micro-MLC defined fields, although the sag in the leaves should be considered as a source of error. There have been methods in the literature that use jaw/MLC/micro-MLC defined fields, but they have the issue of the field defining device sagging. This may be easily fixed for measurements at distinct angles by taking an image at the opposing angle, but it would not be as simple for cine imaging during arc. The method may also be extended to treatments which involve field-defining devices (MLC/jaws) to simultaneously characterize their displacements as a result of gantry rotation.

The method developed in this study can as well be applied to other linacs equipped with EPIDs of different resolutions. In addition, the method is independent of the manufacturer and can be used on other linac brands. It is important to note that possible shifts of the EPID detector due to gravity does not affect the results since the distance between the centres is all that matters.

V. CONCLUSION

The method developed and successfully tested in this study is simple and straightforward. It provides automatic, efficient, accurate, reproducible and repeatable results and provides isocentre deviations for all gantry angles, and is independent of the manufacturer. It is specifically designed for high precision modalities including SRS/SRT/SBRT and is also applicable to IMRT quality assurance procedures. The quality assurance test must be performed before the treatment of each patient as day to day differences have been detected on a single linac due to the effects of mechanical features. The test results are printable and the method has the potential to replace the current quality assurance procedure and enable more frequent mechanical tests on the linac.
ACKNOWLEDGEMENTS

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REFERENCES


CHAPTER 7

Detection and correction for EPID and gantry sag during arc delivery using cine EPID imaging

Pejman Rowshanfarzad, Mahsheed Sabet, Daryl J O’Connor, Peter M McCowan, Boyd MC McCurdy, Peter B Greer

Abstract

**Purpose:** Electronic portal imaging devices (EPIDs) have been studied and used for pre-treatment and in-vivo dosimetry applications for many years. The application of EPIDs for dosimetry in arc treatments requires accurate characterization of the mechanical sag of the EPID and gantry during rotation. Several studies have investigated the effects of gravity on the sag of these systems but each have limitations. In this study, an easy experiment setup and accurate algorithm have been introduced to characterize and correct for the effect of EPID and gantry sag during arc delivery.

**Methods:** Three metallic ball bearings were used as markers in the beam: two of them fixed to the gantry head and the third positioned at the isocentre. EPID images were acquired during a 360° gantry rotation in cine imaging mode. The markers were tracked in EPID images and a robust in-house developed MATLAB code was used to analyse the images and find the EPID sag in three directions as well as the EPID+gantry sag by comparison to the reference gantry zero image. The algorithm results were then tested against independent methods. The method was applied to compare the effect in clockwise and counter clockwise gantry rotations and different source-to-detector distances (SDDs). The results were monitored for one linear accelerator over a course of 15 months and six other linear-accelerators from two treatment centres were also investigated using this method. The generalized shift patterns were derived from the data and used in an image registration algorithm to correct for the effect of the mechanical sag in the system. The Gamma evaluation (3%, 3 mm) technique was used to investigate the improvement in alignment of cine EPID images of a fixed field, by comparing both individual images and the sum of images in a series with the reference gantry zero image.

**Results:** The mechanical sag during gantry rotation was dependent on the gantry angle and was larger in the in-plane direction, although the patterns were not identical for various linear-accelerators. The reproducibility of measurements was within 0.2 mm over a period of 15 months. The direction of gantry rotation and SDD did not affect the results by more than 0.3 mm. Results of independent tests agreed with the algorithm within the accuracy of the measurement tools. When comparing summed images, the percentage of points with Gamma index <1 increased from 85.4% to 94.1% after correcting for the EPID sag, and to 99.3% after correction for gantry+EPID sag.

**Conclusions:** The measurement method and algorithms introduced in this study use cine-images, are highly accurate, simple, fast and reproducible. It tests all gantry angles and provides a suitable automatic analysis and correction tool to improve EPID dosimetry and perform comprehensive linac QA for arc treatments.

**Keywords:** EPID dosimetry, gantry sag, EPID sag, linac, quality assurance
I. INTRODUCTION

During recent decades, there has been considerable improvement in radiotherapy delivery techniques which has also increased the complexity of dose delivery methods. In modern techniques such as intensity modulated arc therapy (IMAT), highly conformal dose distributions are delivered to the target volume in a short time using concurrent gantry motion, multileaf collimator motion, and dose-rate variation. As a result of the application of such complex treatment techniques, there is a growing demand for more accurate methods of treatment plan dose verification.

Although film dosimetry has conventionally been used for quality assurance of treatment plans, application of electronic portal imaging devices (EPIDs) as an alternative has become an interesting subject due to their ease of setup and ability to provide a large two-dimensional array of real-time data. During arc delivery the gantry does not follow a perfect trajectory during rotation as a result of the presence of several heavy components in its head, and mechanical imperfections in the bearing system and junctions. In addition, the EPID which is mounted on the linac by a retractable robotic supporting arm, rotates with the gantry during arc deliveries to acquire dosimetry images. But the supporting arm components are not rigidly fixed to keep the EPID in a perfectly stable position during gantry rotations. In addition to the introduction of inaccuracies to dosimetry measurements, EPID sag causes image artifacts such as spatial distortion and blurring in images. The gravity effect on the gantry and the EPID movements during treatment must be quantified to develop correction methods for accurate dosimetry results and high quality images.

Several research groups have investigated the EPID/gantry excursions during rotation. One method used for the determination of EPID sag in the imager plane perpendicular to the beam axis was to take images of a ball bearing positioned at or close to the linac isocentre at discrete angles or a large number of gantry angle settings. The algorithms developed for marker detection were not described in these studies and the EPID sag was not measured along the beam axis.

Another method investigated changes in position of the centre of a square jaw-defined field. EPID images at discrete gantry angles were generally used or in some cases cine EPID images were acquired during gantry rotations. The EPID radial flex was also measured by detection of changes in the size of the square field. A limitation of these methods is that both gantry and jaw sag affect the measured EPID motion results.

Changes in the EPID images of a grid phantom and a lead block of known dimensions were investigated as a measure of the imager sag along the beam axis. The EPID sag parallel to the beam axis was also measured by detection of changes in the size of the square field.
Some other researchers used specially designed phantoms in conjunction with complicated analysis software\textsuperscript{19-23} to characterise the source and imaging detector geometry from images acquired at discrete gantry angles. Furthermore, there is a known uncertainty in positioning of the markers in phantoms which affects the results, although a method to overcome this problem has been recently reported\textsuperscript{24}. The application of Sobel edge detection filter for marker detection can increases the uncertainties\textsuperscript{25,26}. These measurements also require the availability of special phantoms.

It must be noted that although EPID sag during arc treatments should be corrected for accurate dosimetry results, the application of EPIDs for dosimetric verification of treatment plans also requires consideration of the gantry wobble, since motion irregularities during gantry rotation are not taken into account in treatment plans and this could lead to some differences between the planning system predictions and EPID dosimetry results. Measurement of gantry sag during rotation using EPID has been described in detail by Rowshanfarzad et al. (2011a)\textsuperscript{27}. We also have recently developed a method to perform the Winston-Lutz technique for stereotactic targeting quality assurance using cine-EPID measurements\textsuperscript{28}. This method used laser alignment of the ball and therefore offsets of the ball from isocentre could affect the results.

The aim of this study is to introduce a simple, accurate, fast and automated method for detection and correction of the gantry wobble and three-dimensional EPID sag during 360$^\circ$ gantry rotations based on cine EPID imaging. The advantages of the new method are that it does not require any specifically designed phantoms; it measures the sag over the full 360 degrees of rotation using cine-EPID images; gantry and detector sag are separately quantified from the same images; and the effect of offsets of the ball-bearing from isocentre are detected and corrected. The improvement in the accuracy of dosimetry measurements with the EPID using the derived corrections are also investigated.

II. METHODS AND MATERIALS

II.A. Materials

In this study, a Varian Trilogy linear accelerator (Varian Medical Systems, Palo Alto, CA) operating in the 6 MV photon mode was used for all irradiations. Megavoltage images were acquired with an aS1000 EPID attached to the linac by an E-type supporting arm. The active area of the imager was a 40×30 cm$^2$ matrix containing 1024×768 pixels. The method developed in this study was tested on three Varian Clinacs equipped with aS500 EPIDs with detector arrays of 384×512 pixels, and three Varian Trilogy linacs with aS1000 EPIDs in a collaborating radiotherapy centre.

Images were acquired in DICOM format using 360 MU irradiations at a nominal rate of 600 MU/min in continuous (cine) image acquisition mode (7.5 frames per second, 5 frames per
image) using 18×18 cm² fields with the EPID at 150 and 110 cm source-to-detector distance (SDD) and 360° gantry rotations in both clockwise and counter-clockwise directions, which yielded one image per ~3° rotation. All images were automatically dark-field and flood-field corrected by the imaging system software. The EPID was remotely retracted at zero gantry angle using the imager control box in the treatment console before each series of acquisitions to ensure that the detector was in a reproducible location. According to IEC 61217 (2008), the zero gantry angle is defined as the angle where the x-ray source is at the top.

Three tungsten carbide ball bearings with a diameter of 4.8 mm were used as markers for the measurements. Data analysis and algorithm development were performed using MATLAB programming language and software.

II.B. Measurement Methods

II.B.1. Measurement of the EPID sag

Two tungsten carbide ball bearings (BBs) were embedded in a 2 mm thick solid water slab while positioned 6 cm apart equidistant from the central axis and shifted by 3 cm from the centre toward the gantry. The slab was rigidly secured to the gantry head (Figure 1). The third BB was attached to a plastic rod and fixed at the nominal linac isocentre using the room lasers with both the collimator and couch set at zero angles. The latter is a classic technique commonly used for isocentre verification with the well known Winston-Lutz phantom. The room lasers had a width of ~1 mm and were routinely checked by comparison to the mechanical pointer. More than 100 EPID images were acquired in cine imaging mode during a whole 360° gantry rotation. The experimental setup is schematically shown in Figure 1.

FIG. 1. Schematic illustration of the experimental setup for EPID and gantry sag measurements. The ball bearings (a, b and c) are 4.8 mm in diameter.
A sample image used for the analysis is shown in Figure 2. Briefly, the position of the centre of the third ball bearing fixed at the isocentre (c) in each image is compared with the reference image acquired at zero gantry angle to reveal the EPID sag in the imager plane. Results of this measurement are independent of gantry sag or jaw sag during rotation.

FIG. 2. A sample image acquired for the determination of EPID and gantry sag. The shadows of three markers are labelled as (a), (b) and (c). The distance between (a) and (b) is specified as \((d)\).

In addition, changes in the distance \((d)\) between images of the BB pair \((a,b)\) represent the combined effect of EPID and gantry sag along the incident beam axis. A simple geometric relation (Equation 1) can easily convert the changes in \((d)\) at each angle into changes in the SDD.

\[
\Delta SDD = SDD_0 \left( \frac{d}{d_0} - 1 \right) \tag{Equation 1}
\]

where \(SDD_0\) is the source-to-detector distance at zero gantry angle, and \(d_0\) is the distance between the BBs \((a)\) and \((b)\) at zero gantry angle. \(SDD_0\) is read out from the information in the DICOM image header.

The experiment was performed at SDD=150 and 110 cm, and also tested for clockwise and counter-clockwise gantry rotations.

**II.B.2. Measurement of the combined EPID and gantry sag**

In addition to EPID sag, it is important to characterize the gantry sag during rotation. For this purpose, the combined effect of EPID and gantry sag was also investigated. Centres of the BB pair \((a\ \text{and}\ b)\) were separately tracked in the images, and were then averaged and compared to the image at zero gantry angle. Although tracking only one of the BBs might provide adequate
data to show the combined effect of EPID and gantry sag, the results for two centres were used for higher accuracy. The experiment was performed at SDD=150 and 110 cm, and also tested for clockwise and counter-clockwise gantry rotations.

II.C. Data processing algorithms

II.C.1. Detection algorithm

In order to detect the shifts in projections of the BBs in EPID images at different angular views, a marker-tracking algorithm with sub-pixel accuracy was developed which tracked the position of the BB centres in images acquired during an entire 360° gantry rotation. The steps followed in the algorithm are explained below.

i) Each image is divided into three segments: the upper 1/3 containing images of the BB pair (a and b) which is divided into equal sections on the left and right sides; and the lower 2/3 containing the image of the BB at the isocentre (c). These three regions (shown by thick dashed lines and numbers in Figure 3) are considered as separate parts for the analysis in each projection image. The algorithm is only explained for area (1) here, as similar procedures are followed for the other two regions in each image.

ii) The area containing the BB image is cropped to a region of interest with the aim of removing the field penumbra. To specify this region, the central in-plane ($P_v$) and cross-plane ($P_u$) profiles of the whole EPID image are used. Points with gray levels higher than 95% of the difference between maximum and minimum on each profile are identified as the boundary-defining limits (points A, B, C and D in Figure 3) which define the first/last rows (A,B) or columns (C,D) specifying the regions. The region of interest for the area containing BB(a) is hatched in Figure 3.

FIG. 3. Schematic illustration of the procedure followed in the detection algorithm on one section of a sample EPID image for the measurement of EPID and gantry sag during rotation
iii) The maximum and minimum pixel values are found in the region of interest. The position of the minimum gives the pixel location of the BB centre. In-plane ($P_{v1}$) and cross-plane ($P_{u1}$) profiles passing through this point are obtained. The average of five adjacent profiles in each direction are used for better statistics. This is later used for the determination of the ball bearing centre with sub-pixel accuracy.

iv) The average of the maximum and minimum gray scale values in the region of interest gives the 50% signal intensity as specified in Figure 4. Each of the profiles obtained in step (iii) are queried, and the first and the last pixels with gray scale values equal to or less than 50% of the depth of the valley are selected. The four adjacent data points on each side of the profile (as shown in figure 4) are fitted using a cubic interpolation curve. The exact position of the 50% signal intensity is determined using this curve ($U_L$ and $U_R$), and the average of these two points gives the location of the BB centre ($U_a$) with sub-pixel accuracy. This technique is applied to profiles in both in-plane and cross-plane directions.

v) Similar procedures are followed to determine the geometric parameters for the other two BBs (b and c in figure 2).

vi) The above processes are executed as a core function for each individual image in a series of cine acquisitions, and the data extracted from every single image are automatically stored in six separate vectors, one for each marker in each of the cross-plane and in-plane directions ($u$ and $v$, respectively). Each data point is labelled with the corresponding gantry angle.

vii) As mentioned in section II.B., the algorithm uses the above data to detect the changes in location of each BB centre and the distance between centres of the BB pair (a,b) to
determine the EPID sag in three dimensions in addition to the combined effect of EPID and gantry sag during a 360° rotation.

viii) Finally, the algorithm automatically fits a curve through each series of data points (in each direction) to find the general patterns for EPID sag and the combination of EPID and gantry sag.

In addition to the above features, the algorithm has the ability to detect and correct for any possible displacements of the BB positioned at the isocentre (c in Figures 1 and 2) as a result of the room lasers misalignment or the operator error. It must be noted that shifts along the couch do not affect the results, since they are along the gantry rotation axis and therefore have no elements in the rotation plane. Misalignments in the other two directions (lateral and vertical) introduce a simple periodic function into the geometric location of the marker image on the EPID during a whole gantry rotation, since the gantry moves in a circular path. In the cross-plane direction, this function can be defined as the first harmonic (fundamental) of the Fourier series (Equation 2). The cosine part indicates the displacement in the lateral direction and the sine term gives the displacement in the vertical direction.

\[ f(\theta) = a_0 + \sum_{n=1}^{\infty} a_n \cos(n\theta) + \sum_{n=1}^{\infty} b_n \sin(n\theta) \]  

Equation (2)

where \( a \) and \( b \) are constants (Fourier coefficients), \( \theta \) is the gantry angle in radians, and \( n=1 \) (since it is the first harmonic).

The algorithm automatically fits the first harmonic of the Fourier series to the measured data points and subtracts the result from the raw data. This gives the EPID sag in the cross-plane direction. The \( a_1 \) and \( b_1 \) coefficients represent the BB displacements from the mean mechanical isocentre position in the lateral and vertical directions, respectively. These are back-projected to the isocentre level and reported as useful extra outputs of the algorithm.

II.C.2. Correction algorithm

The correction algorithm was based on an image registration technique. In general, image registration can be a geometrical transformation to align points in one view of an image with corresponding points in the reference view. In other words, the pixels in one view are mathematically mapped to pixels in the reference view. The correction algorithm developed in this study for the EPID and gantry sag during arc was a fully automatic pixel-based image registration algorithm.

To assess the corrections, a 30 mm diameter circular collimator was attached to the gantry head through the accessory tray slot. The purpose of using this collimator was to avoid any possible shifts in field-defining secondary or tertiary collimators during rotation. The drill hole
collimator was firmly bolted and fixed to the gantry head so that it could not introduce any detectable sag with gantry rotation. The jaws were set at 6×6 cm² to avoid the radiation leakage around the collimator. A series of EPID images were acquired at SDD=150 cm with no phantom in the beam. All images in the series were accumulated (Equation 3) and compared with the reference image at zero gantry angle (used as reference).

\[ I(x, y) = \sum_{i=1}^{n} I_i^\theta (x, y) \]  

(Equation 3)

where \( I(x, y) \) is the cumulative image matrix, \( n \) is the number of images in a sequence, and \( I_i^\theta (x, y) \) is the image acquired at each gantry angle \( \theta \).

The same procedure was followed after EPID and gantry sag corrections for each image in the sequence. Corrections were made according to Equations (4) and (5) for the EPID sag and the combined effect of EPID and gantry sag, respectively.

\[ I(x, y) = \sum_{i=1}^{n} I_i^\theta (x - k_i^E, y - m_i^E) \]  

(Equation 4)

\[ I(x, y) = \sum_{i=1}^{n} I_i^{EG} (x - k_i^{EG}, y - m_i^{EG}) \]  

(Equation 5)

where \( I(x, y) \) is the cumulated image matrix after correction for the EPID sag, \( I(x, y) \) is the cumulative image after correction for the combined effect of gantry and EPID sag, \( n \) is the number of images in a sequence, \( k \) and \( m \) are corrections (in pixels) in cross-plane and in-plane directions at each angle \( \theta \).

The cumulative images were normalized to the centre and quantitatively compared with the normalized reference image acquired at zero gantry angle using the Gamma evaluation method\textsuperscript{31} (3%, 3 mm criteria). In practice, for real-time dosimetry during IMAT, it is important to detect the errors and effectiveness of corrections at individual gantry angles; therefore, each of the corrected images in a sequence was also quantitatively compared with the reference image.

In addition, the Canny filter\textsuperscript{32} was used to track the edge pixels of the circular hole images during rotation and provide a visual presentation of the shifts in images resulting from the EPID and gantry sags and their corresponding corrections.

\textbf{II.D. Evaluation of the method}

Although the Gamma evaluation of the corrected images in a sequence (as explained in section II.C.2.) is the main method to assess the efficiency of the sag detection program, the algorithm was also tested using independent methods for more detailed investigations.
II.D.1. Test of the algorithm results for EPID and gantry sag

The algorithm results for EPID sag were assessed by following the position of room lasers on the EPID at cardinal angles. The lasers were first checked for alignment; then the position of lasers at each angle was recorded by marking on a sheet of millimetric paper attached to the imager. Changes relative to zero gantry angle (resulting from EPID sag) were compared with the algorithm results.

Results of the algorithm for the combined EPID and gantry sags were also tested at cardinal angles by tracking the linac crosshair on a sheet of millimetric paper attached to the EPID.

II.D.2. Test of corrections for marker misalignment at the isocentre

Laser misalignment or operator errors will result in offsets of the BB position from the isocentre. The correction method for this possible misalignment is explained in section II.C.1. In order to test the algorithm for this correction, the BB was first positioned at the nominal isocentre and a series of cine EPID images were acquired during a whole gantry rotation (reference series). Then, the BB was deliberately displaced from the nominal isocentre position by known shifts set by a vernier capable of moving in both vertical and lateral directions, and cine EPID images were again taken during a whole gantry rotation. The algorithm was used to detect the shifts by subtraction of the corresponding Fourier coefficients determined for each series. Measurements were carried out three times and the BB was repositioned for each set.

In the next experiment the BB was fixed at the nominal isocentre and a series of cine EPID images were acquired during a 360° gantry rotation. The algorithm was used to find the offsets in the lateral and vertical directions. Then the BB position was corrected according to the algorithm outputs, using a two-dimensional vernier, and another series of cine EPID images were acquired during an entire gantry rotation. No software correction was applied to the latter series of images and the results were compared with the first series (which were already corrected by the algorithm to remove the first Fourier harmonics).

III. RESULTS

III.A. The combined effect of gantry and EPID sag

Results of the algorithm for the combined effect of gantry and EPID sag as a function of gantry angle are given in Figure 5 which shows for each angle the average of positions of the BB pair (a and b in Figure 2). The detector was at SDD=150 cm and measurements were made on the same linac equipped with an aS1000 EPID over the period of 15 months. More than 80 series of measurements were performed, but only five sequences at ~3 months intervals have been presented for clarity. Each experiment was repeated three times and the maximum standard deviation was 0.09 mm in the cross-plane and 0.05 mm in the in-plane direction. The root mean
square deviation (RMSD) between the position of markers (a) and (b) were 0.01 and 0.02 mm in cross-plane and in-plane directions, respectively. Therefore, even tracking one of the BBs could provide the required data with adequate accuracy.

FIG. 5. Results of the combined effect of gantry and EPID sag as a function of gantry angle at SDD=150 cm for the same linac over a period of 15 months in (a) cross-plane and (b) in-plane directions. The fitted sag values in the in-plane and cross-plane directions are compared in (c). Most error bars are smaller than the symbols used for plotting.
Figure 5 shows that the EPID sags in both directions change with gantry angle. The results were very reproducible over the period of experiments and the maximum difference among the sets of measurements was less than 0.2 mm which is smaller than the EPID pixel size (0.392 mm). A curve was fitted through the data points in Figure 5 to generalize the changes in sag as a function of gantry angle and create the transform model for each direction. Fourier series (Equation 2) were the best fits with n=8 for the cross-plane and n=1 for the in-plane data points. The fit formula is given in the appendix. The fitted curves in the in-plane and cross-plane directions are compared in Figure 5(c) to illustrate the effect more clearly.

In order to investigate the effect of the support arm position on sag results, the experiment was repeated with the EPID at SDD=110 cm and the results were compared with SDD=150 cm. The RMSD between the two series was 0.1 mm in cross-plane and 0.2 mm in in-plane direction.

In addition, results of clockwise and counter clockwise gantry rotations were compared and the RMSD between the two series were determined as 0.3 mm and 0.2 mm in the cross-plane and in-plane directions, respectively.

### III.B. Effect of EPID sag

The outputs of the algorithm for the EPID sag in the plane of the imager (derived from the position of the BB at the isocentre (c) in Figures 1 and 2) as a function of gantry angle are given in Figure 6. Measurements were made on the same linac equipped with an aS1000 EPID over the period of 15 months. The raw data before correction for possible misplacement of the BB at the isocentre in the cross-plane direction are shown in Figure 6(a), and the corrected data points after subtraction of the Fourier first harmonic are given in Figure 6(b). The in-plane data points did not require any corrections as explained in section II.C.1. and are plotted in Figure 6(c). Each experiment was repeated three times and the maximum standard deviations were 0.05 mm in both cross-plane and in-plane directions. According to Figure 6, large differences in the amplitude and phase of various data series were observed before correction for BB misplacements, while the results became very reproducible over the period of experiments after the application of the correction method. The maximum differences among the sets of measurements was less than 0.1 mm in cross-plane and 0.2 mm in in-plane direction which are less than the EPID pixel size. A curve was fitted through the data points in Figure 6(b) and (c) to generalize the changes in EPID drift as a function of gantry angle and create the transform model for each direction. Fourier series (Equation 2) were the best fits with n=8 for the cross-plane and n=1 for the in-plane data points. The fit formula is given in the appendix. The fitted curves in the in-plane and cross-plane directions are compared in Figure 6(d) to illustrate the effect more clearly.
FIG. 6. Results of EPID sag measurements in its plane as a function of gantry angle using the algorithm for the same linac over the period of 15 months in (a) cross-plane direction before correction, (b) cross-plane direction after correction for BB misalignment at the isocentre, and (c) in-plane direction. The fitted sag values in the in-plane and cross-plane directions are compared in (d).

The results obtained in clockwise and counter clockwise gantry rotations were compared and their RMSD were 0.03 mm in cross-plane and 0.06 mm in in-plane direction.

The effect of the support arm position on EPID shift was investigated by repeating the experiment at SDD=110 cm and comparison of the results with SDD=150 cm. The RMSD between the two series was 0.02 mm in cross-plane and 0.09 mm in in-plane direction.

III.C. Changes in the SDD
Changes in the distance between source and detector along the beam axis was also given by the algorithm and the results are shown in Figure 7.

Based on Figure 7, the maximum change in the beam direction for SDD=150 cm and SDD=110 cm are 1.5 mm and 1.0 mm, respectively. Such small displacements may only cause up to about 0.1% change in image size and 0.2% change in the delivered dose according to the inverse
square law. These differences are small and would not have a major effect on the measurement results; therefore, no correction method was developed for the effect of changes in SDD during rotation.

The image acquisition took 72 seconds and processing of the 108 cine EPID images to correct for gantry and EPID sag in three directions took only ~60 seconds for aS1000 and ~40 seconds for aS500 EPIDs on a PC with 2.99 GHz CPU and 1.93 GB RAM.

**III.D. Measurements on other linacs**

In addition to the linac used for the development of the algorithm (linac#1 in Figure 8), EPID sag and the combination of EPID and gantry sag measurements were made on six other linacs: three of them (linacs #2 to 4) equipped with aS500 EPIDs at our centre, and the other three (linacs#5 to 7) with aS1000 EPIDs in a collaborating centre. The same algorithm was applied to the images and the results are compared in Figure 8. It must be noted that all EPIDs were mounted on the same type of support arm (E-arm). The curves were not identical for different linac/EPID combinations, as expected. The oldest linac (#4) had the largest differences in in-plane direction mainly due to EPID sag.
III.E. Evaluation of the method

III.E.1. Test of the algorithm results for EPID and gantry sag

Comparison of the changes in position of room lasers on the EPID surface at cardinal gantry angles (section II.D.1.) with the algorithm results showed that the EPID sag in the cross-plane direction was within the thickness of the lasers (1 mm), and in the in-plane direction the algorithm output agreed with the test method results within 0.5 mm which was less than the uncertainty of measurement tools (millimetric paper and 1 mm thick lasers).

Results of the algorithm for combined EPID and gantry sag at cardinal angles compared with the crosshair position on the EPID showed that again in the cross-plane direction the deviation was less than the crosshair thickness, but in the in-plane direction the difference between the results was within 0.2 mm which was below the error of the measurement tools.
III.E.2. Test of corrections for marker misalignment at the isocentre

The marker at the isocentre was intentionally displaced in lateral and vertical directions to different known positions and the shift values were detected by the algorithm. The results are listed in Table I.

<table>
<thead>
<tr>
<th>Shifts set by the 2D Vernier (mm)</th>
<th>Shifts detected by the algorithm (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lateral direction</td>
<td>Vertical direction</td>
</tr>
<tr>
<td>2</td>
<td>0</td>
</tr>
<tr>
<td>3</td>
<td>2</td>
</tr>
<tr>
<td>-2</td>
<td>-3</td>
</tr>
<tr>
<td>0</td>
<td>2</td>
</tr>
</tbody>
</table>

Results of the other experiment for BB positioning showed that the RMSD between image series before and after the application of algorithm corrections was 0.96 mm. When the BB was manually repositioned using the offset values suggested by the algorithm (no software corrections applied), the RMSD between this series and the software-corrected series was 0.03 mm.

III.F. Correction algorithm results

A pixel-based image registration algorithm was used to mathematically map each image to the reference and thus correct the effect of EPID and gantry sag in two separate transformation steps. Cumulative EPID images of the circular drill-hole collimator acquired in cine-imaging mode during gantry rotation were normalized to the centre and quantitatively compared with the normalized reference image acquired at zero gantry angle. The edge pixels of the cumulated images and the Gamma evaluation maps (3%, 3 mm criteria) comparing them with the reference before and after corrections are shown in Figures 9 and the Gamma values are summarized in Table II.
FIG. 9. *Left column:* The edge pixels of the sum of a sequence of EPID images acquired from a circular collimator during a whole gantry rotation: (a-1) before sag corrections, (a-2) after correction for the EPID sag, and (a-3) after correction for the combined effect of EPID and gantry sag. Images are cropped for better visibility. *Right column:* Gamma maps comparing the cumulated cine images acquired during a
whole arc delivery with the reference image taken at zero gantry angle: (b-1) before sag corrections, (b-2) after correction for the EPID sag, and (b-3) after correction for the combined effect of EPID and gantry sag.

TABLE II. The Gamma evaluation results (3%, 3 mm criteria) comparing the cumulated cine-images acquired during a whole gantry rotation with the reference image at zero gantry angle. The average of three sets of measurements (±1SD) have been used for comparison.

<table>
<thead>
<tr>
<th>Correction Type</th>
<th>Gamma&lt;1 (%)</th>
<th>Mean Gamma</th>
</tr>
</thead>
<tbody>
<tr>
<td>No correction</td>
<td>85.43±0.00</td>
<td>62.12±0.01</td>
</tr>
<tr>
<td>EPID sag correction</td>
<td>94.05±0.00</td>
<td>24.28±0.01</td>
</tr>
<tr>
<td>EPID + Gantry sag correction</td>
<td>99.27±0.00</td>
<td>13.69±0.01</td>
</tr>
</tbody>
</table>

A similar investigation was performed on each individual image acquired in cine-mode during a 360° arc delivery. Comparison of each image with the reference (at zero gantry angle) was also performed using the Gamma evaluation method (3%, 3 mm criteria) to find the errors and monitor the improvements due to corrections at each angle. The percentage of points with Gamma index less than 1 are given in Figure 10 before and after the application of sag corrections. The experiment was repeated three times. Distinct control points were also tested at 45° intervals by taking static integrated images and comparing them before and after correction with the reference image (Figure 10). The largest difference between the series results and the control points was 2.57%.

FIG. 10. Comparison of a series of cine EPID images with reference using the Gamma evaluation (3%, 3 mm criteria). The percentage of points with Gamma index <1 are shown before and after corrections for the EPID sag and the combined EPID and gantry sag (±1SD). Control points at 45° intervals are also compared with the data points in the sequence at the same angles.
IV. DISCUSSION

Using EPIDs for pre-treatment or real-time in-vivo dosimetry during complex arc treatments has become a subject of interest during the past years. The EPID and gantry are both affected by gravity during rotations due to their structural features. The resulting geometric uncertainties can affect EPID dosimetry results. Although gantry sag is present during treatments, it is not included in treatment planning system predictions; therefore, it must be taken into account for accurate pre-treatment verification of IMAT plans.

In this study, a simple measurement method is proposed to quantify the EPID and gantry shifts during IMAT treatments with no need for special phantoms or devices. A large amount of information about the system characteristics during arcs is provided from EPID images acquired with just three metallic markers in the beam. The idea of attaching BBs to the gantry head is to make the measurements independent of all other sag effects such as those of jaws or MLCs. This point has not been considered in many previous studies reported in the literature.

The algorithm developed in this study automatically analyses all cine EPID images acquired during a complete arc and calculates their geometric parameters. The algorithm is not complicated and uses the images acquired in only one sequence to yield the EPID shifts in the in-plane and cross-plane directions, the combined effect of EPID and gantry sags in the in-plane and cross-plane directions and changes in source-to-detector distance during rotation. The irregularities in gantry motion alone can easily be derived from the subtraction of the EPID sag data from the combination of EPID and gantry sag for corresponding angles. This could be a useful method for quality assurance of high precision dose delivery techniques such as stereotactic radiosurgery as well as routine linac QA. It could be used as a complementary method to the isocentre verification process introduced in a previous study; although the measurement setups, equipments and outputs of the analysis algorithms are different in the two procedures.

The algorithm is also able to detect and correct for misplacement of the BB at the isocentre caused by laser misalignment or operator error. This correction is tested for intentionally applied offsets and the average deviation of the offsets detected by the algorithm from those set by the two-dimentional vernier device at various positions was 0.03 mm which is smaller than the scale reading error (±0.05 mm). It is important to note that the operator error may be minimised by repeating the measurement, and thus the deviation reported by the algorithm can be attributed to the laser offset. There might only be a scenario in which the EPID moves smoothly with exactly the same phase and frequency of the first harmonic of the Fourier function used for correction of the BB position at the isocentre, such that the correction...
removes the EPID sag pattern in the cross-plane direction. However, this situation seems very unlikely due to the complexity of the robotic arm assembly which has many junctions and several sliding or bolted mechanical parts. The BB at isocentre was repositioned using the determined offsets from the Fourier harmonics and the measurement were repeated without the removal of the first Fourier harmonics. The result was in close agreement (RMSD=0.03 mm) with the corrected data confirming that the actual EPID sag components were not removed. Since all sag values are determined by comparison to the image acquired at zero gantry angle, it is recommended to check the levelling of the imager at this position to exclude any possible errors due to the EPID position at reference conditions.

The results were reproducible to within 0.2 mm over a period of 15 months on the same linac/gantry combination. Changes in the position of support arm and the direction of gantry rotation (clockwise or counter clockwise) did not affect the results and the maximum differences were smaller than the image pixel size in all experiments. According to Figure 8, the sag characteristics for several linacs did not follow the same pattern with varying view angles which can be attributed to differences in the bearing systems of the gantry and EPID support arm components as well as the age of the systems. However, the measurement results on all machines showed that the sag in the in-plane direction was larger than the cross-plane direction, which confirms the previous reports by other research groups15,17,18,33. Apparently, for linacs with very large turbulent EPID and gantry sag values at too many angles such that the BB shadows fall out of the field, no correction could be provided. The overall time estimate for the procedure is about 5 minutes including ~3 minutes for the setup, ~1 minute for image acquisition and ~1 minute for data processing.

The independent methods used to check the algorithms for EPID sag and the combined effect of EPID and gantry sag by following the room lasers and the linac crosshairs on the EPID at cardinal angles had an accuracy of about 1 mm which showed the sag in the in-plane direction but was not enough to reveal it in the cross-plane direction.

Results in the cross-plane direction for the combined effect of EPID and gantry sag showed a jump at zero gantry angle which was attributed to the mechanical structure of the Varian linacs. A similar effect can be seen in measurement reports by other researchers6,17.

General transformation functions were generated for the EPID sag and the combined effect of gantry wobble and EPID sag during whole gantry rotations in the in-plane and cross-plane directions using geometric parameters of the tungsten BB markers on the EPID images. Changes in SDD as a result of gantry rotation were found to be small (up to 1.5 mm) and have a negligible effect on the delivered dose in the linac under investigation. These results were in agreement with the findings of a previous study where the change in SDD was reported to be 1 mm21. However, even 10 mm change in SDD during arc may only affect the dose by 1.4% and the image
size magnification by 0.7%; therefore, the effect of changes in SDD was not corrected. It must be noted that the distance between the ball bearing pair (a and b) should not be very small since it can affect the accuracy of results (6 cm used in this work).

Sag corrections were based on image registration technique to map the images acquired at various gantry angles to the reference image. The Gamma evaluation results showed the effectiveness of corrections which increased the percentage of points with Gamma index <1 by about 8.6% correcting for the EPID sag alone, and by 14% taking the combined effect of gantry wobble and EPID sag into account. Although the Gamma index was improved, the agreement was not perfect, since the sag values found by the detection algorithm were rounded in the correction algorithm as the image registration method was pixel-based. Achieving 100% agreement requires data interpolation to increase the image resolution, but this procedure has its own drawbacks and introduces interpolation artifacts to the original matrix34,35.

Correction of the sag not only increases the accuracy of EPID dosimetry measurements during arcs, but also improves the image quality9-11. It is recommended that these shifts be monitored annually and after every adjustment or repair of the system.

V. CONCLUSION

The method proposed in this study is fast, simple, accurate and reproducible, and the corrections have been shown to be effective by several independent tests. The algorithm is not limited to treatment verification applications and can be used as a robust linac quality assurance technique to characterize and monitor the behaviour of the gantry and EPID during arc treatments. The method is general and applicable to the products of all linac manufacturers. The technique used in this study can also be extended to on-board kilovoltage imagers to improve image-guided radiation therapy procedures. Application of this method can lead to better agreement between EPID dosimetry results and treatment planning system predictions in modern radiotherapy.

ACKNOWLEDGEMENTS

This work was supported by the National Health and Medical Research Council Grant (Grant No. 569211). One of the authors (PR) gratefully acknowledges the award of the UNIPRS scholarship from the University of Newcastle.

APPENDIX: THE FIT FORMULAS FOR EPID AND GANTRY SAG

The formulas found for the best fit through EPID sag and EPID+gantry sag in the cross-plane and in-plane directions are given in Equations (6) and (7), respectively for one of the tested linacs. The sag in both directions followed Fourier series with n=8 for the cross-plane and n=1
for the in-plane. The Fourier coefficients are given in Table III for EPID sag and EPID+gantry sag in each direction.

$$\text{Sag}_{\text{Cross-plane}} = a_0 + \sum_{n=1}^{8} a_n \cos(n\theta) + \sum_{n=1}^{8} b_n \sin(n\theta)$$  \hspace{1cm} (Equation 6)

$$\text{Sag}_{\text{In-plane}} = a_0 + a_1 \cos(\theta) + b_1 \sin(\theta)$$  \hspace{1cm} (Equation 7)

It is important to note that the coefficients introduced here are just given as an example and do not necessarily apply to other linacs. Each machine should be individually tested for its own sag pattern.

TABLE III. Fourier coefficients for the fitted curves through EPID sag and EPID+gantry sag in the cross-plane and in-plane directions for one of the linacs tested in this study

<table>
<thead>
<tr>
<th>Coefficient</th>
<th>EPID+Gantry</th>
<th>EPID Sag</th>
</tr>
</thead>
<tbody>
<tr>
<td>$a_0$</td>
<td>0.035</td>
<td>0.094</td>
</tr>
<tr>
<td>$a_1$</td>
<td>-0.011</td>
<td>-0.163</td>
</tr>
<tr>
<td>$b_1$</td>
<td>-0.000</td>
<td>-0.145</td>
</tr>
<tr>
<td>$b_2$</td>
<td>0.027</td>
<td>0.044</td>
</tr>
<tr>
<td>$b_3$</td>
<td>-0.005</td>
<td>0.139</td>
</tr>
<tr>
<td>$b_4$</td>
<td>-0.035</td>
<td>-0.025</td>
</tr>
<tr>
<td>$a_4$</td>
<td>-0.018</td>
<td>0.031</td>
</tr>
<tr>
<td>$a_5$</td>
<td>-0.004</td>
<td>-0.002</td>
</tr>
<tr>
<td>$b_5$</td>
<td>-0.021</td>
<td>-0.004</td>
</tr>
<tr>
<td>$b_6$</td>
<td>0.003</td>
<td>0.000</td>
</tr>
<tr>
<td>$a_6$</td>
<td>-0.004</td>
<td>0.023</td>
</tr>
<tr>
<td>$a_7$</td>
<td>0.005</td>
<td>-0.035</td>
</tr>
<tr>
<td>$b_7$</td>
<td>0.006</td>
<td>0.003</td>
</tr>
<tr>
<td>$b_8$</td>
<td>0.006</td>
<td>-0.020</td>
</tr>
<tr>
<td>$b_9$</td>
<td>-0.004</td>
<td>-0.001</td>
</tr>
</tbody>
</table>

References:


CHAPTER 8

Investigation of the sag in linac secondary collimator and MLC carriage during arc deliveries

Pejman Rowshanfarzad, Mahsheed Sabet, Daryl J O’Connor, Peter B Greer
Abstract

In modern radiotherapy, it is vitally important to monitor the performance of all linac components including the collimation system. In this study, a simple measurement method and accurate algorithm are introduced for investigation of the secondary and tertiary collimator sag during radiotherapy arc treatments. The method is based on cine EPID images of a ball bearing marker fixed to the gantry head and determines the jaw and MLC sag in all directions relative to the reference at zero gantry angle. Analysis was performed using different field sizes and collimator angles, different linacs and different gantry rotation directions. The accuracy of the method was tested and was less than 0.02 mm. The repeatability and reproducibility of the method was 0.005 and 0.09 mm, respectively. The setup is easy and quick and the algorithm is fast and fully automatic with sub-pixel accuracy. This method is suitable to be included in the routine quality assurance of linacs to monitor the collimator system performance.

Key words: linac, jaw, MLC, sag, quality assurance
1. Introduction

The introduction of multileaf collimators (MLCs) to the structure of linear accelerators was a revolutionary improvement in radiotherapy systems. The MLCs have been used to produce irregularly shaped treatment fields to conform to the tumour outline and thus minimize the dose to healthy tissues. Therefore, rigorous quality assurance (QA) procedures for the MLCs must be included in comprehensive QA schedules to ensure the accuracy of dose delivery (Mubata et al. 1997, Budgell et al. 2000, Boyer et al. 2001, Samant et al. 2002, Ezzell et al. 2003). The QA of the collimation system is even more essential in advanced radiotherapy treatments such as arc-IMRT (RapidArc™), where doses are delivered in more complex plans with the shape of MLC, gantry speed and dose rate changing during treatment (LoSasso 2008, Oliver et al. 2010, Van Esch et al. 2011, Bakhtiari et al. 2011). Rotation of the gantry during arc deliveries can affect the position of heavy-weight components in the gantry head such as secondary collimators (jaws) and MLCs (LoSasso et al. 2008, Samant et al. 2002, Sharpe et al. 2006, Sykes et al. 2008, Ford and Lutz 2004). The MLC system performs computerized and mechanical self-check; however, it is necessary to develop independent methods to monitor its operation (Mubata et al. 1997).

It must be noted that although jaws are not commonly used to shape IMRT or RapidArc treatment fields, there are some examples of junctioned fields, particularly head and neck fields where one side of the fields are defined by jaws. In addition, for large field sizes displacement of the jaws could result in inferior target coverage (Van Esch et al. 2011). The functioning of radiotherapy equipment must remain within the internationally accepted tolerances during their lifetime. This requires QA protocols to routinely check the geometric and mechanical performance of all linac components including jaws and MLCs (Kutcher et al. 1994, Nath et al. 1994, Boyer et al. 2001, Klein et al. 2009). The acceptance limit for deviations in jaw and MLC positioning is within 1 mm (Klein et al. 2009); however, Rangel and Dunscombe have shown that the systematic error in leaf positions should be limited to 0.3 mm for acceptable clinical outcomes (Rangel and Dunscombe 2009). The conventional methods for routine jaw/MLC QA tests which are commonly used in most radiotherapy centres are based on visual inspection of light field and/or film images to check the radiation and light field coincidence (Kirby 1995, Luchka et al. 1996, Mubata et al. 1997, Boyer and Shidong 1997, Boyer et al. 2001, Ezzell et al. 2003, Prisciandaro et al. 2003, Dunscombe et al. 1999, Graves et al. 2001). These methods are not quantitative (Samant et al. 2002) and are usually limited to measurements at a single or few gantry angles. There has been one further study where phosphor plates were used to measure the MLC offsets at a number of distinct collimator and gantry angles by producing MLC-defined star patterns for each measurement setup and using a software to detect the MLC offsets relative to the collimator centre of rotation in X and Y directions (Rosca et al. 2006).
Electronic portal imaging devices (EPIDs) have been used for QA of intensity modulated radiation therapy (IMRT) treatment fields for more than a decade (James et al 2000, Vieira et al 2002, Chang et al 2004, Zeidan et al 2004, Sonke et al 2004, Parent et al 2006, Howell et al 2008, Low et al 2011, Jørgensen et al 2011). They have the advantage of being included in the structure of linacs and have the ability to acquire cine images while the gantry rotates during irradiations (McCurdy and Greer 2009). Measurements with EPIDs are less costly, more reproducible, and much faster than films and therefore if they replace films for linac QA, tests can be performed more frequently (Samant et al 2002, Baker et al 2005, Yang and Xing 2004).

A number of methods have been introduced for QA of individual MLC leaf positions using EPID images (James et al 2000, Vieira et al 2002, Yang and Xing 2004, Sonke et al 2004, Baker et al 2005, Mohammadi and Bezak 2007, Jørgensen et al 2011). These methods were either limited to static gantry angles or did not discriminate between the gantry, EPID and MLC/jaw sag where they were performed in the arc mode. However, leaf position is a function of both carriage position and leaf position within the carriage. Systematic discrepancies in MLC leaf positions may result from displacement of the whole leaf carriage and support assemblies (Ezzell et al 2003). In fact dose errors are more related to gap errors rather than individual leaf position errors (LoSasso 2008, Oliver et al 2010), such that an error of 0.5 mm in the gap size of 1.5 and 2.5 cm could introduce a dose error of about 2% in head and neck or prostate IMRT fields (LoSasso 2008) and an error of 1 mm in 1.9 and 3 cm gaps leads to around 3.5% and 5% dose errors, respectively (Oliver et al 2010).

Therefore, investigation of the gravity effect on the opposed leaf banks and jaws during arc deliveries is an important topic which may not only improve machine QA routines, but also benefit patient specific QA checks using EPIDs for arc treatments.

In a previous study, Samant et al. (2002) measured the shift in MLC leaf banks relative to the central axis using EPID images at zero gantry angle assuming their camera based EPID had a rigid geometry (Samant et al 2002). In practice, the EPID can not be considered as a mechanically stable structure during gantry rotations in arc treatments (Gao et al 2007, Ansbacher 2006, Rowshanfarzad et al. 2012). In addition, the detection precision was restricted to the size of EPID pixels (Yang and Xing 2004). Although the presence of jaw or MLC carriage sag during gantry rotation has been pointed out in previous studies, no other reports were found in the literature about their measurement.

In the present study, a fast, simple measurement method and a fully automatic algorithm are introduced to quantitatively detect the positional displacement in the secondary and tertiary collimator systems of a linear accelerator with sub-pixel accuracy. The method is based on cine EPID images acquired during entire gantry rotations, while it remains unaffected by the EPID and gantry sag.
2. Materials and methods

2.1. Materials
In this study, a Varian Trilogy linear accelerator (Varian Medical Systems, Palo Alto, CA) operating in the 6 MV photon mode was used for all irradiations. The secondary collimators in the linac head consisted of two sets of adjustable opposing tungsten alloy blocks (X₁, X₂, Y₁ and Y₂ in Figure 1) each weighing ~30 kg. Every block is separately controlled by the linac controller. In Varian linacs, jaw collimators are positioned above the MLC leaves and are mainly used to reduce the interleaf leakage and leaf transmission in IMRT/RapidArc treatments. At zero collimator angle, the X-jaws and Y-jaws move in cross-plane and in-plane directions, respectively. The two jaw pairs have different mechanisms for movement. The X-jaws slide by two parallel drive scrolls, while the Y-jaws use rack and track systems for movement (Figure 1). The track for Y jaws is slightly curved and has a small angle with the horizontal level at zero gantry angle.

The linac was equipped with a Varian Millennium MLC-120 multileaf collimator (Varian Medical System, Palo Alto, CA) which consists of two banks of 60 tungsten alloy leaves: 0.5 cm wide for the central 20 cm of the field and 1.0 cm wide for the outer 20 cm (at the isocentre), and can provide a maximum field of 40×40 cm² (Figure 1). The MLC carriage weighs ~36 kg and uses two mechanisms for its movements. Two parallel drive scrolls are used for the linear translation of the leaves and two parallel linear bearing systems hold the carriage in the track (Figure 1). Each individual leaf is driven by an electric motor.

All MLC-defined fields, used in the current study, were generated using the commercial MLC Shaper software (version 6.3, Varian Medical Systems Inc., Palo Alto, CA). The MLC leaves were reinitialized before each measurement series.

DICOM images were acquired using a Varian Portal Vision aS1000 EPID attached to the linac by an E-type supporting arm. The EPID had an active area of 40×30 cm² containing 1024×768 pixels. The method developed in this study was applied and tested on two other Varian Clinacs equipped with aS500 EPIDs with detector arrays of 384×512 pixels.

A 5 mm diameter tungsten carbide ball bearing (BB) was used as marker for the measurements. Data analysis and algorithm development were performed using MATLAB programming language and software (The Mathworks Inc., MA).

2.2. Measurement methods
Images were acquired in DICOM format using 360 MU irradiations at a nominal rate of 600 MU/min in continuous (cine) image acquisition mode at a rate of 7.5 frames per second (5 frames per image). The EPID was positioned at 150 cm source-to-detector distance (SDD).
during 360° gantry rotations, which yielded one image per ~3° rotation. Images were taken for a range of MLC or jaw-defined field sizes (5×5, 10×10, 15×15 and 18×18 cm²). In order to test the effect of direction of gantry rotation on jaw/MLC sag, images acquired in clockwise and counter-clockwise gantry rotations were compared for 10×10 cm² fields. The collimator was set to zero angle for all measurements, but the effect of collimator setting was also investigated using images of a 10×10 cm² field at 90° collimator angle to maximize the gravity effect in the other direction.

All EPID images were automatically dark-field and flood-field corrected by the imaging system software. The EPID was remotely retracted at zero gantry angle using the imager control box in the treatment console before each series of acquisitions to ensure that the detector was in a reproducible location.

The BB marker was embedded in a 2 mm thick solid water slab. The slab was rigidly secured to the gantry head. More than 100 EPID images were acquired in cine imaging mode during a whole 360° gantry rotation. The BB was used as the reference point; therefore, the measurements are independent of any possible sag in the EPID imager or gantry during rotation. This will be explained with more detail in the discussion section. The experimental setup is schematically illustrated in Figure 1. A sample image used for the analysis is shown in Figure 2.

**Figure 1.** Schematic illustration of the linac collimation system and its support assemblies. The BB is fixed to the gantry head as a reference marker for jaw and MLC sag measurements. Note the direction of axes as shown on the EPID with the X-axis from X₁ to X₂ jaw and the Y-axis from Y₂ to Y₁ jaw.
2.3. Detection algorithm

The algorithm developed in this study determines the centre of the BB shadow in each EPID image and considers it as reference. The distance between the field edges and this reference point is detected with sub-pixel accuracy in all images acquired during the 360° arc delivery. The basis of the algorithm has already been explained in a previous study on isocentre verification using EPIDs (Rowshanfarzad et al 2011), but modifications have been made to the code according to the requirements of the present work. The following procedure is performed for all images in a series:

i) Orthogonal profiles are plotted through the centre of the EPID image regardless of the BB position ($P_{X1}$ and $P_{Y1}$ in Figure 2). Each profile is the average of five adjacent profiles to provide better statistics. The in-plane profile ($P_{Y1}$) is slightly shifted from the image centre to avoid the area of leakage between the rounded MLC leaf tips.

![Figure 2](image)

**Figure 2.** A sample aSi1000 EPID image used to detect the sag in MLC carriage during an arc delivery. The analysis method is schematically illustrated. The $P_{X1}$ profile has been intentionally shifted up in the figure for better visibility of the labels and clarity of the procedure.

ii) The average of the maximum and minimum grey scale values in each profile is determined as the 50% signal intensity level. The first and last points with grey scale values higher than this level in each profile give the approximate field edge locations (points 3 on either side of the profile in Figure 3).
iii) A curve is fitted through four adjacent data points around the approximate field edges in each profile (Figure 3) using cubic interpolation. The intersections of this curve with the 50% signal level specify the exact positions of field edges (points $b$ and $c$ in Figure 3).

iv) Points with grey scale levels higher than 95% of the difference between maximum and minimum on each profile are identified as the boundary-defining limits (points $A$, $B$, $C$ and $D$ in Figure 2) to specify the region of interest for BB detection (the hatched area in Figure 2). This step is required to avoid the penumbra region.

v) The position of the minimum value in the region of interest approximately gives the BB location. Orthogonal profiles (average of five adjacent profiles) are passed through this point ($P_{x2}$ and $P_{y2}$ in Figure 2) and a similar method as used in steps (ii) and (iii) is applied to determine the exact central position of the BB shadow (point $a$ in Figure 3).

vi) The distance between the four field edges and the BB centre is tracked in all images in a series. The results are compared with the reference image acquired at zero gantry angle.

vii) The code output is four separate vectors which show the sag for each field-defining component (jaw or MLC) back-projected to the isocentre level and expressed in mm. Each data point is labelled with its corresponding gantry angle. It must be noted that the BB can be anywhere in the field not necessarily at the field centre. The algorithm reads the SDD and EPID pixel resolution from the DICOM image headers; therefore, it can be used for different measurement setups or EPID types.

Figure 3. A sample $P_{x1}$ profile for the determination of field edges and the BB centre
2.4. Evaluation of the method
In order to test the accuracy of the algorithm results, the code was used to detect the manually applied offsets to radiation fields using integrated EPID images acquired at zero gantry angle. For a 10×10 cm² jaw-defined field, offsets of 2, -1, 1 and -2 mm were applied to X₁, X₂, Y₁ and Y₂ jaws, respectively. For a 10×10 cm² MLC-defined field, the offsets were 2 and -1 mm to the left and right banks, respectively. The jaws/MLC leaves were repositioned (parked and set to the field size again) five times for each setup. The amounts of shifts were detected relative to the reference 10×10 cm² field.

2.5. Repeatability
The algorithm was applied to five successive integrated EPID images of a 10×10 cm² field to test the repeatability of the method. The experiment was performed on both jaw-defined and MLC-defined fields.

2.6. Reproducibility
The method was tested for short term reproducibility using integrated EPID images acquired at zero gantry angle, and also cine EPID images acquired during a 360° arc. The 10×10 cm² field (both jaw and MLC-defined fields) was reset before each measurement. Long term reproducibility was also investigated for all field sizes during 6 months using the same linac and setup.

3. Results
3.1. MLC carriage sag
The algorithm results for quantification of the sag in MLC carriage during a 360° arc are shown in Figure 4 for different field sizes. Directions are specified as L: left, R: right, G: gun and T: target sides (Figure 4(a)). The maximum detected sag was 0.33, 0.31, 0.09 and 0.07 mm in L, R, G and T directions, respectively.
Figure 4. Comparison of the sag in MLC carriage during an entire gantry rotation using different field sizes at zero collimator angle in: (a) left, (b) right, (c) gun and (d) target directions. Directions are shown in (a).

The root mean square deviation (RMSD) between the detected MLC carriage sag in clockwise and counter-clockwise gantry rotations using a 10×10 cm² field was 133, 217, 60 and 12 µm in L, R, G and T directions, respectively.

The MLC carriage sag results are compared with those of two other linacs for a 15×15 cm² field in Figure 5. The overall trends are similar for all machines although they have been in service for different durations of time.
Figure 5. Comparison of the sag in MLC carriage during gantry rotation for three linacs using a $15 \times 15$ cm$^2$ field in: (a) left, (b) right, (c) gun and (d) target directions.

The effect of collimator angle on MLC carriage sag is shown in Figure 6. The rotation has affected the results for R and L directions as expected, since the rotation has resulted in the movement of the flat side of the MLC leaves to the cross-plane direction. Note the change in MLC positions at each direction as shown in Figure 6(a). The change in collimator angle did not have a major effect in the G-T direction since it is perpendicular to the axis of gravity and therefore has no component along it.
Figure 6. Comparison of the sag in MLC carriage for a 10×10 cm² field with 90° collimator rotation in: (a) left, (b) right, (c) gun and (d) target directions. The change in MLC direction due to collimator rotation is shown in (a).

The RMSD between the data sets at 0° and 90° collimator angles was 0.14, 0.13, 0.02 and 0.02 mm in L, R, G and T directions, respectively.

3.2. Jaw sag
The algorithm results for quantification of the jaw sag during a 360° arc are shown in Figure 7 for different field sizes. The maximum detected sag was 0.37, 0.52, 0.69 and 0.57 mm for X₁, X₂, Y₂ and Y₁ jaws, respectively.
Figure 7. Comparison of the sag during an entire gantry rotation using different jaw-defined field sizes at zero collimator angle for: (a) X₁, (b) X₂, (c) Y₂ and (d) Y₁ jaws. Jaw positions are shown in (a).

The root mean square deviation (RMSD) between the detected jaw sag in clockwise and counter-clockwise gantry rotations using a 10×10 cm² field were 27, 43, 53, 50 µm in L, R, G and T directions, respectively.

The jaw sag results are compared with those of two other linacs for a 15×15 cm² field in Figure 8. Each machine has a different jaw sag pattern since small differences in the mechanical drive system as a result of aging, wearing, etc. could affect the movements of these heavy weight components.
Figure 8. Comparison of the jaw sag during gantry rotation for three linacs using a 15×15 cm² field for:
(a) X₁, (b) X₂, (c) Y₂ and (d) Y₁ jaws.

The effect of collimator angle on jaw sag is shown in Figure 9. The rotation has affected the results for all directions as expected, since the X and Y jaws have replaced each other as a result of collimator rotation. Note the change in jaw positions at each direction as shown in Figure 9(a). The reason for change in all directions is that unlike MLCs which can not move in in-plane direction, jaws have no rigid side and can move in both in-plane and cross-plane directions.
The RMSD between the data sets at 0° and 90° collimator angles was 0.39, 0.34, 0.11 and 0.08 mm in L, R, G and T directions respectively.

It is noteworthy that the acquisition of cine images took 72 seconds and running the algorithm for more than 100 images took only about ~5 seconds for aS1000 and ~4 seconds for aS500 EPID images on a PC with 2.99 GHz CPU and 1.93 GB RAM. The run-time was the same for jaw and MLC-defined fields.

3.3. Accuracy of the method

Results of the algorithm accuracy tests are listed in Table 1. The average difference between the values of intentionally applied shifts and the algorithm predictions was ~0.02 mm.
Table 1. Results of the algorithm accuracy test for jaw and MLC-defined fields

<table>
<thead>
<tr>
<th></th>
<th>Jaw Defined Field</th>
<th>MLC-Defined Field</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>X₁</td>
<td>Y₁</td>
</tr>
<tr>
<td>Nominal Applied Shift</td>
<td>2.00</td>
<td>1.00</td>
</tr>
<tr>
<td>Algorithm Result (mm)</td>
<td>2.02±0.00</td>
<td>1.00±0.01</td>
</tr>
</tbody>
</table>

3.4. Repeatability

The standard deviation of the code results for five successive measurements on jaw-defined and MLC-defined fields in four directions were less than 5 and less than 4 microns, respectively.

3.5. Reproducibility

The short term reproducibility results for 10×10 cm² jaw and MLC-defined fields are summarized in Table 2.

Table 2. Results of the algorithm for short term reproducibility test in jaw and MLC-defined fields

<table>
<thead>
<tr>
<th></th>
<th>Static Mode, Integrated Image</th>
<th>Arc Mode, Cine Images</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Jaw-Defined</td>
<td>MLC-Defined</td>
</tr>
<tr>
<td>Maximum SD in All Directions(μm)</td>
<td>63</td>
<td>62</td>
</tr>
</tbody>
</table>

The long term stability was investigated by determination of the RMSD for all jaw and MLC-defined field sizes (5×5, 10×10, 15×15 and 18×18 cm²) during a 6 month period. The maximum RMSD values were 70 and 25 microns for jaw and MLC-defined fields, respectively.

4. Discussion

The accuracy of complicated radiotherapy treatments is affected by the behaviour of linac components. This implies the importance of QA techniques to specifically characterize all linac features particularly field shaping elements. Although the sag of jaw and MLC carriage during arc treatments can affect the accuracy of dose delivery (Ezzell et al 2003, Van Esch et al 2011, Mu et al 2008), no report was found in the literature to comprehensively analyse this effect.

In the present study, a simple and quick measurement method is used to investigate the jaw and MLC carriage sag during arc treatment deliveries. The experiment setup takes less than a minute and does not require any special phantom design.

The method usually suggested for determination of the collimator position using EPIDs is through averaging their position in images acquired at opposing collimator settings to find the field centre (Sharpe et al 2006, Korevaar et al 2011) which is set as reference. This method introduces up to 0.5 mm inaccuracy due to mismatch of the mechanical and radiation centres.
which results in systematic errors in collimator position (Baker et al. 2005). This process is not feasible for investigation of cine EPID images during arc delivery. In this study, a high density ball bearing marker is used as reference. It is independent of the collimator position and provides better accuracy. In addition, attaching the BB to the gantry head makes the method independent of all other mechanical imperfections in the system such as gantry sag. According to a previous comprehensive study on EPID sag during arc delivery (Rowshanfarzad et al. 2012), the EPID could shift in the beam direction by a maximum of 1.5 mm, which can only change the measured distance between the field edge and the BB centre by 0.1%. Even assuming an unlikely extreme case of 10 mm EPID shift in the beam direction, the distance between the field edge and the BB centre would only change by 0.7%. The EPID also moves in the in-plane and cross-plane directions, but since the distance between the BB centre and the field edges are measured for each individual image, it is independent of the BB position in the previous or next images. Therefore, the method is not dependent on the EPID sag.

An algorithm is developed in this study to quantify the sag values for each jaw or MLC bank in a 360° gantry rotation using cine EPID images acquired during the arc. The algorithm provides automatic data on individual jaw/MLC sag at different gantry angles compared to their position at zero angle (reference) with sub-pixel accuracy.

Results for the MLC carriage sag showed that the effect was similar for different field sizes and was larger in the L-R direction than the G-T (Figure 4), due to the fact that the G-T direction is perpendicular to the gantry rotation plane as well as the axis of gravity. Therefore, it is not subject to changes in the gravitational component during rotation. Figure 10 illustrates the changes in the sag of the left MLC bank at various gantry angles due to variations in the gravitational component during arc. The BB position remains unchanged since it is fixed to the gantry head. Each arrow represents a measure of the amount and direction of the left MLC bank sag at the corresponding gantry angle. At -90°, the distance between the BB and left MLC bank is at its maximum since the gravity force is strongest on the left bank enforcing it to move to its maximum possible shift. The distance gradually decreases and reaches its minimum at 90° gantry angle. The pattern at the centre of the figure shows this effect and is similar to the graph in Figure 4(a). Obviously, the pattern will be in the opposite direction for the right bank and will be similar to the effect shown in Figure 4(b).
Although comparison of the sag pattern for MLC carriage on the tested linacs did not reveal major differences (Figure 5), the measurement should be performed on each machine separately. The MLC carriage sag is affected by the collimator angle as shown in Figure 6. At 90° collimator angle, a sudden change of about 0.1 mm is visible in the L-R direction near zero gantry angle. This effect is attributed to the change in direction of gravitational component of the heavy MLC carriage that mainly affects the collimator bearing system (which is not firmly fixed). The direction of gantry rotation did not affect the MLC sag by more than 0.2 mm.

Another outcome of this study is that it proves the MLC control points which monitor and correct the position of MLC leaves are unable to detect the shifts in MLC position resulted by the sag in MLC carriage. The reason is that the whole MLC frame -including the leaf position encoders- is moving. It is similar to a moving coordinate system encompassing a stationary observer (moving with the system) that can not detect the motion of his own reference frame.

The jaw sag effect is quite different from and generally larger than the MLC sag. Jaw sag can be affected by field size such that its variation for different field sizes amounted up to 0.3 mm for the linac discussed in Figure 7. This effect is attributed to the differences in individual driving systems for each jaw. It must be noted that field size dependence patterns were not similar for different machines, for example the linac used for Figure 7 showed different field size patterns only for the X₁ and Y₂ jaws, while another linac showed differences in X₁, X₂ and Y₁ jaws. It must be noted that unlike the MLC, jaws showed large sag in in-plane direction during gantry rotation. This movement is attributed to the oblique track of the Y jaws (Figure 1) which gives them a gravitational component that drives them either toward or away from one another (Figure 7(c,d)).

The jaw sag patterns can be completely different on various machines (Figure 8) mainly due to differences in the age of the linacs and the mechanical performance of their junctions and
support assemblies. Changes in collimator angle strongly affected the jaw sag results in L-R direction (Figure 9) due to the effect of gravity on the swapped jaws.

It is important to note that although the differences observed between the expected and measured positions of field defining elements may seem small, their effect may be significant in a number of clinical situations. If the sag values are large, re-adjustment or change of the support systems is recommended.

The method proposed in this study has an accuracy of less than 0.02 mm and is highly repeatable (0.005 mm) and reproducible (0.09 mm). The algorithm is not limited to a particular linac brand and can be applied to the linacs of all other manufacturers.

5. Conclusion

In this work a novel EPID-based technique has been developed to check the positional displacement of secondary and tertiary collimators during arc radiotherapy treatments. The presented method is simple, accurate, robust, and suitable for routine QA of the collimation system performance for every linac make and model.

Acknowledgments

This work was supported by the National Health and Medical Research Council Grant (Grant No. 569211). The authors would like to thank the Electronics group at the radiation oncology department of Calvary Mater Newcastle Hospital for their support. The first author gratefully acknowledges the award of the UNIPRS scholarship from the University of Newcastle, Australia.

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Gantry angle determination during arc-IMRT: Evaluation of a simple EPID-based technique and two commercial inclinometers

Pejman Rowshanfarzad, Mahsheed Sabet, Daryl J O’Connor, Peter M McCowan, Boyd MC McCurdy, Peter B Greer

Abstract:

The increasing popularity of Intensity Modulated Arc Therapy (IMAT) treatments requires specifically designed linac quality assurance (QA) programs. Gantry angle is one of the parameters that have major effect on the outcome of IMAT treatments since dose reconstruction for patient-specific QA relies on the gantry angle; therefore, it is essential to ensure its accuracy for correct delivery of the prescribed dose.

In this study, a simple measurement method and algorithm are presented for QA of gantry angle measurements based on integrated EPID images acquired at distinct gantry angles and cine EPID images during an entire 360° arc. A comprehensive study was carried out to evaluate this method as well as two commercially available inclinometers (NG360 and IBA GAS supplied in conjunction with popular array dosimeters: Delta4 and MatriXX Evolution, respectively) by comparison of their simultaneous angle measurement results with the linac potentiometer readouts at five gantry speeds.

In all tested measurement systems, the average differences with the reference angle data were less than 0.3° in static mode. In arc mode, at all tested gantry speeds the average difference was less than 0.1° for the IBA GAS and the proposed EPID-based method, and 0.6 for the NG360 after correction for the inherent systematic time delay of the inclinometer. The gantry rotation speed measured by the three independent systems had an average deviation of about 0.01 degrees/s from the nominal gantry speed.

Keywords: gantry angle, arc IMRT, quality assurance, inclinometer
I. INTRODUCTION

Intensity modulated arc therapy (IMAT) is a novel form of radiotherapy treatment that allows the radiation dose to be delivered in one or two arcs.\(^{(1,2)}\)

This technique offers precise target coverage using lower target doses and shorter delivery times compared with intensity modulated radiotherapy (IMRT). The method is more complex than IMRT since the gantry speed, dose rate and the MLC-defined field shape are varied during the delivery.\(^{(1,3,4)}\) Therefore, the QA programs developed for IMRT do not sufficiently address the requirements for IMAT.\(^{(5)}\) Due to the increasing worldwide interest in this technique, it is essential to develop more sophisticated QA programs that take all components which affect the accuracy of IMAT delivery into account.\(^{(2)}\)

One of the major considerations for implementation of new treatment techniques is verification of the predicted doses. In the case of IMAT treatments, this involves gantry angle-resolved dosimetric information.\(^{(5-8)}\) Misalignment of the linac angular settings could severely affect the dose distribution of an IMAT plan delivery and have serious clinical consequences due to the steep dose gradients and complex MLC shapes.\(^{(9)}\)

In routine QA of linacs, a level is positioned on a flat surface of the gantry head close to the graticule and the gantry is rotated until the bubble settles at the centre. Using this method, the gantry angle indicator can be checked only for cardinal angles and the flatness of the surface usually remains unchecked.\(^{(9,10)}\) Another method suggested for QA of the angle indicator is to perform a star shot on film at distinct static gantry angles and determine the angle based on the film setup.\(^{(9)}\) This method is not suitable for testing in arc mode and introduces the difficulties of film measurements and processing. A ±1 degree limit has been recommended as the action level for the gantry angle readout system by the AAPM Task Group 40.\(^{(11)}\)

In a study on linac gantry angles during arc treatments, cine images were acquired during IMAT deliveries using an electronic portal imaging device (EPID) with a specially designed phantom in the beam. The phantom consisted of a pair of wires wound around a cylinder. The accuracy of gantry angles recorded in the header of these DICOM images were investigated by comparison to the angles derived by following the points of intersection of wires in each image.\(^{(12,13)}\) It was found that the same angle may be repeated in headers of successive images due to the low frequency of angle readout update.

Other proposed methods for the determination of gantry angle were based on EPID images of phantoms containing a number of radio-opaque fiducial markers. Edge detection filters or thresholding methods were used to detect the marker edges and numerical optimization functions were applied to find the center of each ball bearing. Gantry angles were derived from the relative positions of the markers in the images.\(^{(14,15)}\)
Due to the importance of dose verification especially in arc treatments, in the present study three independent measurement systems have simultaneously been used for gantry angle determination. A simple and easy-to-use phantom is suggested and a fast accurate algorithm is used to determine the gantry angle during a 360° arc using cine EPID images. The measurement results are evaluated by comparison to the linac log files used as reference. Furthermore, the accuracies of two types of inclinometers supplied with commercial array dosimeters which are commonly used for pre-treatment verification of IMAT plans have been investigated.

II. METHODS

A Varian Trilogy linear accelerator (Varian Medical Systems, Palo Alto, CA) was used for the experiments. These systems are equipped with two encoding potentiometers that replicate each other and provide signals linearly proportional to the gantry angle. The signals are sent to a digital readout system and as a result the gantry angle is demonstrated on the angle indicator (Figure 1) and the console. In addition, the gantry angles detected by the potentiometers are saved in the linac delivery log files (MLC DynaLog files, which are referred to as DynaLog files throughout the text). These files are generated by the Varian MLC control software and updated every 0.05 seconds and are only accessible after the delivery is completed. In the present study, the gantry angles saved in the DynaLog files were used as reference. The accuracy of potentiometer measurements was first evaluated by moving the gantry to cardinal angles using a spirit-level placed on a flat surface of the head and comparing to the potentiometer readouts.

FIG. 1. Schematic illustration of the experimental setup for three methods of simultaneous gantry angle measurements. The ball bearings are positioned in the beam at the isocentre level and two inclinometers are fixed to the gantry head.
A. Inclinometer measurements

Two types of inclinometers have been studied in the present work. They are supplied in conjunction with well-known commercially available array dosimeters that are commonly used for dosimetric verification of IMRT/IMAT plans:

i) Nordic Transducer NG360 (Hadsund, Denmark): This digital inclinometer is supplied with the Delta^4 dosimetry device (ScandiDos AB, Uppsala, Sweden) which contains two orthogonal matrices of diodes enclosed in a cylindrical phantom. The NG360 is a liquid capacitive based inclinometer which is attached to the gantry and has the ability of measuring the tilt angle with respect to gravity over the range of 360°. The inclinometer has 0.01° resolution and its maximum readout frequency is 1 Hz. The accuracy of its measurements is claimed as ±0.25°.(16)

The NG360 was firmly bolted to a steel frame and the frame was attached to the gantry head through an accessory tray slot such that it could not move during gantry rotation (Figure 1). It was aligned at zero gantry angle (IEC scale) before each series of measurements by comparison to the linac gantry angle indicator. The NG360 was connected to a PC through a converter and the collected data were processed by the accompanying program supplied by the vendor (GetAngle.exe).

ii) IBA Gantry Angle Sensor (IBA GAS): This inclinometer is supplied with the MatriXX Evolution dosimetry device (IBA Dosimetry GmbH, Schwarzenbruck, Germany) which includes a two-dimensional array of ionization chambers and is used for pre-treatment verification of IMRT/IMAT plans. The snapshots are recorded in movie mode with their corresponding measured angles and are thus used for three-dimensional dose calculations as well as corrections for optimization of the angular dependency of the array. The accuracy of its measurements is claimed as ±0.6°.(17)

The IBA GAS was attached to the gantry head (Figure 1) and levelled by two locking screws. It was adjusted to the linac gantry angle indicator at 0° and 90° angles and was finally aligned using the four setup LEDs on the device. The sampling frequency was set to 1 Hz.

The MatriXX was not positioned in the beam, but was connected to the GAS, power source, and a PC. Measurement results were processed using OmniPro-I'mRT v.1.7.0007 software (IBA Dosimetry GmbH, Schwarzenbruck, Germany) which was installed on the PC.

It must be noted that the data measured by both inclinometers were only accessible after the delivery was completed.
B. EPID-based angle measurement setup

Another independent measurement method used in this study was based on EPID images of a dual ball bearing (BB) phantom. Two 4.8 mm diameter tungsten carbide BBs were fixed on a thin plastic plate 14 cm apart and positioned in the beam at the isocentre level. An amorphous silicon aSi1000 EPID attached to the linac was used for image acquisition. The phantom was fixed to the top edge of the couch (toward the gantry) so that the shadow of the couch could not affect the BB images (Figure 1). Irradiations were carried out using 6 MV treatment beams at a variety of doses and delivery rates to perform the test at different gantry speeds. The EPID has an array of 1024×768 pixels in a 40×30 cm² area and produces images in DICOM format. The imager was remotely retracted at zero gantry angle using the imager control box in the treatment console before each series of acquisitions to ensure that the detector was in a reproducible location.

The centres of the BBs were automatically detected in each image using an algorithm already developed and explained in detail by Rowshanfarzad et al. (18,19,20) in the MATLAB programming language (MathWorks Inc., Natick, MA) with minor modifications. The algorithm finds the position of BBs by determining the minimum signal value in a region of interest with sub-pixel accuracy. Variation of the distance between the BBs (d) in cine images indicated the changes in gantry angle. The gantry angle for each projection image was determined using a calibration curve already derived from images acquired at distinct gantry angles (from -180° to 180° in 30° intervals) which yielded the gantry angle as a function of distance (d).

C. Measurement methods

The experiments were performed in static and arc modes. The three independent systems were simultaneously used for gantry angle measurements.

C.1 Measurements in static mode

The readouts of inclinometers and EPID-based angle measurements were first evaluated by comparison of the results for distinct static gantry angles with the linac angle indicator (which originates from the potentiometer). Measurements were made at gantry angles from -180° to 180° in 30° intervals and each series was tested three times to ensure the reproducibility. EPID images were acquired in integrated mode using 100 MU irradiations at 300 MU/min delivery rate.

C.2 Measurements during arc delivery

In order to make measurements that cover the whole range of possible gantry speeds (1.5, 2, 3, 4 and 5 degrees per second), various MUs were delivered at a nominal rate of 300 MU/min
(Table 1) during 360° arcs, while cine EPID images were acquired in service mode at an acquisition rate of 7.5 frames per second (5 frames per image). The imager and both inclinometers were started simultaneously ~6 seconds before the beam was turned on and continued to measure a few seconds after the beam was turned off. The angle data in the linac Dynalog files were used as reference for comparisons.

With the start of beam delivery, the Dynalog file begins to record the angles (with 20 Hz frequency) and the EPID starts to acquire images. To synchronize the BB phantom method with the Dynalog file, half of the time required for the acquisition of the first image (~0.36 s) was considered as the start of EPID imaging and its angle was compared with the corresponding data point in the Dynalog file. The timing for data acquisition is shown in a diagram in Figure 2. Furthermore, as the beam is turned on, the inclinometer readouts start to change with gantry rotation. The angles measured by the inclinometers were compared with those recorded in the Dynalog file at corresponding times.

![FIG. 2. Schematic diagram showing the timing for data acquisition with the measurement devices used in this study](image)

| TABLE 1. Measurement settings for beam delivery at various gantry speeds during 360° arcs |
|-----------------------------------------------|-----|-----|-----|-----|-----|
| Gantry Speed (degrees/s) | 5   | 4   | 3   | 2   | 1.5 |
| MU Setting              | 360 | 450 | 600 | 900 | 1200|
| Beam-on time (s)        | 72  | 90  | 120 | 180 | 240 |

The gantry speeds measured by the three methods were derived from the data sets \( \dot{\theta} = \frac{\Delta \theta}{\Delta t} \) and compared with the nominal gantry speeds.
III. RESULTS

A. Calibration curve for the EPID-based method

The algorithm results for the distance between the BB pair at distinct gantry angles are shown in Figure 3. The negative values indicate the change in the relative positions of the BBs in image projections during gantry rotation.

![Calibration Curve for the EPID-Based Method](image)

FIG. 3. Results of the algorithm for angle measurements at distinct gantry angles. The dotted line shows the curve fitted through the data points which was used to derive the calibration function for the EPID-based measurement method.

A curve was fitted through the data points and was used to determine the calibration function to find the angle corresponding to each BB distance for the entire 360° arc (Equation 1).

\[
\theta_d = \frac{1}{0.01746} \times \left( \sin^{-1} \left( \frac{d}{210.7} \right) - 1.571 \right),
\]

(Equation 1)

Where \((\theta_d)\) is the gantry angle in degrees and \((d)\) is the pixel distance between the BBs.

B. Static gantry measurements

The evaluation of linac potentiometer accuracy showed an average 0.02 degree difference between the angle settings using the spirit-level and the potentiometer-based angle indicator for cardinal angles.

The angles measured by both inclinometers and the EPID-based method at distinct angles were compared with the reference gantry angle indicator values which are read out from the linac potentiometer (Figure 4).
FIG. 4. Comparison of the three independent measurement systems with the reference gantry angle indicator for distinct static angles

The average differences (±1SD) for the NG360, IBA GAS and the BB phantom measurements with the reference were 0.15±0.13, -0.29±0.18 and 0.20±0.16 degrees, respectively.

C. Measurements during arc
Comparison of the experimental results during beam delivery in 360° arcs with the linac Dynalog files (used as reference) revealed some deviations in all measurements. The results for NG360 were discussed with the manufacturer and an inherent time delay of 0.56 s was reported by the company. Detailed results for each measurement system (including the corrected NG360 readings) are shown in Figure 5 for each gantry speed and the average deviation for each speed (±1SD) is given in Table 2.
FIG. 5. Deviations of the results of different angle measurement methods from the linac Dynalog files over 360° arcs in: (a) 5, (b) 4, (c) 3, (d) 2, and (e) 1.5 degrees/s nominal gantry rotation speeds.

Gantry speeds were also determined from the angle measurement data during arcs. The results are compared in Figure 6 for different nominal gantry rotation speeds. Details are given in Table 2.
Fig. 6. Measured gantry rotation speeds from the angle data sets during arc using the three simultaneous measurements for: (a) 5, (b) 4, (c) 3, (d) 2, and (e) 1.5 degrees/s nominal speeds.
TABLE 2. Average deviations of the measured gantry angles over 360° arcs rotating at different speeds compared with the linac Dynalog files, and the gantry speeds measured using the three independent measurement methods. All values are given ±1SD.

<table>
<thead>
<tr>
<th>Nominal Gantry Speed (Degrees/s)</th>
<th>Average Deviation from Reference (Degrees)</th>
<th>Measured Average Speed (Degrees/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>NG360 (corrected)</td>
<td>IBA</td>
</tr>
<tr>
<td>5.0</td>
<td>-3.11±0.37</td>
<td>-0.31±0.37</td>
</tr>
<tr>
<td>4.0</td>
<td>-2.81±0.25</td>
<td>-0.14±0.24</td>
</tr>
<tr>
<td>3.0</td>
<td>-0.87±0.19</td>
<td>0.81±0.19</td>
</tr>
<tr>
<td>2.0</td>
<td>-0.56±0.16</td>
<td>0.55±0.16</td>
</tr>
<tr>
<td>1.5</td>
<td>-0.14±0.11</td>
<td>0.70±0.11</td>
</tr>
</tbody>
</table>

IV. DISCUSSION

Implementation of new radiotherapy techniques requires accurate and efficient quality assurance procedures. Characterization of the changes in gantry angle is essential for the QA of machine performance in IMAT deliveries and also for real-time dosimetry or three dimensional dose reconstruction.

In this study, an EPID-based measurement method with a simple phantom consisting of two ball bearings was proposed in conjunction with a robust algorithm to measure the gantry angle during an entire 360° arc with sub-pixel accuracy. To achieve a highly accurate algorithm and more reliable results, it is recommended to position the BBs as far apart as possible and use most of the active length of the detector. The ball bearings should be large enough to involve sufficient number of pixels, and have high density to provide the required level of image contrast for data processing.

Results of this method were compared with two existing commercial devices currently used for dosimetry, and the simultaneous measurements made by the three independent systems were evaluated by comparison to the reference data provided by the linac potentiometer. Using the DICOM image headers as reference could be an option, but was abandoned since up to three images in a dataset had the same angle in their headers. This confirmed previous findings by Ansbacher et al. (12)

Results of angle measurements in static mode showed that the NG360 inclinometer provided the closest data to the linac potentiometer, while in arc mode it had the largest deviations from the potentiometer at all gantry speeds. This was attributed to the changes in the capacity of the liquid-based sensor system during gantry rotation in addition to a possible delay in the readout communications. After correction for the 0.56 s systematic time delay as recommended by the manufacturer, the deviation of readings with reference data became less than 1 degree for all
gantry speeds, although the correction was more effective for higher gantry speeds (4 and 5 degrees/s).

According to Figure 5, although the average IBA GAS inclinometer readings are in good agreement with the reference data, there are fluctuations in their readouts at high gantry speeds. However, the range of variations is generally within the tolerance limit for gantry angle (±1 degree). The reason for such noisy results is not clear due to the lack of information about the measurement mechanism and the sensor structure.

Comparison of the gantry rotation speed measured by the three independent methods showed that on average all systems had a deviation of up to 0.01 degrees/s from the nominal gantry speed. A jump was observed around the middle of beam delivery in Figure 6 (~zero gantry angle), which was attributed to the mechanical structure of Varian linacs. A similar effect has been reported in the literature.(19,21,22) The angle measurements with the EPID-based BB phantom method and algorithm used in this work were within 0.1 degrees of the reference data at all gantry speeds. The measured data were not noisy and required no time delay corrections. More importantly, the angle information is derived separately from the analysis for each image and there is no need for synchronization of the gantry angles and images, which makes this method superior to the indirect readouts from the potentiometer or inclinometers. However, setting up the BB phantom for pre-treatment dose verification or real-time dosimetry without modification of the MLC and jaw positions is not feasible. One suggested method would be to open a pair of the outer leaves (far from the treatment field) and set the ball bearings at the isocentre level such that their EPID images could be acquired to provide the angle data for each image independently. The dose corresponding to this part of the image could easily be excluded for dose reconstruction. However, using the inclinometers (for real-time investigations) or the linac Dynalog files for pre-treatment dose verification or real-time dosimetry may be more appropriate options.

V. CONCLUSION

The present study showed that the proposed EPID-based BB phantom and algorithm can accurately measure the gantry angles for static and arc deliveries and be used as a reliable method for gantry angle measurements. At high gantry speeds, the IBA GAS inclinometer provides noisy readings within the gantry angle tolerance limits and the NG360 inclinometer data fit within the ±1 degree tolerance levels after a time delay correction.

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CHAPTER 10

EPID-based verification of the MLC performance for dynamic IMRT and VMAT

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Abstract

**Purpose:** In advanced radiotherapy treatments such as intensity modulated radiation therapy (IMRT) and volumetric modulated arc therapy (VMAT), verification of the performance of the multi-leaf collimator (MLC) is an essential part of the linac QA program. The purpose of this study is to use the existing measurement methods for geometric QA of the MLCs and extend them to more comprehensive evaluation techniques, and to develop dedicated robust algorithms to quantitatively investigate the MLC performance in a fast, accurate and efficient manner.

**Methods:** The behaviour of leaves was investigated in the step-and-shoot mode by the analysis of integrated EPID images acquired during picket fence tests at fixed gantry angles and arc delivery. The MLC was also studied in dynamic mode by the analysis of cine EPID images of a sliding gap pattern delivered in a variety of conditions including different leaf speeds, deliveries at fixed gantry angles or in arc mode and changing the direction of leaf motion. The accuracy of the method was tested by detection of the intentionally inserted errors in the delivery patterns.

**Results:** The algorithm developed for the picket fence analysis was able to find each individual leaf position, gap width, and leaf bank skewness in addition to the deviations from expected leaf positions with respect to the beam central axis with sub-pixel accuracy. For the three tested linacs over a period of 5 months, the maximum change in the gap width was 0.5 mm, the maximum deviation from the expected leaf positions was 0.1 mm and the MLC skewness was up to 0.2°. The algorithm developed for the sliding gap analysis could determine the velocity and acceleration/deceleration of each individual leaf as well as the gap width. There was a slight decrease in the accuracy of leaf performance with increasing leaf speeds. The analysis results were presented through several graphs. The accuracy of the method was assessed as 0.01 mm for both the gap size and peak position determination.

**Conclusions:** This study provides fast, easy and accurate test methods for routine QA of the MLC performance and helps in faster troubleshooting of MLC problems in both IMRT and VMAT treatments.

Key words: linac QA, MLC performance, VMAT, IMRT, EPID
I. INTRODUCTION

In intensity modulated radiation therapy (IMRT) treatments, non-uniform dose distributions of megavoltage beams are delivered to the patient in complex shapes. Intensity modulation is performed by means of motorized multi-leaf collimators (MLCs). Therefore, it is essential for patient safety to routinely monitor the MLC performance through strict quality assurance programs to ensure the accuracy and reproducibility of leaf motion in every fraction of the treatment plan.\textsuperscript{1,2} In advanced radiotherapy treatments such as arc-IMRT (volumetric modulated arc therapy or VMAT), the QA of the collimation system may be more crucial, since doses could be delivered in more complex plans with the MLC shape, gantry speed and dose rate changing during treatment in single or multiple gantry rotations.\textsuperscript{3,4} In fact, it is rather challenging to provide appropriate techniques for the QA of linacs used for such sophisticated techniques.

Errors in MLC positioning may be either due to the inaccurate positioning of individual leaves or the result of systematic shifts in the leaf banks\textsuperscript{5} or MLC carriage.\textsuperscript{6} The accuracy of individual leaf positions may be affected by several factors including mechanical imperfections, gradual degradation of the performance of each motor,\textsuperscript{2} cable communication malfunctions,\textsuperscript{7} and loss of counts by potentiometer encoders.\textsuperscript{2} Leaves may slow down, get stuck, be misaligned, skewed or be affected by inertia or gravity. In addition, the drive support assemblies could fail or have delay in communications with the MLC controller.\textsuperscript{8,9} The MLC bank alignment during gantry rotation may be affected by gravity which causes drifts in the carriage drive and its supporting assemblies,\textsuperscript{2,6,10} since they are heavy components and their weight could affect the window widths during gantry and collimator rotations.\textsuperscript{2,6,9} Errors may also occur in the leaf calibration process\textsuperscript{2,5,11,12} which are probably the main cause for systematic errors in leaf banks.\textsuperscript{5,13}

Dosimetric consequences of systematic errors in leaf banks are reported to be proportional to the extent of errors and are more pronounced for treatments with small window openings.\textsuperscript{5,14} Every 1 mm shift of the leaf banks has been predicted to result in 2.7\% and 5.6\% change in the reference equivalent uniform dose for prostate and head and neck fields, respectively.\textsuperscript{5} In another study, Monte Carlo simulations have predicted \(~6\% dose difference as a result of 1 mm MLC systematic error in prostate step and shoot IMRT treatment plans.\textsuperscript{14} An error of 1 mm in dynamic IMRT delivery with a window of 2 cm has been reported to cause 5\% dose error.\textsuperscript{15} However, measurements of the effect of 0.5 mm systematic offset in the leaves have shown up to 12\% and 6\% dose difference for head and neck and prostate fields, respectively.\textsuperscript{16}

Budgell et al.\textsuperscript{17} have shown that accurate dose delivery for IMRT fields requires better than 1 mm accuracy in leaf positioning. The AAPM task group report 142 recommends \(\pm 1\) mm as the MLC positioning tolerance\textsuperscript{18} while the ESTRO guidelines propose \(\pm 0.5\) mm as the acceptance criterion.\textsuperscript{19}
Different devices have been used for the QA of MLC leaves for IMRT/VMAT delivery. The conventional method for two-dimensional (2D) checking of MLC leaf positions was to use film images of a dynamic MLC (DMLC) leaf pattern. Film images were either visually inspected or scanned, digitized and contrast enhanced. Visual inspection is a subjective and inaccurate method for checking the leaf pair alignment and uniformity of the gap widths. Using the scanned films is a better option but it is time-consuming and labour-intensive. Another disadvantage of films is their pixel to pixel noise. 

Electronic portal imaging devices (EPIDs) provide images in digital format which can be used to quickly provide the image data required for analysis. They are easy to use and have a similar level of sensitivity as films for MLC QA applications while their imaging speed can be adjusted to catch up with the leaf motion. Therefore, EPIDs were considered as more efficient alternatives to films, allowing the test to be performed more frequently. EPIDs have been used for the QA of MLC performance in several studies. There have also been reports on the application of 2D array detectors such as MapCheck, MatriXX and PTW-729 for this purpose. The major concern about these devices is their low spatial resolution.

Another method for the QA of MLC leaves would be to use the dynamic log files (or Dynalog files) created at the end of each IMRT delivery by the Varian MLC controller software. Dynalog files have been tested and used as reference in many studies on the leaf positioning accuracy. However, the validity of the data in Dynalog files strongly depends on the accuracy of leaf position readouts. Any possible drift in the encoder readings could be temporarily improved by the MLC re-initialization, but it must be routinely checked by independent imaging methods.

Accurate delivery of a dynamic IMRT/VMAT treatment requires not only accurate leaf positioning, but also correct leaf speeds. The leaf speed and its acceleration/deceleration should be investigated since they could affect the beam delivery and lead to artifacts in the beam intensity profile. The MLC leaf positioning error is reported to be proportional to the leaf speed. Investigation of the stability of MLC has been performed by film imaging or looking at the MLC log files at cardinal angles.

In the present study, some important MLC characteristics during IMRT/VMAT deliveries have been examined, and reliable QA techniques for MLC leaves have been implemented based on some popular test patterns that have already been accepted and used worldwide. The elements which have been quantitatively investigated in this work include: (a) the accuracy and stability of each individual MLC leaf position and their resulting gap widths; (b) the skewness of leaf banks/carriage; (c) the speed of each leaf over the whole range of its positions; and (d) the acceleration/deceleration of each individual leaf. These items have either not been studied in previous studies or not been measured with this level of accuracy. The study is based on the
acquisition of EPID images in integrated and cine imaging modes and development of robust and highly accurate codes to automatically analyse the image data. Finally, a large amount of useful information on the behaviour of MLC leaves is presented in an efficient manner. The ultimate aim of this study is to develop and implement faster and more accurate QA techniques for MLC leaves in both dynamic and arc IMRT deliveries.

II. METHODS AND MATERIALS

II.A. Materials

A Varian Trilogy linear accelerator (Varian Medical Systems, Palo Alto, CA) operating in the 6 MV photon mode was used for all irradiations. The linac was equipped with Millennium™ 120 leaf MLC, which includes two banks (A and B) each with 60 tungsten alloy round-ended leaves mounted on a carriage. The 40 central leaves in each bank are 0.5 cm thick (at the isocentre level) and are called inner leaves. The peripheral 20 leaves are 1.0 cm thick and are called outer leaves. Each leaf is equipped with a motor which is driven by the MLC controller and drives the leaf along the carriage. The leaf extension range is 14.5 cm at the isocentre level. Further leaf motions require the movement of the carriage. The maximum leaf speed is 3 cm/s.

Megavoltage images were acquired with an aS1000 EPID attached to the linac by an E-type supporting arm. The active area of the imager was a 40×30 cm² matrix containing 1024×768 pixels. The method developed in this study was tested on three other Varian Clinacs equipped with Millennium 120 leaf MLC and aS500 EPIDs with detector arrays of 384×512 pixels. EPID images were acquired in DICOM format and were automatically dark-field and flood-field corrected by the imaging system software. The gantry angles were set according to the IEC convention.° Data analysis and algorithm development were carried out using the MATLAB programming language and software (The Mathworks Inc., Natick, MA) in a PC with 3.10 GHz CPU and 4.0 GB RAM.

II.B. Measurement Methods

II.B.1. Accuracy of leaf positioning

The leaf pattern used for the investigation of geometric accuracy and the reproducibility of positioning for each individual leaf was defined by leaf pairs moving across the field forming a sliding slit which was stopped at a number of equally-spaced positions yielding narrow hot spots. This pattern was first introduced by Chui et al. in 1996° and has since been widely used and well known as the picket fence pattern. In this work, for measurements at fixed gantry angles, slits of 2×20 cm² dimensions stopping every 2 cm were used to produce 7 pickets. For measurements in arc mode, 10 pickets of 0.1×20 cm² size were formed by a sliding gap (dMLC)
stopping at 1.5 cm intervals. The test was performed in both conditions to cover the QA method for both IMRT and VMAT deliveries.

For fixed-gantry conditions, the EPID was positioned at 150 cm source to detector distance (SDD), the collimator angle was set to zero, and the jaws were parked at X=18 cm and Y=20 cm. The test with fixed gantry was repeated at cardinal angles to maximize the gravity effect at 90° and 270°.

The picket fence pattern explained above for fixed-gantry conditions was also used with the jaws parked at X=20 cm and Y=40 cm, the EPID at SDD=100 cm and the collimator angle was set to 90°. Thus, all MLC leaves were included in the test. The gantry was fixed at zero angle. EPID images were acquired in integrated mode during the delivery of 100 MU beams at a rate of 300 MU/min.

For the experiments in the arc mode, the EPID was positioned at SDD=105 cm and the jaw-defined field size was 20×20 cm². The linac was running at a nominal dose rate of 600 MU/min delivering 480 MU during a 352° gantry rotation. The test pattern for arc mode was based on the pattern introduced by Ling et al.²² It was provided by Varian Medical Systems and was available via myVarian homepage (myVarian.com). All plans were imported into the ECLIPSE™ treatment planning system, and scheduled and measured through the ARIA® information system. EPID images were saved in ARIA and exported in DICOM format. The angular dependency of MLC performance can also be checked with this test.

An algorithm was developed which provided four main outputs from the picket fence test, including: (a) the deviation of the peak position of each individual leaf pair from the average of all peak positions in the picket, which shows the error for each leaf pair; (b) the deviation of each actual peak position from its expected position (with respect to beam central axis), which reveals any systematic shift in leaf positioning; (c) the distance between opposing leaves (gap width) for each individual leaf pair; and (d) the skewness of leaf banks/carriage and differentiating it from the collimator misalignment and EPID rotation in its plane.

In order to check the effect of EPID and gantry sag on the picket fence test results during arc deliveries, cine EPID images acquired during the delivery were corrected based on the method introduced in a previous study⁴¹ using the sag pattern for the system under test. The corrected and uncorrected cine images were separately added up and their picket fence test results were compared. The effect of EPID sag in the beam direction on gap size was less than 0.003 mm and the combined effect on the gap size of EPID sag in its plane plus the gantry sag was found to be less than 0.06 mm. These values were negligible and were not corrected for in the algorithm.

Furthermore, picket fence images acquired immediately after initialization and after four hours of using the linac were compared to investigate the effectiveness of reinitializing the MLC.
II.B.2. Dynamic MLC performance

Although the leaf positioning can be investigated by the picket fence test, it cannot provide sufficient information on all aspects of the MLC system performance, since it can only give the data for certain points where the sliding gap stops. To verify the stability of leaf speed in a dynamic IMRT/VMAT delivery and the acceleration and deceleration of leaves as well as investigation of the consistency of gap width, a 0.5×20 cm² MLC-defined slit was moved across a 14×20 cm² field. The linac was running at a dose rate of 600 MU/min and both the collimators and gantry were at zero degrees for the fixed-gantry measurements. The EPID was positioned at SDD=150 cm and images were acquired in cine mode using the IAS3 software (version 7.3.15). The image acquisition time was set to 133.3 ms by capturing only one frame per image (7.5 frames per second). The slit was moved through five different patterns to comprehensively test the leaf performance step by step in various possible conditions:

(a) Unidirectional, starting from one end of the field and stopping at the other end, travelling with a constant speed with the gantry fixed at zero degree. Using different monitor units for this test led to a range of leaf speeds from 6.7 to 29.0 mm/s. The test was repeated at cardinal angles to investigate the angular dependence of the leaf speed and the gap width. In addition, it was repeated at 90° collimator angle (zero gantry angle) with a wider field to include the outer leaves and to investigate the consistency of the dynamic properties of the inner and outer leaves.

(b) Unidirectional, starting from one end of the field and stopping at the other end, travelling with a constant speed in an entire 360° gantry rotation with the collimator at zero angle. Using 360 MU irradiations led to a nominal leaf speed of 2.0 mm/s.

(c) Starting from one end of the field and stopping at the other end, returning to the start point with the same constant speed with the gantry fixed at zero degree. Application of 150 MU provided a nominal leaf speed of 19.3 mm/s.

(d) Unidirectional, starting from one end of the field and stopping at the other end, travelling in five sequences of uniform motion each with a different speed with the gantry fixed at zero degree. A range of dose fractions were used for similar distances to provide different leaf speeds for each sequence ranging from about 2 mm/s to 20 mm/s.

(e) Finally, a wider sliding gap (14×20 cm²) was used from an acceptance test plan provided by Varian (dynamic arc leaf speed test). The gap width was 14 cm and moved across a large 28 ×20 cm² field to and from one side to the other. Ten oscillations were made while the gantry rotated from 90° to 270°. Thus, an overall distance of 140 cm was
covered. The collimator was set at zero angle. Measurements were performed using 500 MU which led to a nominal leaf speed of 27.5 mm/s.

All MLC-defined fields for this study were generated using the commercial MLC Shaper software (version 6.3, Varian Medical Systems Inc., Palo Alto, CA). The speed of each individual leaf was determined in addition to the width of the sweeping gap for each test. The stability of leaf speeds in each sequence of test (d) was assessed. The last test was used to investigate the acceleration and deceleration of each leaf, since the speed gradients were larger.

The gantry and EPID sag effects were not considered for the sliding gap tests in arc mode, since cine EPID images were acquired every ~0.7 degrees and the sag effect would be negligible in such a small angle and short duration of time.

II.B.3. Data processing
II.B.3.1 Picket fence analysis algorithms

An algorithm was developed to detect the position of each individual leaf from the picket fence images. The steps followed in the algorithm are briefly explained below.

(i) The image borders were determined based on 50% of the signal on the central cross-plane and in-plane profiles (points U, D, L and R in Figure 1(a)).

(ii) A profile was plotted in the in-plane direction through the left border between the upper and lower borders (dotted line (PL) in Figure 1(a)).

(iii) A Fourier first harmonic curve was fitted through the data points of the profile (PL) to find the spatial frequency of leaf positions (k in Figure 1(b)). The spatial frequency for the outer leaves was twice that of the inner leaves and this was considered by automatic determination of the field length.

(iv) The Canny edge detection filter was applied to the image (Figure 1(c)) to roughly find the position of leaves and the area between each two pickets (sub-areas A₁ to A₆). It must be noted that different edge detection filters were tested for detection of the leaf positions and the Canny filter was found to be the optimum technique for this purpose.

(v) The mean pixel value for each sub-area was determined and their minimum was subtracted from the whole image matrix to eliminate the background and increase the image contrast (Figure 1(d)).

(vi) Using the leaf positions found in step (iii), cross-plane profiles were plotted through each leaf. The average of five cross-plane profiles was used for the analysis. Depending on the MLC pattern measured (fixed-gantry or arc), a certain number of peaks should be observed on the profiles (dotted curves in Figure 1(e)).
Based on a previously developed algorithm already explained in detail in other publications, the position of the 50% value of the intensity profile on the left and right edges of the peaks were determined with sub-pixel accuracy (Figure 1(e)).

The distance between the right and left edges of each peak (the full width at half maximum or FWHM) gives the gap widths (w in Figure 1(e)), and their mean position indicates the peak position with sub-pixel accuracy (point (P) in Figure 1(e)). The measured position of each peak was compared with its expected position to find any possible systematic shift. The pixel positions were calibrated to absolute coordinates relative to the centre of the beam on the EPID (CAX). This point which was used as reference has already been determined using a simple method explained in the literature and was read out from a text file.

Considering the SDD and EPID resolution, all results were automatically scaled to millimetre distances at the isocentre level.

A line was fitted through the peak positions along each picket. If the lines were not vertical, the skewness of the system was determined from the slope of the lines (α). To differentiate between the leaf bank skewness relative to the collimator and the effects of collimator angle misalignment and EPID rotation in its plane, the Radon transform was applied to one of the sub-areas between the pickets (e.g. A3 in Figure 1(c)) to determine the orientation of interleaf leakages (β). The resulting angle was subtracted from (α) and gives the net skewness of the leaf bank (θ). All MLC positions were subsequently corrected for the determined angle misalignment (α).

The results were automatically illustrated in five separate graphs showing:
- The gap width for each leaf pair: if the gap width violates the accepted range for tolerance (±1 mm), the corresponding position was flagged.
- The angle of skewness for each picket (θ)
- The peak positions for each leaf pair relative to their average over each picket.
- The peak positions for each leaf pair relative to their expected positions
- Three histograms showing the percentage frequency of: deviations from the average, deviations from the expected positions, and the distribution of the gap widths

The data presented in the graphs for each leaf were also automatically saved in the form of a text file in the folder which contains the DICOM images. The file is named as the date of the test and known as the picket fence log file. Having access to the log files from different dates would be useful for future reference.

It must be noted that the SDD and date of the test were read out from the DICOM image headers, and the EPID resolution (aS1000/aS500) and the collimator angle (0°/90°) were
automatically determined in the algorithm by finding the matrix size and image profile characteristics after the Canny filter is applied.

FIG. 1. Illustration of the main steps of the algorithm used for analysis of the picket fence images: (a) determination of image borders (U, D, L and R), a profile (P_L) is passed through the left border; (b) Fitting the first Fourier harmonic curve through the data points of (P_L) and finding the spatial frequency (k) of
the leaves; (c) Application of the Canny filter and determination of the edges in the image; (d) subtraction of the background signal from the image to increase the contrast; (e) A sample profile passed through one leaf in the cross-plane direction. The vertical bars are results of the Canny filter with additional offsets in both left and right directions to include the full width of the picket. They are used to define the region of interest to find the 50% of signal for each picket. The horizontal dashed lines indicate the position of the 50% signal level for each picket. The peak position (P) and the gap width (w) are shown on the magnified picket on the left.

II.B.3.2 Analysis algorithm for the sliding gap images

Another algorithm was developed to analyse each individual image acquired in cine mode during the delivery of sliding gap patterns. The steps followed in the algorithm are briefly explained below.

(i) The first three steps were similar to the algorithm for the picket fence analysis (steps (i) to (iii) in section II.B.3.1).

(ii) Using the leaf positions obtained from the Fourier fit, cross-plane profiles were plotted along the centrelines of the leaves. The average of five cross-plane profiles was used for each leaf to improve the statistics. Only one peak was expected on each profile.

(iii) Based on the method shown in Figure 1(e), the position of the left and right gap edges (and subsequently the gap width) were determined individually for each leaf. The movements of the leaves during each frame acquisition are not expected to affect the accuracy of gap width determination, since the leaf pairs travel simultaneously with the same speed in the same direction. The distances were converted into millimetres and back-projected to the isocentre level.

The above processes were executed as a core function for each individual image in a series of cine acquisitions, and the data extracted from each single image were automatically stored in separate vectors, each including the data for all leaves that form the gap. The vectors were concatenated to form matrices of data (e.g. gap widths) with rows containing the data for each leaf pair. The number of matrix rows and columns were equal to the number of leaves and the number of cine images, respectively.

(iv) A correction was applied in the code for the known effect of rounded leaf edges based on the leaf position offset file (MLCTABLE.txt) provided by Varian for geometric corrections.

(v) The number of frames and applied dose rates (300 or 600 MU/min) were readout from the DICOM header. These were used for determination of the time for acquisition of each image (t) that is equal the number of frames divided by the frame acquisition rate (which is 6.67 s⁻¹ for 300 MU/min, and 7.50 s⁻¹ for 600 MU/min dose rate).
(vi) The speed of each leaf was determined using the leaf positions in each two consecutive images and the time (t).

(vii) The acceleration/deceleration of each leaf was found in a similar manner to step (v).

(viii) The results were automatically illustrated in separate figures showing:

- The gap width for each leaf pair versus position while crossing the field; if the gap width was outside the accepted tolerance of ±1 mm from the planned width, a red bullet appeared beside the figure, otherwise the bullet was green.
- The deviation of gap widths from the average of gap widths defined by the corresponding leaf pair across the field (this quantifies the behaviour of each leaf pair separately)
- Two figures showing the leaf speeds versus position while crossing the field for each leaf bank separately; if the leaf speed was out of the accepted tolerance of ±5 mm/s from the planned speed, a red bullet appeared beside the figure, otherwise the bullet was green.
- Two figures showing the leaf acceleration/deceleration at different positions while crossing the field for each leaf bank separately

(ii) The data presented in the figures for each leaf were also automatically saved in the form of a text file in the folder which contained the DICOM images. The file was named as the date of the test and known as the sliding gap log file. Having access to the log files from different dates would be useful for future reference.

II.B.4. Accuracy of the algorithm

Intentional known errors were inserted in a number of leaf positions in the picket fence pattern by applying a range of displacements from 0.1 to 2.5 mm to the gap width (at the isocentre level), and also by changing the gap position in randomly selected leaf pairs without modification of the gap width. The ability of the algorithm to detect these errors could indicate its level of accuracy. Each measurement was repeated three times to reduce the uncertainties.

The accuracy of MLC skewness detection depends on the accuracy of the Radon transform. The latter was tested by rotating the image by different angles in a range of 0.1° to 1° at 0.1° intervals and detecting the angles by the Radon transform method.

III. RESULTS

III.A. MLC positioning

A series of graphical outputs produced by the algorithm is shown in Figure 2 for a picket fence test with intentional errors of 0.5 mm, 1.0 mm and 1.5 mm in the gap width of pickets 5, 6 and 7.
The first figure (2(a)) shows the EPID image from step (v) of section (II.B.3.1) to provide a visual illustration of the leaf positions. In this figure, the two orthogonal lines through the centre of radiation field (CAX) in the in-plane and cross-plane directions are automatically plotted to help visual detection of any systematic shifts in the leaf positions. Figure 2(b) includes a series of graphs showing the gap widths for each individual leaf pair in all pickets. The deviations of peak positions from their average and from their expected positions (with respect to beam central axis) are plotted for each leaf pair in Figures 2(c) and 2(d), respectively. The points where deviations are larger than the tolerance limit are automatically flagged by the error value and the corresponding leaf number (leaves 16 and 21 in pickets 6 and 7, respectively). In cases where errors are detected in peak positions, the flags will appear on graphs (c) and (d). If such results appear in routine tests, the corresponding leaves are considered out-of-range and investigation would be required. It should be noted that the 0.5 mm intentional error in picket 5 has not triggered a flag, since it is within the tolerance level. The total skewness of the system measured for all pickets at each leaf position is shown in Figure 2(e). The average skewing of the system and the MLC bank alone are given in the graph title. Finally, three histograms are generated in Figure 2(f) which statistically summarize the data in Figures 2(b), 2(c) and 2(d).
FIG. 2. The graphs automatically produced by the algorithm for the picket fence test analysis including 0.5, 1.0 and 1.5 mm intentional errors inserted in the gap width in leaves 26, 21 and 16 at pickets 5, 6 and 7, respectively: (a) the improved EPID image; (b) the gap width for each individual leaf at all pickets including flags for the out-of-range gap widths (leaves 21 and 16 in pickets 6 and 7); (c) deviation of peak positions from their average in each picket; (d) deviation of peak positions from their expected positions in each picket; (e) skewness of the system at each picket; and (f) histograms of the peak positions, expected peak positions and the gap width distribution.

The maximum time required to perform each measurement was less than 2 minutes and the data processing required less than 15 seconds for all tests.
The picket fence test was performed on all tested linacs for five months and the results of MLC performance investigations are summarized in Table I.

**TABLE I.** Results of the picket fence algorithm on three linacs over five months. The average of the maximum deviations measured at each session is presented ±1SD for each linac. The nominal gap width was 1 mm.

<table>
<thead>
<tr>
<th>Tested linac</th>
<th>#1</th>
<th>#2</th>
<th>#3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum gap width (mm)</td>
<td>1.52±0.14</td>
<td>1.48±0.20</td>
<td>1.23±0.15</td>
</tr>
<tr>
<td>Maximum deviation from average peak position (mm)</td>
<td>0.12±0.04</td>
<td>0.09±0.02</td>
<td>0.09±0.02</td>
</tr>
<tr>
<td>Maximum deviation from expected peak position (mm)</td>
<td>0.80±0.07</td>
<td>0.56±0.06</td>
<td>0.17±0.04</td>
</tr>
<tr>
<td>Maximum MLC skewness (degrees)</td>
<td>0.16±0.03</td>
<td>0.04±0.01</td>
<td>0.07±0.02</td>
</tr>
</tbody>
</table>

The picket fence test was repeated (three times) with the collimator at 90° to include all MLC leaves. The results showed that for the outer leaves, the average gap width was 0.01±0.00 mm larger than the average gap width of the inner leaves and the average peak position was 0.08±0.00 mm displaced from those of the inner leaves.

The picket fence test results were compared after the MLC initialization and four hours later. The results showed a shift in the peak position of up to 0.1 mm and 0.2 mm for 2% and 1% of the 420 data points, respectively. The change in the gap widths was up to 0.1 mm for 3%, 0.2 mm for 3.5%, 0.3 mm for 5% and 0.4 mm for 0.5% of the data points.

**III.B. Dynamic MLC performance**

Several measurements of the MLC speed, variations of the gap width, and leaf acceleration/deceleration were carried out using cine EPID images acquired during sliding gap deliveries. A sample of each set of the algorithms outputs is given in this section. Similar types of graphical outputs are presented in Figures 3 to 7 for a number of different measurement conditions. Figures 3 and 4 are the result sheets for a unidirectional crossing gap moving with a constant speed for both the fixed-gantry measurements (~10 mm/s), and arc mode (~2 mm/s). Another sliding gap starting from one end of the field and stopping at the other end, returning to the start point with the same constant speed is analysed in Figure 5. Results for a unidirectional gap travelling with different but uniform speeds are shown in Figure 6, and finally the analysis for an oscillating sliding gap during arc delivery is shown in Figure 7. There are four components (a) to (d) in each of these figures:

(a) shows the gap width at all different positions/gantry angles/delivery times across the field for each individual leaf pair depending on the measurement condition;
(b) illustrates the deviations of gap widths from the average of gap widths defined by the corresponding leaf pair across the field; (c) and (d) show the speeds of the left and right MLC banks at each individual leaf position/gantry angle/delivery time, respectively. Similar patterns are expected for the left and right banks.

The oscillating gap test was a suitable case to investigate the leaf acceleration/deceleration due to the hysteresis resulting from moving the leaves back and forth. Therefore, two more graphs (e) and (f) are added in Figure 7 to show the acceleration/deceleration of each leaf for different gantry angles. According to these graphs, the leaves had consistent speeds in each sequence with narrow junctions which indicate rapid changes in the leaf direction, as expected.

The presence of hot or cold spots on any of the above patterns could be the result of inconsistencies in the leaf speeds. Figures 3, 5 and 6 each represent the data from more than 4100 points, while the number of data points included in Figures 4 and 7 are 21400 and 14250, respectively.

FIG. 3. The graphical output sheet for the analysis of a unidirectional 5 mm sliding gap with a constant speed of ~10 mm/s using a fixed gantry angle: (a) the gap width pattern for different leaf positions across the field; (b) deviation of the gap widths from the average of gap widths defined by the corresponding leaf pair across the field; (c) the speed of each leaf in the left bank; and (d) the speed of each leaf in the right bank.
FIG. 4. The graphical output sheet for the analysis of a unidirectional 5 mm sliding gap with constant speed (~2 mm/s) delivered during an entire 360° arc: (a) the gap width versus gantry angle; (b) deviation of the gap widths from the average of gap widths defined by the corresponding leaf pair across the field; (c) the speed of each leaf in the left bank; and (d) the speed of each leaf in the right bank.

FIG. 5. The graphical output sheet for the analysis of a 5 mm sliding gap travelling with constant speed (~19 mm/s) across the field from one end to the other and returning to the original position using a fixed gantry angle: (a) the gap width pattern for different leaf positions across the field; (b) deviation of the gap widths from the average of gap widths defined by the corresponding leaf pair across the field; (c) the speed of each leaf in the left bank (left to right movement is considered as the positive direction); and (d) the speed of each leaf in the right bank.
FIG. 6. The graphical output sheet for the analysis of a 5 mm sliding gap travelling across the field in five sequences each with a different constant speed (20.1, 2.2, 16.7, 5.0 and 13.4 mm/s from left to right) using a fixed gantry angle: (a) the gap width pattern for different delivery times across the field; (b) deviation of the gap widths from the average of gap widths defined by the corresponding leaf pair across the field; (c) the speed of each leaf in the left bank; and (d) the speed of each leaf in the right bank.
FIG. 7. The graphical output sheet for the analysis of a 140 mm sliding gap travelling back and forth across the field in arc delivery mode: (a) the gap width pattern for different gantry angles; (b) deviation of the gap widths from the average of gap widths defined by the corresponding leaf pair across the field; (c) the speed of each leaf in the left bank (left to right movement is considered as the positive direction); (d) the speed of each leaf in the right bank; (e) the acceleration/deceleration of each leaf in the left bank; (f) the acceleration/deceleration of each leaf in the right bank.

The maximum time required to perform each measurement was less than 2 minutes and the data processing required less than 20 seconds for the tests at fixed gantry angles. The results for the arc mode including the unidirectional sliding gap and the oscillating gap test took 60 and 90 seconds to process, respectively.

III.C. Comparison of the results of DMLC tests

A unidirectional sliding gap delivered at a fixed gantry angle was tested for four different leaf speeds and the results of their statistical analysis are summarized in Table II. The change in leaf speed was achieved by applying a different number of monitor units. The average detected and nominal leaf speeds were compared for each experiment, and the maximum deviation of the detected leaf speeds (among all leaf positions) from the global average are listed for each
measurement. The root mean square deviation (RMSD) between the leaf speeds at corresponding positions in the left and right banks are calculated. In addition, the gap widths have been investigated for all leaf pairs by looking at their maximum deviation from the nominal gap width (5 mm) across the field, and by evaluation of the RMSD of the measured and nominal gap widths at all positions across the field.

TABLE II. The analysis results for testing the consistency of gap width and leaf speed for a unidirectional 5 mm sliding gap delivered at a fixed gantry angle. The average of three measurements are presented with ±1SD.

<table>
<thead>
<tr>
<th>Nominal leaf speed (mm/s)</th>
<th>6.73</th>
<th>9.67</th>
<th>19.33</th>
<th>29.01</th>
</tr>
</thead>
<tbody>
<tr>
<td>Average detected speed (mm/s)</td>
<td>6.73±0.06</td>
<td>9.69±0.21</td>
<td>19.42±0.22</td>
<td>29.17±0.35</td>
</tr>
<tr>
<td>Maximum deviation from average leaf speed (mm/s)</td>
<td>0.25±0.01</td>
<td>0.64±0.03</td>
<td>0.92±0.06</td>
<td>1.58±0.06</td>
</tr>
<tr>
<td>RMSD between left and right bank speeds (mm/s)</td>
<td>0.10±0.00</td>
<td>0.21±0.00</td>
<td>0.32±0.01</td>
<td>0.49±0.01</td>
</tr>
<tr>
<td>Maximum deviation from nominal gap width (mm)</td>
<td>0.22±0.04</td>
<td>0.24±0.06</td>
<td>0.25±0.06</td>
<td>0.35±0.08</td>
</tr>
<tr>
<td>RMSD of measured and nominal gap widths (mm)</td>
<td>0.09±0.00</td>
<td>0.09±0.00</td>
<td>0.09±0.00</td>
<td>0.09±0.00</td>
</tr>
</tbody>
</table>

Table II shows that there is a slight decrease in the accuracy and precision of leaf performance with increasing leaf speed.

The unidirectional sliding gap delivered at a fixed gantry angle was also repeated at 90° collimator angle to include the outer leaves (only for the 9.67 mm/s leaf speed). The results showed that the inner and outer leaves had the same speeds, while the average gap width for the outer leaves was 0.24±0.32 mm larger than the inner leaves.

Results of the other experiments on leaf speed and acceleration/deceleration are summarized in Table III. The RMSD of the measured and nominal gap widths for all leaf positions/gantry angles are presented along with the maximum difference between the measured and nominal gap widths and the RMSD between the measured speeds for corresponding leaves in the left and right banks.
### TABLE III. Summary of the statistical investigation of the results for different tests on the leaf speed. The average of three measurements are presented with ±1SD.

<table>
<thead>
<tr>
<th>Test</th>
<th>Delivery mode</th>
<th>Leaf speed (mm/s)</th>
<th>RMSD of measured and nominal gap widths (mm)</th>
<th>Maximum deviation from nominal gap width (mm)</th>
<th>RMSD between left and right bank speeds (mm/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Unidirectional gap Arc</td>
<td>Arc</td>
<td>2.02</td>
<td>0.11±0.01</td>
<td>0.39±0.04</td>
<td>0.37±0.05</td>
</tr>
<tr>
<td>Bidirectional gap Static</td>
<td>Static</td>
<td>19.33</td>
<td>0.08±0.01</td>
<td>0.29±0.06</td>
<td>0.48±0.01</td>
</tr>
<tr>
<td>Unidirectional gap, different speeds Static 2.23 to 20.08</td>
<td>Static</td>
<td>2.23 to 20.08</td>
<td>0.09±0.00</td>
<td>0.31±0.05</td>
<td>0.39±0.00</td>
</tr>
<tr>
<td>Oscillating gap Arc</td>
<td>Arc</td>
<td>27.50</td>
<td>0.21±0.02</td>
<td>0.97±0.08</td>
<td>1.35±0.03</td>
</tr>
</tbody>
</table>

Tables II and III provide detailed data for comparison of several different leaf performance conditions. The arc delivery data showed larger deviations due to the gravity effect.

### III.D. Angular dependence

The picket fence test results measured at cardinal angles and during arc delivery are compared in Table IV. The data are averaged over all leaf pairs and picket positions, and are given ± 1SD. The number of data points were (40×7=280) for the fixed-gantry measurements and (40×10=400) in arc mode.

### TABLE IV. Comparison of the picket fence results at cardinal angles and during arc delivery. The average data over all leaf pairs are given ± 1SD. The nominal gap width was 1 mm.

<table>
<thead>
<tr>
<th>Gantry Angle (degrees)</th>
<th>0</th>
<th>90</th>
<th>180</th>
<th>270</th>
<th>Arc</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gap width (mm)</td>
<td>0.99±0.12</td>
<td>0.98±0.13</td>
<td>1.00±0.13</td>
<td>1.00±0.12</td>
<td>0.99±0.10</td>
</tr>
<tr>
<td>Deviation from average peak position (mm)</td>
<td>0.01±0.02</td>
<td>0.02±0.02</td>
<td>0.01±0.03</td>
<td>0.03±0.02</td>
<td>0.03±0.02</td>
</tr>
<tr>
<td>Deviation from expected peak position (mm)</td>
<td>0.21±0.05</td>
<td>0.23±0.06</td>
<td>0.24±0.15</td>
<td>0.33±0.07</td>
<td>0.30±0.07</td>
</tr>
<tr>
<td>MLC skewness (degrees)</td>
<td>-0.23±0.01</td>
<td>-0.31±0.00</td>
<td>-0.21±0.00</td>
<td>-0.09±0.01</td>
<td>-0.23±0.01</td>
</tr>
</tbody>
</table>

Table IV shows that the gantry rotation does not have a major effect on gap width (up to 0.02 mm). The maximum effect of gantry angle on the deviation of the peak positions from their expected positions (with respect to the beam central axis) was about 0.1 mm. The maximum effect on MLC skewness was ~0.1°. Similar results were obtained for the other two tested linacs. A similar investigation on gantry angle dependence was performed on DMLC using a unidirectional 5 mm sliding gap travelling at 19.3 mm/s speed. The results are given in Table V.
TABLE V. Comparison of the results of the unidirectional 5 mm sliding gap test at cardinal angles and during arc delivery. The average of three measurements is presented with ±1SD.

<table>
<thead>
<tr>
<th>Gantry Angle (degrees)</th>
<th>0</th>
<th>90</th>
<th>180</th>
<th>270</th>
<th>Arc</th>
</tr>
</thead>
<tbody>
<tr>
<td>RMSD of measured and nominal gap widths (mm)</td>
<td>0.09±0.00</td>
<td>0.09±0.00</td>
<td>0.12±0.00</td>
<td>0.10±0.00</td>
<td>0.11±0.00</td>
</tr>
<tr>
<td>Maximum deviation from nominal gap width (mm)</td>
<td>0.25±0.06</td>
<td>0.29±0.04</td>
<td>0.31±0.05</td>
<td>0.55±0.07</td>
<td>0.42±0.04</td>
</tr>
<tr>
<td>RMSD between left and right bank speeds (mm/s)</td>
<td>0.32±0.01</td>
<td>0.34±0.03</td>
<td>0.37±0.01</td>
<td>0.40±0.04</td>
<td>0.39±0.05</td>
</tr>
</tbody>
</table>

Table V shows that the RMSD of the measured and nominal gap width was not affected by the gantry rotation by more than 0.02 mm. The maximum deviation from nominal gap width at different gantry angles were up to 0.3 mm. Changes in the RMSD of the leaf speeds in the left and right banks at different gantry angles were affected by up to 0.1 mm/s. Similar results were obtained for the other two tested linacs.

**III.E. Accuracy of the algorithm**

Results of the detection of deliberately inserted errors in leaf positions in the picket fence pattern, and the detection of intentional changes made in gap positions are given in Table VI.

TABLE VI. Results of the algorithm for detection of the intentionally inserted errors in the picket fence pattern and in the gap positions. The average of three measurements is presented with ±1SD.

<table>
<thead>
<tr>
<th>Nominal inserted shift (mm)</th>
<th>0.1</th>
<th>0.2</th>
<th>0.3</th>
<th>0.5</th>
<th>0.7</th>
<th>1.5</th>
<th>2</th>
<th>2.5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Detected change in gap width (mm)</td>
<td>0.11±0.02</td>
<td>0.20±0.03</td>
<td>0.31±0.02</td>
<td>0.52±0.01</td>
<td>0.69±0.02</td>
<td>1.50±0.01</td>
<td>2.01±0.02</td>
<td>2.49±0.02</td>
</tr>
<tr>
<td>Detected change in leaf pair position (mm)</td>
<td>0.10±0.00</td>
<td>0.21±0.02</td>
<td>0.31±0.02</td>
<td>0.51±0.01</td>
<td>0.70±0.01</td>
<td>1.49±0.02</td>
<td>2.00±0.03</td>
<td>2.50±0.01</td>
</tr>
</tbody>
</table>

The average difference between the nominal and measured shifts in the gap widths and leaf pair positions were both 0.01 mm.

According to Table II, the sliding gap algorithm detected the leaf speed for the range of tested velocities for a unidirectional gap with an average accuracy of 0.07 mm/s. The RMSD of the measured and nominal gap widths across the field was 0.09 mm for all tested leaf speeds. The detected angles of the rotated image were compared with the applied (nominal) rotation angles. The result was an RMSD of 0.03 degrees which was considered as the accuracy of the Radon transform.

**IV. DISCUSSION**

In advanced radiotherapy treatments, errors in the width of MLC-defined gaps could lead to inaccurate delivery of the prescribed doses, and errors in the gap position could result in
geometric inaccuracies in tumour targeting. Correct MLC positioning is more important for highly modulated IMRT fields and arc treatments, especially for fields with small apertures, or at the edges of the subfields where dose errors are proportional to the penumbra slope.

Considering the complexity of IMRT/VMAT treatments, accurate delivery of the prescribed doses requires stable and reliable electromechanical systems which in turn necessitate extensive knowledge of MLC positioning and accurate quality assurance methods. Ideally, the best practice would be to develop simple methods to cover more efficient QA programs. In this study, robust algorithms are developed to extract greater information from some well-known tests by observing the collected measured data from a different point of view, and by adding new tests for more comprehensive investigation of the system.

The picket fence test was used to provide information on leaf positions, gap widths and skewness of the system over the whole range of leaf positions. The distinction of the picket fence test results in the present work compared with previous studies is in finding the deviation of each detected peak position from its expected position with respect to the beam central axis (to find systematic shifts), and also in determination of the system skewness. In addition, the leaf-by-leaf analysis performed in this study provides quantitative data for each individual leaf with sub-pixel accuracy, and the analysis results are presented through several graphs to make it easier for the user to detect the leaf tolerance violations and to quickly review the analysis results. The log files provide a useful source of archived information for long term investigations of the stability of the MLC performance.

Deviations up to 0.52 mm in gap width and 0.80 mm from the expected peak positions were recorded among three linacs over five months of observation. Performing the test in both fixed-gantry (at cardinal angles) and arc modes generalized the QA method to both IMRT and VMAT deliveries. The results with the gantry fixed at cardinal angles showed that the peak positions were displaced from their expected positions by up to 0.1 mm, while the effect of gravity and friction on the gap widths was negligible. This confirmed the report by Vieira et al. (2002). It should be emphasized that the EPID and gantry sag effect on the tests performed in arc mode were found to be negligible in the tested linacs and were not considered in this study.

The heavy weight of the MLC assembly results in excessive force imposed on the carriage bearings which may lead to the skewing of the leaf banks. There is also a possibility of some small EPID skewing relative to the linac head. In this study, the overall skewness of the system including the MLC assembly, collimator and EPID was investigated and the contribution of the MLC system alone was separately evaluated for the tested linacs, which was less than 0.2°. Misalignments up to 1 degree would only result in negligible change in dose distribution as reported by Bhagwat and is in accordance with the AAPM task group report 142. If the system exceeds the tolerance levels for skewness or has a large systematic shift in leaf
positioning, the results may be attributed to carriage sag and adjustment or replacement of carriage bearings or adjustment of its belt tension may be required. However, the possibility of an error in the calibration procedure should also be considered.

Some leaf positioning errors could be temporarily eliminated by re-initializing the MLC. It was shown in this study that four hours of linac work (after MLC initialization) could affect 3% of the peak positions in a picket fence pattern by a maximum of 0.2 mm and could change 12% of the gap widths by up to 0.4 mm. Therefore, it would be a good idea to re-initialize the MLC every four hours. It is important to note that errors in the leaf calibration process could be the main cause of systematic errors in leaf banks. These errors are affected by the stability of the infrared source used for MLC calibration.

Deliberately inserted errors in leaf positions in a range of 0.1 mm to 2.5 mm as well as changes in the gap positions were detected by the algorithm with an average accuracy of 0.01 mm. These results indicate the superiority of the method used in this work to previous studies which were based on visual inspection of the errors on film or EPID images.

In addition to the picket fence test which evaluates the MLC performance in static leaf positions, a comprehensive QA program would require investigation of the behaviour of leaves in dynamic beam deliveries. In this study, assessment of the dynamic MLC performance was based on the evaluation of sliding gaps travelling across the field in different conditions in both gantry-fixed and arc modes. In arc deliveries, gravity and friction between leaves (e.g. due to the accumulation of dirt or grease) may affect the stability of MLC speed during gantry rotation. The sliding gap test is sensitive to small changes in leaf position and speed across the range of leaf travel which could be affected by motor fatigue and lead to changes in the leaf acceleration/deceleration.

Using the EPID in cine imaging mode provides an opportunity to analyse the images of a sliding gap as time-lapse series. Thus, another dimension was added to the results and changes in the gap widths, leaf speeds and leaf acceleration/deceleration could be examined. This is a major advantage over film measurements. It must be noted that movements of the gantry and imager during the delivery are not expected to cause blurring at the edge of images since only one frame is used per image.

The accuracy and precision of leaf positioning is slightly reduced for higher leaf speed deliveries. This finding is in accordance with previous studies. The leaf speed for a unidirectional gap in arc mode was lower than the fixed-gantry setup, since the aim was to perform the test in an entire 360° arc delivery while the maximum gantry speed is limited to 5 degrees per second. However, the gap width and leaf speeds were both affected by gravity during arc delivery. The change in the direction of leaf travel in a bi-directional gap in fixed-gantry condition resulted in an increase in the RMSD between the speeds of corresponding
leaves in the left and right banks compared with a unidirectional sliding gap moving with the same leaf speed. Application of speed gradients to the unidirectional sliding gap (average speed ~10 mm/s) led to larger differences between the nominal and measured gap widths and larger RMSD between the speeds of corresponding leaves in the left and right banks when compared to a gap travelling with a constant speed of 9.67 mm/s. The results for the oscillating gap test had the largest deviations and RMSDs, as expected. Although the test conditions are extremely stringent and are unlikely to be applied in clinical practice, it is a suitable test for QA purposes. The accepted tolerance for deviation of leaf speed from its nominal value was ±5 mm/s in this study, based on the AAPM task group report 142. However, in the most difficult test conditions (oscillating gap during arc with high speed), the maximum deviation of leaf speeds from the nominal value was less than 3 mm/s.

The tests performed in this study in conjunction with the algorithms could be included in the routine QA tests as part of the MLC maintenance procedures for IMRT/VMAT to identify problems with MLC motors before they become clinically significant. The MLC interlock is usually activated at 2 mm leaf positioning error to avoid unacceptable number of beam interruptions. Therefore, much tighter tolerances are required for the QA of IMRT/VMAT. Jorgensen et al. have proposed 0.3 mm and 0.5 mm tolerance levels for leaf positioning and gap width errors in VMAT QA which seem reasonable based on the results of this study.

Leaving a slow moving leaf unchanged could finally lead to its getting stuck during a treatment session. Based on the results of this study, considering the quick test procedure and the short algorithm processing times, a timetable is suggested for the MLC QA tests as scheduled in Table VII. The daily picket fence test after the last treatment helps in identification of the faulty motors more easily due to the longer time elapsed since the last MLC re-initialization. It is also necessary to perform the picket fence test after any changes to the optical system for MLC calibration.

<table>
<thead>
<tr>
<th>Test</th>
<th>Fixed gantry angles</th>
<th>Arc</th>
<th>At cardinal angles</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Picket fence</strong></td>
<td>• Daily, after machine warm-up</td>
<td>Weekly</td>
<td>Monthly</td>
</tr>
<tr>
<td></td>
<td>• Daily, after the last treatment</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>• Before patient specific QA</td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Unidirectional sliding gap</strong></td>
<td>Daily</td>
<td>Weekly</td>
<td>Monthly</td>
</tr>
<tr>
<td><strong>Other sliding gap tests</strong></td>
<td>Monthly</td>
<td>Monthly</td>
<td>Monthly</td>
</tr>
</tbody>
</table>

After changing a leaf motor it would be required to check its performance by carrying out a picket fence test and a unidirectional sliding gap test at fixed gantry angles.
It must be noted that comprehensive QA for VMAT treatments requires two other tests. These tests were introduced by Ling et al. in 2008\textsuperscript{22} and include investigation of the effects of varying dose rate/gantry speed, and varying dose rate/MLC speed. EPID images can be used for these tests as addressed by Jorgensen et al.\textsuperscript{7}

V. CONCLUSION

This study provides convenient test methods for routine QA of the MLC performance as well as troubleshooting of the MLC problems in both IMRT and VMAT treatments. The fast, easy and accurate measurement methods and the robust algorithms introduced in this study help in performing the tests more frequently and therefore improve the clinical outcomes. The large amounts of information shown on several graphs enable the physicists to have immediate access to many quantities which could help finding the problems by looking at them from various aspects and decision making on selection of the most convenient rectification strategy.

ACKNOWLEDGMENT

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References:


Conclusion
Advances in radiotherapy have increased the complexity of treatment delivery techniques. The complexity of plans, with dynamic variation of field shape, gantry speed and dose rate require highly accurate techniques for quality assurance of the treatment machines and dosimetric verification of the treatment plans.

There has been a growing interest on the application of electronic portal imaging devices (EPIDs) for dosimetry applications and quality assurance testing of linear accelerators (linacs). This thesis has focused on development of EPID-based techniques to ensure more accurate treatment deliveries.

The work was divided into two sections. The first section is based on improvement of the accuracy of EPID dosimetry with Varian systems by either accounting for, or reduction of, the effect of backscattered radiation from the treatment room walls and the EPID support arm.

In Chapter 2, the effect of backscattered radiation from the EPID support arm was predicted using a measured backscatter kernel. This was included in an existing EPID dose prediction model to improve the EPID dosimetry results. Although the kernel did not take the exact backscatter distribution from the complex structure of the arm, it improved the outcome of EPID dosimetry calculations for all tested IMRT fields.

The next approach (Chapter 3) was to add a large sheet of lead between the EPID and support arm to block the non-uniform arm backscattered radiation. The optimum thickness of the sheet was determined as 2 mm. This led to better symmetry (for both 6 and 18 MV beam energies), and reduced non-uniform arm backscatter; while the image quality, EPID sag and patient skin dose remained unchanged or had negligible differences.

Finally, in Chapter 4 a combination of the above methods was used to improve the accuracy of EPID dosimetry. A small sheet of 5 mm thick lead was fixed on the arm area to reduce the effect of structural non-uniformities. The lead was considered as part of the arm. This setup helped avoid the excessive amount of weight added to the imager as a result of the insertion of a large sheet of lead. The resulting backscatter kernel was measured and included in the existing EPID dose prediction model which improved the dosimetry results for the tested IMRT fields. However, adding the piece of lead alone (without applying the kernel in the model) can reduce the non-uniform backscatter to a reasonably low level. This could be useful for radiotherapy centres which do not have access to EPID dose prediction models.

Although the impact of the backscattered radiation from the walls, ceiling and floor of the bunker was expected to be very small but it was worthwhile to perform a systematic and quantitative study on the subject. In Chapter 5, the effect of backscattered radiation from the...
treatment room structure was quantified for the first time using a number of measurement setups including different field sizes, SDDs, with and without a phantom in the beam. In addition, comparisons were made with measurements in the presence of an independent portable wall. It was shown that the effect can be ignored altogether for pre-treatment verifications with the imager panel at the isocentre (SDD= 100 or 105 cm), but the effect gradually increases with increasing SDD and even more so when the larger SDD is combined with transit dosimetry. Fortunately, even in this ‘worst case scenario’ the effect still remains limited to 1% at its maximum.

In the second section of this thesis, EPID-based measurement methods were used and new algorithms were developed for faster, easier, more robust, more accurate quantitative techniques to characterize the linac components. The results could be used for improvement of EPID dosimetry measurements and/or be included in the linac quality assurance program.

Chapter 6 includes two papers on isocentre verification methods suitable for stereotactic radiosurgery/radiotherapy. First, a review article on the existing techniques and comparison of their characteristics with the aim of helping physicists make a decision on selection of their preferred quality assurance routine. Second, introduction and testing of a method based on cine EPID imaging of a Winston-Lutz phantom, and development of an algorithm to find the isocentre position. The method was simple, fast, highly accurate, and independent of the manufacturer. This method is sufficiently quick to be performed before each stereotactic treatment.

Chapters 7 and 8 include studies on determination of the sag in EPID, gantry, jaws and MLC systems during arc IMRT/VMAT deliveries. All measurement methods were based on cine EPID imaging of simple and easily available ball-bearing phantoms which can easily be included in the linac QA programs as routine tests. The algorithms developed for each part of these studies are accurate, fast and robust and can also be used to improve EPID dosimetry results.

In Chapter 9, a measurement method and algorithm are presented for QA of gantry angle, based on integrated EPID images acquired at distinct gantry angles and cine EPID images during an entire 360° arc. A comprehensive study was carried out to evaluate this method as well as two commercially available inclinometers by comparison of their simultaneous angle measurement results with the dynamic log files (provided by the manufacturer) at five gantry speeds.

Finally in Chapter 10, a comprehensive study is explained on some important MLC characteristics during IMRT/VMAT deliveries. Reliable QA techniques for MLC leaves were implemented based on some popular test patterns that have already been accepted and used worldwide. The algorithms developed in this work for MLC QA can automatically provide a large amount of data on the behaviour of each individual leaf and present them in an efficient manner.
Robust algorithms were developed in-house to quantitatively investigate: (a) the accuracy and stability of each individual MLC leaf position and their deviation from the expected leaf positions, in addition to the gap widths for each leaf pair; (b) the skewness of leaf banks/carriage; (c) the speed of each leaf over the whole range of its positions and the stability of gap width during a sliding gap delivery; and (d) the acceleration/deceleration of each individual leaf. All this was only made possible with the algorithms. Quantitative information were extracted from well-known QA tests by observing the collected measured data from a different point of view, and by adding new tests for more comprehensive investigation of the MLC system. The main goal of this study was to develop and implement comprehensive, faster, and more accurate QA techniques for MLC leaves in both dynamic IMRT and VMAT deliveries.

The methods proposed in the second part of this thesis have been tested and are applicable for quality assurance of the linear accelerators used for advanced treatment techniques with all linacs, independent of their make and model. They make up a useful collection of QA tools for routine application in the clinic.
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**APPENDIX I: LIST OF ACRONYMS**

<table>
<thead>
<tr>
<th>Acronym</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>AMFPI</td>
<td>Active Matrix Flat Panel Imager</td>
</tr>
<tr>
<td>a-Si</td>
<td>amorphous Silicon</td>
</tr>
<tr>
<td>BB</td>
<td>Ball Bearing</td>
</tr>
<tr>
<td>BSK</td>
<td>Backscatter Kernel</td>
</tr>
<tr>
<td>CAX</td>
<td>Central Axis</td>
</tr>
<tr>
<td>CNR</td>
<td>Contrast to Noise Ratio</td>
</tr>
<tr>
<td>CT</td>
<td>Computed Tomography</td>
</tr>
<tr>
<td>DICOM</td>
<td>Digital Imaging and Communications in Medicine</td>
</tr>
<tr>
<td>DMLC</td>
<td>Dynamic Multileaf Collimator</td>
</tr>
<tr>
<td>EPID</td>
<td>Electronic Portal Imaging Device</td>
</tr>
<tr>
<td>FF</td>
<td>Flood Field</td>
</tr>
<tr>
<td>FWHM</td>
<td>Full Width at Half Maximum</td>
</tr>
<tr>
<td>IC</td>
<td>Integrated Circuits</td>
</tr>
<tr>
<td>IDU</td>
<td>Image Detection Unit</td>
</tr>
<tr>
<td>IMAT</td>
<td>Intensity Modulated Arc Therapy</td>
</tr>
<tr>
<td>IMRT</td>
<td>Intensity Modulated Radiation Therapy</td>
</tr>
<tr>
<td>IMSR</td>
<td>Intensity Modulated Stereotactic Radiotherapy</td>
</tr>
<tr>
<td>kV</td>
<td>Kilovoltage</td>
</tr>
<tr>
<td>Linac</td>
<td>Linear accelerator</td>
</tr>
<tr>
<td>MeV</td>
<td>Mega Electron Volts</td>
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<tr>
<td>MLC</td>
<td>Multi-Leaf Collimator</td>
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<tr>
<td>MRI</td>
<td>Magnetic Resonance Imaging</td>
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<tr>
<td>MRS</td>
<td>Magnetic Resonance Spectroscopy</td>
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<tr>
<td>MU</td>
<td>Monitor Unit</td>
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<tr>
<td>MV</td>
<td>Megavoltage</td>
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<tr>
<td>NRMSD</td>
<td>Normalized Root Mean Squared Deviation</td>
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<tr>
<td>PECVD</td>
<td>Plasma Enhanced Chemical Vapour Deposition</td>
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<tr>
<td>Acronym</td>
<td>Description</td>
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<tr>
<td>PET</td>
<td>Positron Emission Tomography</td>
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<tr>
<td>PMMA</td>
<td>Polymethylmethacrylate</td>
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<td>PSM</td>
<td>Pixel Sensitivity Matrix</td>
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<td>RD</td>
<td>Relative Difference</td>
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<td>RF</td>
<td>Radiofrequency</td>
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<td>RMSD</td>
<td>Root Mean Square Deviation</td>
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<td>SBRT</td>
<td>Stereotactic Body Radiotherapy</td>
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<tr>
<td>SLIC EPID</td>
<td>Scanning Liquid-Filled Ionization Chamber EPID</td>
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<tr>
<td>SSD</td>
<td>Source to Surface Distance</td>
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<td>SPECT</td>
<td>Single Photon Emission Tomography</td>
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<td>Total Energy Released per unit Mass</td>
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<td>Thin Film Transistor</td>
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<td>Thermoluminescent Dosimeter</td>
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<td>VEPID</td>
<td>Video based EPID</td>
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<td>VMAT</td>
<td>Volumetric Modulated Arc Therapy</td>
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