Dynamic Electron Arc Radiotherapy (DEAR): A New Conformal Electron Therapy Technique

by

Anna Elisabeth Rodrigues

Graduate Program in Medical Physics
Duke University

Date:_______________________

Approved:

Qiuwen Wu, Supervisor

Fang-Fang Yin

Mark Oldham, Chair

Rachel Blitzblau

Martin Tornai

Dissertation submitted in partial fulfillment of the requirements for the degree of Doctor of Philosophy in the Graduate Program in Medical Physics in the Graduate School of Duke University

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ABSTRACT

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Abstract

Electron beam therapy represents an underutilized area in radiation therapy. While electron radiation therapy has existed for many decades and electron beams with multiple energies are available on linear accelerators – the most common device to deliver radiation therapy – efforts to advance the field have been slow. In contrast, photon beam therapy has seen rapid advancements in the past decade, and has become the main modality for radiation therapy treatment.

This doctoral research project comprises the development of a novel treatment modality, dynamic electron arc radiotherapy (DEAR) that seeks to address challenges to clinical implementation of electron beam therapy by providing a technique that may be able to treat specific patient subsets with better outcomes than current techniques. This research not only focused on the development of DEAR, but also aimed to improve upon and introduce new tools and techniques that could translate to current clinical electron beam therapy practice.

The concept of DEAR is presented. DEAR represents a new conformal electron therapy technique with synchronized couch motion. DEAR utilizes the combination of gantry rotation, couch motion, and dose rate modulation to achieve desirable dose distributions in patient. The electron applicator is kept to minimize scatter and maintain
narrow penumbra. The couch motion is synchronized with the gantry rotation to avoid collision between patient and the electron cone.

First, the feasibility of DEAR delivery was investigated and the potential of DEAR was demonstrated to improve dose distributions on simple cylindrical phantoms. DEAR was delivered on Varian’s TrueBeam linac in Research Mode. In conjunction with the recorded trajectory log files, mechanical motion accuracies and dose rate modulation precision were analyzed. Experimental and calculated dose distributions were investigated for a few selected energies (6 MeV and 9 MeV) and cut-out sizes (1x10 cm$^2$ and 3x10 cm$^2$ for a 15x15 cm$^2$ applicator). Our findings show that DEAR delivery is feasible and has the potential to deliver radiation dose with high precision (RMSE of <0.1 MU, <0.1° gantry, and <0.1 cm couch positions) and good dose rate precision (1.6 MU/min). Dose homogeneity within ±2 % in large and curved targets can be achieved while comparable penumbra to a standard electron beam on a flat surface can be maintained. Further, DEAR does not require fabrication of patient-specific shields, which has hindered the widespread use of electron arc therapy. These benefits make DEAR a promising technique for conformal radiotherapy of superficial tumors.

Next, an accurate dose calculation framework for DEAR was developed since current commercial dose calculation systems cannot handle the dynamic nature of the DEAR. Comprehensive validations of vendor provided electron beam phase space files for Varian TrueBeam linacs against measurement data were assessed. In this framework,
the Monte Carlo generated phase space files were provided by the vendor and used as
input to the downstream plan-specific simulations including jaws, electron applicators,
and water phantom computed in the EGSnrc environment. The phase space files were
generated based on open field commissioning data. A subset of electron energies of 6, 9,
12, 16, and 20 MeV and open and collimated field sizes 3×3, 4×4, 5×5, 6×6, 10×10, 15×15,
20×20, and 25×25 cm² were evaluated. Measurements acquired with a CC13 cylindrical
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were compared for a water phantom geometry. The evaluation metrics include percent
depth dose, orthogonal and diagonal profiles at depths R₁₀₀, R₅₀, Rₚ, and Rₚ⁺ for standard
and extended source-to-surface distances (SSD), as well as cone and cut-out output
factors. Agreement for the percent depth dose and orthogonal profiles between
measurement and Monte Carlo were generally within 2% or 1 mm. The largest
discrepancies were observed for depths within 5 mm from the phantom surface.
Differences in field size, penumbra, and flatness for the orthogonal profiles at depths
R₁₀₀, R₅₀, Rₚ, and Rₚ⁺ were within 1 mm, 1 mm, and 2%, respectively. Simulated and
measured orthogonal profiles at SSDs of 100 and 120 cm showed the same level of
agreement. Cone and cut-out output factors agreed well with maximum differences
within 2.5% for 6 MeV and 1% for all other energies. Cone output factors at extended
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Monte Carlo simulation framework for electron beam dose calculations for Varian
TrueBeam linacs for electron beam energies of 6 to 20 MeV for open and collimated field sizes from 3×3 to 25×25 cm² were studied and results were compared to the measurement data with excellent agreement.

DEAR uses the superposition of many small fields for its delivery, as such accurate planning requires the knowledge of accurate small field dosimetry. Prior research has shown that previous versions of the clinically used eMC dose calculation algorithm (Varian Medical Systems, Inc., Palo Alto, CA) cannot accurately calculate small static electron fields, leading to discrepancies in the dose distributions and output. Further, the clinical treatment planning system, Eclipse, currently does not support the planning of dynamic electron radiation therapy. Therefore, the aforementioned validation was extended to small fields and compared to dose calculations from the treatment planning system.

Subsequently, small field optimization was explored. Monte Carlo simulations were performed using validated Varian TrueBeam phase space files for electron beam energies of 6, 9, 12, and 16 MeV and square (1x1, 2x2, 3x3, 4x4, and 5x5 cm²) and circular (1, 2, 3, 4, and 5 cm diameter) fields. Resulting dose distributions (kernels) were used for subsequent calculations. The following analyses were performed: (1) Comparison of composite square fields and reference 10x10 cm² dose distributions and (2) Scanning beam deliveries for square and circular fields realized as the convolution of kernels and scanning pattern. Preliminary beam weight and pattern optimization were also
performed. Two linear scans of 10 cm with/without overlap were modeled. Comparison metrics included depth and orthogonal profiles at \( d_{\text{max}} \). (1) Composite fields regained reference depth dose profiles for most energies and fields within 5%. Smaller kernels and higher energies increased dose in the build-up and Bremsstrahlung region (30%, 16 MeV and 1x1 cm\(^2\)), while reference \( d_{\text{max}} \) was maintained for all energies and composite fields. Smaller kernels (<2x2 cm\(^2\)) maintained penumbra and field size within 0.2 cm, and flatness within 2 and 4% in the cross-plane and in-plane direction, respectively. Deterioration of penumbra for larger kernels (5x5 cm\(^2\)) was observed. Balancing desirable dosimetry and efficiencies suggests that smaller kernels should be used at the target edges and larger kernels in the center of the target. (2) Beam weight optimization improves cross-plane penumbra (0.2 cm) and increases the field size (0.4 cm) on average. In-plane penumbra and field size remain unchanged. Overlap depends on kernel size and optimal overlap results in flatness \( \pm 2\% \). Dynamic electron beam therapy in virtual scanning mode is feasible by employing small fields to achieve desired dose distributions and acceptable efficiencies.

Further, tools to generally improve upon limitations in Monte Carlo simulations for electron beams were investigated. The phase space file contains a finite number of particle histories and can have very large file size, yet still contains inherent statistical noises. A characterization of the phase space file was investigated to overcome its inherent limitations. To characterize the phase space file, distributions for energy,
position, and direction of all particles types were analyzed as piece-wise parameterized functions of radius. Subsequently, a pseudo phase space file was generated based on this characterization. Validation was assessed by directly comparing the original and pseudo phase space file, and by comparing the resulting dose distributions from Monte Carlo simulations using both phase space files. Monte Carlo simulations were run for energies 6, 9, 12, and 16 MeV and all standard field sizes 6x6, 10x10, 15x15, 20x20, and 25x25 cm². Percent depth dose and orthogonal profiles at depths $R_{100}$, $R_{50}$, and $R_p$ were evaluated. Histograms of the original and pseudo phase space file agree very well with correlation coefficients greater than 0.98 for all particle attributes. Dosimetric comparison between original and pseudo dose distributions yielded agreement within 2%/1mm for PDDs and profiles at all depths for all field sizes 6x6, 10x10, 15x15, 20x20, and 25x25 cm² and energies 6, 9, 12, and 16 MeV. Phase space files were found to be successfully characterized by piece-wise distributions for energy, position, and direction as parameterized functions of radius and polar angle. This facilitates generation of sufficient particles at any statistical precisions.

Additionally, new hardware for improved DEAR capability was investigated. Few leaf electron collimators (FLEC) or electron MLCs (eMLC) are highly desirable for dynamic electron beam therapies as they produce multiple apertures within a single delivery to achieve conformal dose distributions. However, their clinical implementation has been challenging. Alternatively, multiple small apertures in a single cut-out with
variable jaw sizes could be utilized in a single dynamic delivery. A Monte Carlo simulation study was performed to investigate the dosimetric characteristics of such an arrangement. Investigated quantities included: Energy (6 and 16 MeV), jaw size (1x1 to 22x22 cm$^2$; centered to aperture), applicator/cut-out (15x15 cm$^2$), aperture (1x1, 2x2, 3x3, and 4x4 cm$^2$), and aperture placement (on/off central axis). Three configurations were assessed: (a) single aperture on-axis, (b) single aperture off-axis, and (c) multiple apertures. Reference was configuration (a) with the standard jaw size. Aperture placement and jaw size were optimized to maintain reference dosimetry and minimize leakage through unused apertures to <5%. Comparison metrics included depth dose and orthogonal profiles. Configuration (a) and (b): Jaw openings were reduced to 10x10 cm$^2$ without affecting dosimetry (gamma 2%/1mm) regardless of on- or off-axis placement. For smaller jaw sizes, reduced surface (<2%, 5% for 1x1 cm$^2$ aperture) and increased Bremsstrahlung (<2%, 10% for 1x1 cm$^2$ aperture) dose was observed. Configuration (c): Optimal aperture placement was in the corners (order: 1x1, 4x4, 2x2, 3x3 cm$^2$ for quadrants I, II, III, and IV) and jaw size were 2x2, 2x2, 3x3, and 7x7 cm$^2$ and 7x7, 7x7, 10x10, and 10x10 cm$^2$ for apertures: 1x1, 2x2, 3x3, 4x4 cm$^2$ and energies 6 and 16 MeV, respectively. Asymmetric leakage was found from upper and lower jaws. Leakage was generally within 5% with a maximum of 10% observed for the 1x1 cm$^2$ aperture irradiation. Multiple apertures in a single cut-out with variable jaw size can be used in a single dynamic delivery, thus providing a practical alternative to FLEC or eMLC.
Based on all the results from this project, DEAR has been found to be a feasible technique and demonstrates the potential to improve electron therapy.
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Oma, Opa,

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Andrea, Ernst,

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List of Abbreviations

CR       Computed radiography
DAH      Dose area histogram
DVH      Dose volume histogram
DEAR     Dynamic electron arc radiotherapy
EAT      Electron arc therapy
eMC      Electron Monte Carlo algorithm
eMLC     Electron multi-leaf collimator
EGSnrc   Electron gamma shower national research council
EPID     Electronic portal imaging device
FLEC     Few leaf electron collimator
FWHM     Full width at half maximum
GB       Gigabyte
IMN      Internal mammary lymph nodes
KB       Kilobyte
LINAC    linear accelerator
MERT     Modulated electron radiation therapy
MeV      Mega-voltage, energy of the incident electron beam
MC       Monte Carlo
MU       Monitor unit
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<tr>
<td>NTCP</td>
<td>Normal tissue complication probability</td>
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<td>OF</td>
<td>Output factor</td>
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<td>PDD</td>
<td>Percent depth dose</td>
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<tr>
<td>QUANTEC</td>
<td>Quantitative analysis on normal tissue effects in the clinic</td>
</tr>
<tr>
<td>SSD</td>
<td>Source to surface distance</td>
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<tr>
<td>TPS</td>
<td>Treatment planning system</td>
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1. Introduction

1.1. Treating cancer with radiation

Cancer is a major public health issue and currently ranks as the second-leading cause of death in the United States. On average, the lifetime risk of developing any form of cancer is 43% and 37% for males and females, respectively. Improvements in medicine and healthcare have been positively correlated with increasing age, such that while we are curing many chronic diseases, cancer incidences will likely increase, and cancer has been projected to soon surpass heart disease as the leading cause of death (Siegel et al 2015). Thus the treatment and management of cancer plays an important and central role in improving public health.

Cancer is a class of diseases defined by an abnormal, out-of-control cell growth, leading to malignant tumors that are characterized by their ability to metastasize to neighboring or distant regions in the body. If left untreated, cancer can cause solid organ failure, brain and nervous system problems, pain, and death. Treatment options include surgery, chemotherapy, and radiation therapy, with approximately two-thirds of cancer patients in the United States receiving radiation therapy as treatment (2003). Radiation therapy is a non-invasive treatment technique using ionizing radiation to kill cancerous cells. Herein, only teletherapy techniques are detailed, however, brachytherapy using
radioactive sources placed inside or close to the tumors either permanently or transiently are successfully used to treat a variety of cancers.

Ionizing radiation kills cells by depositing energy by either directly or indirectly destroying the cell’s DNA. In the optimal scenario, targeted radiation renders the cell unable to replicate and kills the cancerous cell. While both photons and charged particles are used as radiation sources to treat cancer, their physical interaction with matter differs, therefore depending on the cancer type and site, one source may be preferred over another.

1.2. Delivery of ionizing radiation

1.2.1. Physical interaction of radiation with matter

The energy deposition for megavoltage photon and electron beams in water as a function of depth is characterized by the percent depth dose (PDD) curve and several examples are shown for typical photon and electron beam energies in Figure 1.
Figure 1. Percent depth dose (PDD) curves in water for a typical electron (left) and photon (right) beam energies. The PDD curves were obtained from clinical commissioning data for a Varian TrueBeam linac.

The electron PDD is characterized by a steep build-up region with a peak at the depth of maximum dose $d_{\text{max}}$, followed by a fall-off region, and a constant Bremsstrahlung region. Various metrics can be used to describe the PDD, and are described in Figure 2: The range concept defines depths $R_{\text{dose}}$% on the electron PPD curve at which the PDD attains a value of [dose]%. Thus, e.g., $R_{100}$ and $R_{50}$ describe the range for PDD values of 100% (or $d_{\text{max}}$) and 50% dose. The practical range $R_p$ is defined as the depth at which the tangent plotted through the fall-off region intersects the extrapolation of the Bremsstrahlung background (Podgorsak and IAEA 2005).
Figure 2. A typical electron PDD describing the definitions of $R_{100}$, $R_{50}$, and $R_p$ that are used to characterize an electron PDD.

The interaction with matter is different for electrons than for photons: Photons lose their energy mainly through Compton scattering in human tissue for energies in the megavoltage range with typical photon energies of 6, 10, and 15 MV. Electron interactions are dominated by elastic and inelastic Coulomb force interactions of the electron with the atoms in the media. An electron can lose its energy in two ways: through inelastic collisional or radiative losses. Collisional losses occur when an electron
excites a bound electron or ionizes the atom, freeing a bound electron, and thus
generating a secondary δ-electron. Radiative losses occur when an electron passes close to
a nucleus, thus experiencing an electrostatic force that leads to the deceleration and
change in direction of the electron under the emission of Bremsstrahlung x-rays. The
rate of energy loss for collisional losses depends on the electron density of the media,
while the rate of energy loss for radiative losses is proportional to the energy of the
electron and $Z^2$. Collisional losses dominate in tissue, while radiative losses become
larger for higher Z material due to the $Z^2$ dependence.

As the charged particle passes through matter, the electrons undergo multiple
scattering due to the aforementioned interactions which cause the electrons to slow
down and change their direction, thus gaining velocity components and displacements
transverse to the original direction of motion. This leads to energy and range straggling
of the electrons. Eventually, an electron will lose all of its kinetic energy and come to
rest. The sharp fall-off after the build-up region can be explained by a dramatic increase
in stopping power as electrons become less energetic. There will be a finite distance,
beyond which there will be no electrons and just dose due to Bremsstrahlung – this is
defined as the maximum range of the electron. This is in contrast to photons, whose
attenuation in media can be described in a first approximation of simple absorption and
no scattering or secondary radiation, by an exponential. In context of dose deposition in
human tissue, as described in Figure 1 by the PDD curves in a homogenous water phantom, electrons deposit their energy within a few centimeters of the surface. It is thus clear that – at least intuitively – electrons would be the preferred means by which to treat superficial cancers.

### 1.2.2. Electron beam generation

Electron beams are available on linear accelerators (linac) with many discrete energies (e.g. 6, 9, 12, 16, 20, and 22 MeV). Once the electron exits the accelerator, the beam needs to be scattered to a broad beam and shaped laterally to provide a useful beam. Figure 3 provides an overview of the important beam-line components that shape the beam before it enters the patient.
Figure 3. (left) Photo of a linac with an electron applicator (shown inside the dashed box) and a water tank. The three levels of the applicator are denoted. (right) Schematic of beam-line components in the linac treatment head that shape the electron beam after exiting the bending magnet.

As the electron beam exits the bend magnet, it is a focused pencil beam with a diameter of approximately 2 - 3 mm. More realistically, the beam has some spatial distribution, which can be approximated by a 2D Gaussian. The beam also has some angular divergence, i.e. the electron beam is not a parallel beam. Further, the energy spectrum can also be approximated by a Gaussian distribution. The energy spread is dictated by the tolerance of the energy slits in the bending magnet, which typically only
allow electrons with an energy of ±3% to pass through (Karzmark et al 1993). Generally, the spatial, angular, and energy distribution depend on the linac construction and materials.

Important beam-line components that further shape the beam both along the beam axis and laterally after exiting the bending magnet include: an energy dependent dual-scattering foil system, secondary, and tertiary collimation. The dual scattering foil shapes the beam along the beam axis, providing a uniform, flat, and broad beam. To mitigate the effects of beam divergence and scattering at the beam edge, secondary (jaws) and tertiary (electron applicator/cone) collimation must be used. The downstream edge of the electron applicator is located close to the treatment surface (~5 cm) to maintain narrow beam penumbra. Standard electron applicator aperture ranges from 6x6 cm\(^2\) to 25x25 cm\(^2\). For each applicator, the jaws are set slightly larger than the applicator size to minimize scatter from the jaws of finite thicknesses. For customized field shaping, Cerrobend cut-outs are placed at the bottom of the electron applicator (Figure 3, third level closest to the patient).

Cerrobend cut-outs are usually patient-specific i.e. they conform to the beam-eye view of the target shape. They can be produced by generating the appropriate shape as a Styrofoam mold, then pouring molten Cerrobend around the mold. Once hardened, the Styrofoam mold can be removed, leaving the desired aperture shape.
1.3. **Superficial cancers and their treatment**

Common sites for superficial cancers located with 5 cm of the surface that can be treated with electron beams with energies of 6 – 20 MeV include the skin, head and neck, scalp, extremities, spinal cord, intraoperative sites, breast, and chest wall. The main advantage of electrons for these cases is that they spare dose to deeper-seated normal tissue structures. The following sections will briefly describe the goals for radiation therapy and current clinical techniques for breast and chest wall as these were the targeted geometries for this dissertation research.

1.3.1. **Breast and chest wall**

The goal of radiation therapy is primarily and most importantly to reduce the risk of local recurrence and improve survival from breast cancer. Secondary to that goal is the sparing of dose to underlying normal tissues, specifically dose to lung and heart for breast and chest wall cases. Treatment of breast and chest wall varies substantially, as it depends on the location, cancer type, extent of the disease, and patient anatomy. Whole breast, partial breast, post-lumpectomy, or post-mastectomy chest wall irradiations are possible.

During a lumpectomy only the tumor is removed and the breast is conserved. Post-lumpectomy radiation therapy can involve either the whole breast or partial breast. Whole breast radiation therapy may additionally include irradiation of the lymph nodes.
A mastectomy involves the excision of both the tumor and the entire breast. The goal of post-mastectomy radiation therapy is to treat the entire chest wall and lymph nodes as to reduce the risk of recurrence on the chest wall. In summary, the targets for radiation therapy of breast and chest wall include: Chest wall and lymph nodes, whole breast with or without lymph nodes, and tumor bed with a margin to account for microscopic diseases. The combination of the aforementioned variability of patients and treatment sites generally leads to patient-specific treatment planning.

A common approach in delivering radiation therapy involves the use of fractionated treatment using two tangential photon beams to treat the whole breast, partial breast, or chest wall. The tangential direction is preferred because photons are penetrating and this arrangement can minimize the normal tissues including lung and heart inside the beam path. *En face* electron beams, and less commonly photon beams, are sometimes used to treat internal mammary nodes (IMN). Supraclavicular and axillary nodes are treated with oblique photon fields. Additionally, an electron field may be administered in an abbreviated course (1-2 weeks, usually at the end of treatment) for the tumor bed or scar. This is called a boost field (i.e. tumor bed boost or scar boost). Clinical trials have shown that improved local control can be achieved by adding additional electron boost fields resulting in a higher radiation dose directed to the tumor bed (Hayman *et al* 2000, Benda *et al* 2003). Electron beams are used in this instance they
are less penetrating, and can minimize additional dose to heart and lung while still providing adequate target coverage.

For breast and chest wall, targets such as the tumor bed in the breast and lymph nodes are typically contoured and normal tissue dose-volume constraints are targeted to be as low as reasonably achievable, without sacrificing target coverage. Due to patient variability, no single “hard” radiation dose-volume threshold is deemed optimal to reduce normal tissue complications. Rather, clinical experience and recommendations provided by literature reviews such as the quantitative analysis on normal tissue effects in the clinic (QUANTEC) for heart (Gagliardi et al 2010) and lung (Marks et al 2010), help guide treatment planning. Further, normal tissue dose constraints may follow the ALARA principle – as low as reasonably achievable – without compromising the target coverage. Dose constraints for specific organs or structures are described by a cumulative dose volume histogram (DVH), which relates the radiation dose to the tissue volume.

1.3.1.1. Heart

Generally, treatment planning should be focused on reducing the heart dose-volume to be as low as reasonably achievable (Blitzblau and Horton 2013). As stated previously, no firmly agreed upon limits for the heart exist, rather historical outcomes and experience guide the dose constraints. Radiation-associated cardiac toxicities can be
observed for breast and chest wall irradiation and are clearly linked to increased non-
cancer related complications due to breast radiation therapy (Darby et al 2005, Darby et

However, recent advancements and increased complexity in radiation therapy
techniques have been used to reduce the dose to the heart. Breath-hold technique, prone
positioning of the patient, addition of mixed beam energies, electronic compensation,
and IMRT are all techniques that have been used to improve plans. For example, by
using a breath-hold technique, which maximizes the distance between the chest wall
and heart, dose to the heart can be reduced. However, despite these reductions in heart
dose, it remains unclear whether modern radiation therapy techniques have translated
to a reduction in radiation-induced cardiac diseases (Shafman et al 2005) as the follow-
up time since introduction of these techniques has been too short.

1.3.1.2. Lung

The main normal tissue complication expected in the lung is radiation-induced
pneumonitis, which occurs in approximately 1-5% of breast and chest wall patients
(Marks et al 2010, Blitzblau and Horton 2013). In our clinic, the main DVH parameter
used as a marker for the lung is the volume of the ipsilateral lung receiving at least 20
Gy (V20), which is targeted to be between 18 and 30%. If the IMN are included, the V20
should be preferably less than 35%.
1.3.1.3. Recurrences

Normal tissue may have a limited lifetime tolerance to radiation. Recurrences pose an additional constraint on these normal tissues as they have already been exposed to radiation in prior treatments. Thus while the main goal for recurrences is to treat the recurrent tumor, special attention may be placed on the brachial plexus and chest wall with clinical endpoints of reducing complications such as non-healing wounds and brachial plexopathy. Electrons are used because recurrences are often very superficial and electrons are less penetrating, thus providing good coverage to superficial targets, while sparing deeper tissues. Deeper recurrences are not treated with electrons.

While recurrences represent a relatively small patient population, investigation of improved techniques and technologies that can balance the trade-off between local control and normal tissue complications is warranted to improve the therapeutic ratio.

1.4. A brief history of electron therapy

Photon beam therapy has enjoy rapid advancements over the past few decades with efforts in physics, biological, and clinical research leading to advanced photon therapy techniques and accompanying technology that have solved many challenges and limitations previously imposed on the field. Conversely, advancements in electron beam therapy have been much slower, leading to its current meager participation in the radiation therapy modality space.
The following sections aim to briefly discuss the history of electron beam therapy and describe the current challenges and limitations facing electron beam therapy.

1.4.1. Challenges facing electron therapy

On one hand, electron beam interaction with media is advantageous for superficial tumors, as they drastically reduce their dose deposition after a certain depth, thus significantly reducing the dose to normal tissues, as well as maintaining a high dose to the surface. On the other hand, electrons suffer from a wide penumbra and must be collimated with additional collimators that must be placed close to the patient surface. In the clinical setting, electron beam therapy is often underutilized due to the lack of advanced planning and delivery tools, lack of proper training and technology, and advancements of other treatment modalities, such as intensity modulated radiation therapy (IMRT) and volumetric modulated arc therapy (VMAT). Photon beam therapy was at one point a crude technique and the rapid advancements leading to the developments of IMRT and VMAT were due to concerted research efforts and a considerable push for clinical translation. As such, these photon techniques have surpassed the efficiency and current clinical implementation of electron beam therapy. While photon beam therapy has increased its complexity by e.g. incorporation of the multi leaf collimators (MLCs) to improve target homogeneity and reduce normal tissue dose, implementation of more accurate planning algorithms such as the analytical...
anisotropic algorithm (AAA), and complex treatment techniques, electron beam therapy has remained practically unchanged for decades, and does not allow for the flexibility in planning, delivery, and verification that photon beam therapy enjoys. This has led to the much reduced role of electron beams in the clinic. Electron beam therapy rarely acts as the sole treatment modality mode and is mostly used in boost treatment concurrent to a photon beam therapy.

Electron beam therapy will remain underutilized and underdeveloped unless the radiation therapy community demands the proper technology and tools to aid translation to the clinic (Hogstrom and Almond 2006). Development of all aspects of the radiation therapy treatment chain including planning, delivery, and verification need to be addressed. Pursuing advanced electron beam therapy with dose distributions and outcomes that are equivalent or superior to current techniques may increase the use of electron beam therapy in the clinic.

1.4.2. Electron conformal therapy

Targets on a flat surface are typically treated with a single electron beam of a fixed energy. For a target consisting of large area along a curved surface, there are no satisfactory solutions. A single large electron beam will produce a non-uniform dose distribution inside the target and a wide penumbra at the target boundary. Clinical techniques such as single or multiple abutting static electron beams often show
heterogeneous dose distributions that suffer from undesirable hot and/or cold spots at the beam junction region (Figure 4). Electron conformal therapy is typically defined such that the target is contained within the 90% isodose line (Hogstrom and Almond 2006).

Figure 4. Static (a) single field and (b) two field delivery and resulting dose distribution on a curved target. The arrows indicate the wide penumbra generated in the single field delivery and the high dose at the beam junction causing dose heterogeneities for the two field case. These dose distributions were generated in the Varian clinical treatment planning system, Eclipse, using the eMC algorithm.
As stated previously, for customized field shaping lateral to the beam direction, Cerrobend cut-outs can be produced and placed in the bottom of the electron applicator. Further, for irregular treatment surfaces, a customized bolus consisting of a human tissue equivalent material can be used to compensate the irregularities and even out the surface, thus providing additional target conformity along the beam axis. Bolus also be increases the surface doses and reduces electron beam penetration. Bolus electron conformal therapy has been shown to provide good conformal electron therapy plans (Hogstrom and Almond 2006). An advantage of bolus electron conformal therapy is that it does not require modification to the treatment machine. Limitations of this technique include higher skin surface dose as well as the need to manufacture patient-specific bolus. The bolus can either be made manually in-house (requiring skilled staff), custom milled and ordered (Low et al 1992, Low and Hogstrom 1994, Low et al 1995, Kudchadker et al 2002, Kudchadker et al 2003), or more recently 3D printed (Su et al 2014).

Some examples of clinically used static fields are discussed: Electron fields for a chest wall scar boost often require the use of multiple patient-specific cut-outs, due to the extent of these scars. Usually large applicators (15x15 cm² and larger) must be used, and collision with the patient must be assessed. The margins for the scar are contoured by hand on the patient using a 2 cm margin. While tangential photon beams are
currently the dominant treatment modality for treatment of breast and chest wall, static electron beams as the sole modality for post-mastectomy chest wall irradiation has a long-running history, with some centers almost exclusively treating post-mastectomy patients with electron beams (Hardenbergh et al 1999, Feigenberg et al 2003). While conformal electron beam therapy for post-mastectomy chest wall is tedious, it has shown good locoregional and long-term disease control. Even with these findings, electron beam therapy remains a marginal treatment modality.

1.4.3. Advanced electron beam therapy techniques

The use of intensity and energy modulation is beneficial for conforming the beam to complex target shapes and reducing scattered dose and dose to normal tissue. Previous efforts to improve electron conformal therapy include electron arc therapy (EAT) and modulated electron radiation therapy (MERT).

1.4.3.1. Electron arc therapy (EAT)

has been primarily implemented clinically for the treatment of chest wall disease, since it spares the underlying lung tissue, unlike tangential photons. EAT is planned and delivered (Figure 5) such that the linac isocenter is positioned at a point inside the patient and with the electron cone removed to avoid collision, since the source-to-skin distance (SSD) is much smaller than 100 cm. To generate a narrow beam penumbra, secondary and tertiary collimators must be used. In addition, a shield or mold covering the entire patient surface must be made to improve the penumbra and protect non-target area.

Figure 5. A schematic of an EAT delivery.
The advantage of this technique is that for large treatment target area with
topographical variations, EAT can produce homogeneous dose distributions while
sparing surrounding and distal normal tissues. The disadvantage of this technique is
that it requires the labor-intensive manufacturing of patient-specific casts and
customized bolus to compensate for variable target thickness. Further, EAT requires
time-consuming planning and a rigorous physics support program, which is not
available at many clinics. Therefore, its clinical implementation was limited to only a
few clinics with expertise in this procedure. When these challenges are overcome, EAT
has shown clinically favorable outcomes compared to standard photon beam therapy
techniques for chest wall treatment (Gaffney et al 1997, Gaffney et al 2001, Gaffney et al
2003).

Substantial research has been undertaken in applying EAT to the treatment of
disease located on the chest wall. A 20-year experience published by Gaffney et al.,
summarizes the clinical outcome for 156 patients treated with EAT at the University of
Utah for post-mastectomy chest wall irradiation (Gaffney et al 2001). EAT was found to
provide excellent locoregional control for the post-mastectomy chest wall, while
covering the IMN without additional abutting fields, and minimizing the dose to lung
and heart.
1.4.3.2. Modulated electron radiation therapy (MERT)

Modulated electron radiation therapy (MERT) achieves conformal electron therapy by utilizing both energy modulation and intensity modulation (Hyodynmaa et al 1996, Lief et al 1996, Zackrisson and Karlsson 1996, Ebert and Hoban 1997, Karlsson et al 1998, Karlsson et al 1999). Dose conformity along the beam axis is achieved by modulating the electron energy, while a cut-out, scanned beam, or electron multi-leaf collimators (eMLC) can provide lateral dose constriction and intensity modulation. eMLCs replace the cut-out and provide variable apertures that must be manually driven or computer controlled. Material for eMLC construction can be steel, brass, or tungsten with typical leaf thickness of approximately 2-3 cm. Generally, the choice of material is a trade-off between cost, ease of machinability, and Bremsstrahlung production. One of the more advanced computer-controlled prototype eMLCs is shown in Figure 6 (Eldib et al 2013). This prototype eMLC consists of 2 cm thick tungsten leaves with straight leaf edges. The leaf width projected to isocenter is 6 mm.
Figure 6. A prototype eMLC consisting of 2 cm thick tungsten leaves with straight leaf edges. The leaf width projected to isocenter is 6 mm. (Image courtesy of Eldib et al. 2013 Phys. Med. Biol. 58 5653)

For most linacs, intensity modulation using the photon MLC without modification of the treatment head does not produce useful beam penumbra because of the large distance between the downstream edge of the MLC to the isocenter and the large extent of the virtual source to the dual scattering foil (Klein et al 2009b). The large thickness of the photon MLC and the rounded edge design also contributes significantly to the scatter downstream. Further, useful intensity modulation can be achieved only when the SSD is much reduced.

Monte Carlo simulation of MERT using eMLCs has been performed to investigate its characteristics (Ma et al 2000, Lee et al 2001a, Deng et al 2002, Ma et al 2003, ...
Al-Yahya et al 2005, Vatanen et al 2009, Alexander et al 2011). The use of eMLCs has remained mostly a research endeavor, and commercial eMLCs are currently only available through one company (Euromechanics GmbH, Schwarzenbruck, Germany). Thus the clinical implementation of eMLCs for MERT has lagged behind.

1.4.3.3. Scanning electron beam

A drawback of using a scattering foil is the production of Bremsstrahlung contamination, which is proportional to the energy and is typically about 1% for 6 MeV and 4% for 20 MeV electron beams for a dual scattering foil system. While Bremsstrahlung production occurs in the collimators, ionization chambers and air, the scattering foil is the largest contributor to X-ray contamination. Most modern linacs utilize scattering foils to produce large electron fields with uniform fluence. Another technique utilizes a scanning electron beam in place of a scattered beam to produce a beam large enough for clinical use. While not common, linacs capable of producing a scanning electron beam do not have scattering foils rather they utilize two computer-controlled magnets, which deflect the pencil beam in two orthogonal planes thus generating a constantly moving electron beam across the field. This system is advantageous as it does not introduce Bremsstrahlung contamination from the scattering foil, resulting in superior PDD curves. A disadvantage which lead to the ultimate failure of the aforementioned scanning beam technique is the complex
electronic system required to steer the electron beam. There have been some well
documented cases of this steering system failing, thus resulting in mistreatments of patients (Leveson and Turner 1993) and generally in the discontinuation of this technique.
1.5. Project focus and specific aims

The aforementioned discussion motivated the development of a new technique that seeks to address the questions posed by current limitations of advanced electron beam therapy. We aimed to produce a technique that can treat specific patient subsets with better outcomes than current techniques. Further, exploitation of current technological linac advancements was targeted thereby reducing the need for additional hardware. Additionally, we developed novel tools for planning, delivery, and verification of electron beam therapy. Thus, not only was foundational research on the development of a novel conformal electron therapy technique undertaken, but also new tools and techniques that could translate to current clinical electron beam therapy practice were also investigated.

The work presented herein aims to build upon previous research and contribute to novel solutions in the area of conformal electron beam therapy. This project comprises the development of a novel treatment modality, dynamic electron arc radiotherapy (DEAR). As such, every step of the radiation therapy treatment process must be investigated. The specific aims for DEAR are discussed below. These aims are listed in chronological order and represent the overarching goals for DEAR. Thus additional aims are listed that are beyond the scope of this dissertation, but are kept for the sake of completeness.
### Specific aim 1. Feasibility of dynamic electron arc radiotherapy (DEAR)

1.1. Evaluation and analysis of the feasibility of DEAR by investigation of preliminary DEAR design, delivery, and analysis methods

### Specific aim 2. Development of accurate dose calculation algorithms for DEAR

2.1. Validation of an electron Monte Carlo dose calculation framework on TrueBeam linacs
2.2. Evaluation of small field electron Monte Carlo for DEAR on TrueBeam linacs
2.3. Characterization of electron Monte Carlo phase space

### Specific aim 3. Development of treatment planning techniques for DEAR

3.1. Development of a platform to integrate Monte Carlo dose calculation, aperture optimization, energy modulation, and motion trajectory optimization for scanning beams
3.2. Application of the technique to selected clinical treatment sites

### Specific aim 4. Development of treatment verification techniques for DEAR

4.1. Development of accurate dosimetric verification techniques
4.2. Development of secondary and independent delivery verification of linac component motions

### Specific aim 5. Design of new hardware for improved DEAR capability

5.1. Investigation of multiple apertures in a single cut-out
5.2. Simulation of both static and dynamic modes for Motorized electron collimator (MEC) and Electron MLC (eMLC)
This dissertation constitutes foundational research for DEAR. The goal of specific aim 1, which is presented in Chapter 2, was to assess the feasibility of DEAR by evaluating DEAR deliverability and dosimetry by investigation of preliminary DEAR design, delivery, and analysis methods. Specific aim 2, which is detailed in Chapter 3, investigates the development of an accurate dose calculation algorithm for DEAR. Chapter 3 consists of three subsections: Firstly, the comprehensive validation of a Monte Carlo dose calculation framework for Varian TrueBeam linacs is presented (specific aim 2.1.). Secondly, this framework was extended to small electron fields and evaluation of small fields for DEAR was performed (specific aim 2.2.). Thirdly, tools for Monte Carlo dose calculations involving the characterization of the electron beam phase space file were developed (specific aims 2.3.). Specific aim 5, which is discussed in Chapter 4, presents the development of a new delivery technique using multiple apertures in a single cut-out to improve DEAR capability. Chapter 5 comprises future work and additional preliminary investigation of the application of DEAR for specific treatment sites (specific aim 3) and development of accurate delivery and dosimetric verification techniques for DEAR (specific aim 4).
2. Feasibility of dynamic electron arc radiotherapy (DEAR)

2.1. Introduction

As stated in section 1.4.1, compared to other radiation therapy modalities, clinical electron beam therapy has remained practically unchanged for the past few decades (Hogstrom and Almond 2006) even though electron beams with multiple energies are widely available on many linacs. The physical principles of electron interaction with matter make electrons well-suited for treatment of superficial tumors (<5 cm deep), as they deposit most of their energy within a few centimeters from the surface (Podgorsak and IAEA 2005). Electron beams in the range of 6 – 20 MeV are therefore advantageous for treatment of shallow disease with critical organs and normal tissues at the distal edge, such as the head and neck, skin and lip, chest wall, and extremities. However, as previously shown, challenges and limitations exist in electron beam therapy.

Recent advancement in linac technology and robotic feedback control has made many sophisticated radiation therapy techniques possible. Examples are dynamic photon therapies such as IMRT via dynamic MLCs at multiple static gantry angles, and more recently VMAT which delivers intensity modulated photon beams in an arc fashion to create highly conformal three dimensional dose distributions. Even more advanced techniques are possible, but rigorous research and extensive validations are necessary before clinical implementation. The TrueBeam linac (Varian Medical Systems,
Inc., Palo Alto, CA) provides a special Research Mode, for non-clinical use, which can be used for testing advanced delivery techniques. Research Mode is driven by treatment plan files in XML (extensible markup language) format. This allows users to design and deliver complex non-standard plans, imaging, and gating techniques with arbitrary motion trajectories in MU-Position space with high temporal resolution. The delivery is recorded in a trajectory log file for subsequent analysis.

The goal was to develop a novel electron beam therapy technique which aims to improve upon previous electron beam therapy research. We proposed a novel technique for conformal electron beam therapy called dynamic electron arc radiotherapy (DEAR). In DEAR, radiation is delivered in arc mode with the electron applicator kept in place. The couch motion is synchronized with gantry rotation to avoid collisions between patient and the linac during treatment delivery. Unlike EAT or MERT, DEAR does not require extraneous collimation or patient-specific shielding equipment. The following will describe design, delivery, and analysis of DEAR on simple cylindrical phantoms as a proof-of-concept of this technique. DEAR deliverability was investigated and its potential was demonstrated to improve dose distributions in comparison to static field electron beam techniques (Rodrigues et al 2014).
2.2. Materials and Methods

In DEAR, radiation is delivered while gantry rotation and dose rate are modulated. The electron applicator and cut-out provide the lateral constriction of the beam and are therefore able to maintain the penumbra. The couch motion also permits target area larger than the size of applicator to be treated. The combined motion of couch and gantry is such that the electron beam is always normally incident on the surface at a constant SSD of 100 cm, therefore avoiding problems of oblique incidence found in fixed electron field techniques (Ekstrand and Dixon 1982, Biggs 1984). Normality of the electron beam to the surface ensures maximum penetration to the therapeutic depth and uniform penumbra. Combining these aforementioned constraints, a technique can be generated where the linac isocenter is not a single point inside the patient, but forms a continuous line along the surface. Figure 7 illustrates a simplified “chest wall” DEAR delivery as seen from the patient view and the room view. The patient view represents the coordinate system that moves with the couch, i.e. the patient is stationary in patient view. In room view, motions of both the gantry and couch (patient) are visible.
Figure 7. Example of DEAR delivery as seen from the (a) patient view and the (b) room view in an axial plane. The semi-circle represents the patient skin with target area (blue) on the surface extending over an arc. The red dots represent the linac isocenter at the start and stop of the arc. As the radiation is delivered in an arc, the couch is moved synchronously with gantry motion, such that the beam is always normal to the surface and maintains a constant SSD. In patient view, the isocenter forms a continuous line along the surface, while in the room view the couch motion in lateral and vertical as well as the gantry rotation about the machine isocenter is visible.

To explore the feasibility of DEAR, plans for cylindrical water equivalent phantoms (25 cm and 32 cm diameter) were designed and delivered for various energies and cut-out sizes. This simple geometry allowed us to understand the deliverability accuracy and dosimetry of DEAR. The following sections discuss in detail several steps of this study: plan design, delivery, and analysis, which are summarized in the work flow schematic in Figure 8.
Figure 8. DEAR study work flow. Dose kernels created in the TPS are exported via the DICOM protocol to MATLAB where a DEAR plan is created. The plan (defined in extensible Markup Language (XML)) is loaded onto the TrueBeam linac in Research Mode, and the delivery is recorded in the trajectory log file and on radiographic film for subsequent analysis.

2.2.1. DEAR plan design

DEAR plans were generated in three steps: (1) Dose kernel creation, (2) DEAR plan creation, and (3) Static to dynamic beam conversion to obtain 2D axial dose distributions.

The first step in creating DEAR plans is to generate accurate and high resolution dose distributions for single static electron fields on the cylindrical phantom. As a first approximation, dose distributions computed using the electron Monte Carlo (eMC) dose calculation algorithm in the Varian Eclipse Treatment Planning System (TPS, v.10) are used. A dose calculation grid size of 1 mm in all dimensions and endpoint for dose
accuracy of 1% was chosen (Popple et al 2006b, Xu et al 2009). Dose kernels for a combination of two Cerrobend cut-outs of size 1x10 cm$^2$ and 3x10 cm$^2$ within a 15x15 cm$^2$ electron applicator (Figure 9) and two electron energies (6 MeV and 9 MeV) were generated.

![Figure 9. Cerrobend cut-outs 1x10 cm$^2$ (left) and 3x10 cm$^2$ (right).](image)

In this feasibility study, dose calculations were performed using the clinically available dose eMC algorithm in the Varian Eclipse treatment planning system. The dose kernels are exported via the RT Plan and RT Dose DICOMs to an in-house software and processed for symmetry and smoothness in the MATLAB computing environment.
In this study, DEAR was simulated as a sequence of multiple static beamlets, as has been previously shown for EAT (Khan et al 1977, Lam et al 1987). This superposition of weighted dose kernels leads to the calculated dose distributions for the rotational therapy. However, unlike EAT, the fields in DEAR do not have the common isocenter of the treatment machine, but rather have a simulated isocenter line that arises from the combination of the gantry and couch motion for a given cylindrical diameter e.g. a cylindrical phantom of 30 cm diameter will have a common equivalent isocenter of 115 cm for all fields, with the isocenter at SAD following a line along the phantom surface. It follows that a realistic patient geometry will not have a common equivalent isocenter for all fields. The superposition algorithm for all beamlets is summarized in Figure 10.
Figure 10. Superposition of $N$ beamlets on cylindrical phantom with radius $r_p$. Each beamlet is described by gantry angle $\theta_i$ and beam weight $w_i$ (or $MU_i$) where $i$ is the beamlet index from 0 to $N$. A point in the target is defined by the coordinates $r_j$ and $\alpha_j$, where $j$ is the index for all points in the target from 1 to $M$.

A point in the target can be described in polar coordinates relative to the center of the phantom. The dose for a point in the target $D_j(r_j, \alpha_j)$ is expressed in Equation 1. Gantry angle $\theta$ is defined with respect to the machine isocenter. Since the dose kernel $d$ is defined at gantry angle $\theta = 0^\circ$, the term $d_i(r_j, \alpha_j - \theta_i)$ gives the dose $D_j$ at the $j^{th}$ point in the target $(r_j, \alpha_j)$ for the $i^{th}$ beam.
\[ D_j(r_j, \alpha_j) = \sum_{i=1}^{N} w_i d_i(r_j, \alpha_j - \theta_i) \]  

Equation 1

where \( j \) in the index for all points in the target from 1 to \( M \), and \( i \) is the index for all beams from 1 to \( N \). The \( w_i \) is the beam weight, or MU.

The 1D beam weight optimization was performed to investigate improvement in dose distribution. Equal beam weighting was determined as the ratio of the prescription dose \( D_0 \) and the mean target dose. Optimal beam weighting was realized through a gradient search algorithm (Wu and Mohan 2000). Subsequently, the software converts this plan of multiple static beamlets to a dynamic arc plan in XML format for delivery. Each static beamlet serves as a control point (CP). The conversion scheme is described in Table 1 in conjunction with Figure 10.
Table 1. The static beamlets to dynamic arc conversion scheme: There are $N$ static beamlets, each delivers a certain $MU_i$. The cumulative $MU$ index at each control point, $MU_{CP}$, is then given by the formula below.

<table>
<thead>
<tr>
<th>CP</th>
<th>Beamlet</th>
<th>Angle</th>
<th>$MU_{CP}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>0</td>
<td>$\theta_1$</td>
<td>0</td>
</tr>
<tr>
<td>1</td>
<td>1</td>
<td>$\theta_1$</td>
<td>$\frac{MU_1}{2}$</td>
</tr>
<tr>
<td>2</td>
<td>$1+2$</td>
<td>$\theta_2$</td>
<td>$MU_1 + \frac{MU_2}{2}$</td>
</tr>
<tr>
<td>...</td>
<td>...</td>
<td>...</td>
<td>...</td>
</tr>
<tr>
<td>$N$</td>
<td>$(N-1)+N$</td>
<td>$\theta_N$</td>
<td>$\sum_{i=1}^{N-1} MU_i + \frac{MU_N}{2}$</td>
</tr>
<tr>
<td>$N+1$</td>
<td>$N$</td>
<td>$\theta_N$</td>
<td>$\sum_{i=1}^{N} MU_i$</td>
</tr>
</tbody>
</table>

2.2.2. DEAR plan delivery

2.2.2.1. Varian TrueBeam Research Mode

DEAR plans are delivered in Research Mode on a TrueBeam linac. Research Mode is a relatively new non-clinical mode that allows for non-clinical and/or novel treatment and imaging techniques (Ling et al 2011) expressed in the XML format to be delivered. Rather than communicating a plan in the DICOM format, users can directly script their plans in XML, which is the machine-internal language. Due to its repetitive nature, various tools were developed to aid the writing of DEAR plans.
In this study, the intended trajectory is defined as the ideal trajectory specified in the XML file, which may not consider the linac limitations. After the XML file is loaded into the TrueBeam, the expected trajectory is determined from the intended trajectory while taking into account the machine limitations. This expected trajectory is defined as the “MU versus Position Trajectory Model” which treats the positions of all mechanical axes as a function of MU. The trajectory is defined by a finite number of discrete control points (CPs) with the segment between two successive control points being linear. Since the position on every axis is related to MU, the relative position between consecutive CPs can also be determined. Therefore, during delivery each axis follows a predictable and reproducible path as specified by the trajectory model. Delivery is regulated by the computer feedback control system in real-time with high temporal resolution of 10 ms. Further, every 20 ms during delivery, the computer system records the actual trajectory for all axes values to a trajectory log file. This log file also includes the expected trajectory as well, such that delivery can be retrospectively reconstructed and analyzed.

2.2.3. DEAR Analysis

DEAR feasibility was analyzed in terms of deliverability and dosimetry accuracy.

2.2.3.1. Deliverability

For reference, Figure 11 shows a TrueBeam linac with pertinent couch and gantry motion axes investigated in this study.
Figure 11. Photograph of a TrueBeam linac with motion axes for gantry, collimator, and couch indicated. The machine isocenter defines the origin of the coordinate system.

The absolute values for the motion axes on the TrueBeam linac can be defined by the Varian IEC 60601-2-1 or Varian 61217 scales. The gantry rotates about the machine isocenter. $0^\circ$ is defined when the gantry head is vertical and perpendicular to the floor.
with the beam towards the floor. The angles increase in the clockwise direction. Further, the gantry can rotate in both directions. The collimator rotates about its own center. The couch has four motion axes: Couch rotation, longitudinal, lateral, and vertical motions are possible.

2.2.3.1.1. Analysis of trajectory log file

Deliverability was evaluated by verifying the accuracy of the MU, gantry angle, and couch positions: The agreement between expected and actual axes values recorded in the trajectory log during delivery was quantified by the root mean square error (RMSE) given by the Equation 2:

\[
RMSE = \sqrt{\frac{1}{n} \sum_{k=1}^{n} (y_k - \hat{y}_k)^2}
\]

with \(y_k\) and \(\hat{y}_k\) being the expected and actual axes values, respectively. The velocity is calculated by taking the numerical derivative from the trajectory by the central difference method and smoothed with a moving average kernel.

2.2.3.1.2. Determination of maximum velocity

Since DEAR utilizes the simultaneous motion of multiple mechanical axes (couch, gantry, and collimator), the maximum velocities of these axes were measured using the trajectory log file. To obtain the maximum velocity, each axis was investigated
independently. XML files were designed to hold all mechanical axes constant, and only vary one axis at a time. During delivery, the axis of interest was asked to cover a large distance/rotation while a small fixed number of MU was delivered at the highest dose rate. In the example shown in Figure 12, the couch was asked to move vertically by 10, 20, 30, 32.5, 35, 37.5, 40, and 50 cm (corresponding to eight different scenarios 1 – 8, respectively) while 250 MU of a 6 MeV beam was delivered at a dose rate of 1000 MU/min for each of these scenarios. As the distance the couch must travel increases, the couch velocity increases (scenario 1: 0.67 cm/s, scenario 2: 1.34 cm/s, and scenario 3: 2 cm/s), as can be seen for both the intended and actual couch velocity. For scenario 4, the intended couch velocity is 2.17 cm/s, however, since the couch cannot move faster than 2 cm/min, the dose rate drops from the intended 1000 MU/min to the actual 923 MU/min such that dose of 250 MU can be delivered in time. Scenario 5, 6, 7, and 8 continue to show this pattern.

The mechanical axes velocities are inherently limited by the design, and can therefore not exceed a certain threshold. As a consequence, the linac control system lowers the dose rate, such that the expected trajectory model can be followed. Therefore, observation of a reduced actual dose rate (given in the trajectory log file) from the intended dose rate (given by the XML file) during delivery indicates that the maximum mechanical axes velocity has been reached. Measurements for each axis were repeated multiple
times to find their maximum velocity and compared to the manufacturer’s specifications.

Figure 12. An example of determining the maximum axis velocity. The dose rate is shown in red according to scales on left and the couch vertical velocity in blue to scales on right. The solid lines with circles denote the actual delivery, while the dotted lines with triangles denote the intended delivery. As long as intended delivery is within the mechanical capabilities of the machine, the delivery will follow the intended trajectory (i.e. circles and triangles coincide). However, when the intended couch vertical velocity exceeds the actual maximum couch vertical velocity, the dose rate must drop to compensate.

2.2.3.1.3. Dose rate modulation

Dose rate modulation is a feature that distinguishes the DEAR technique from EAT and conventional static electron beam therapy. It is commonly employed in the dynamic photon therapy such as IMRT and VMAT. It was investigated for a 6 MeV
electron beam and compared to a 6 MV photon beam. Dose rates from 600 to 50 MU/min in increments of 50 MU/min for 10 s for each constant dose rate level were coded in an XML file and delivered. The expected and actual dose rates were then extracted from the trajectory log file. Data for each constant dose rate level was down-sampled to 0.5 s intervals and the standard deviation of these points over that level were computed and compared.

2.2.3.1.4. Gantry velocity modulation

Another feature of DEAR that is different from EAT is the capability of modulating gantry rotation speed. This was investigated qualitatively by requesting a change of gantry rotation speed in a single XML file.

2.2.3.1.5. Motions in multiple axes

The complexity of DEAR was explored by introducing motions along multiple mechanical axes during delivery. Test plans included sweeping beam, beam hold, collimator motion, and additional couch motions in vertical, lateral, longitudinal, and rotational directions.

2.2.3.2. Dosimetry

One goal of the feasibility study was to achieve penumbra for DEAR on a curved surface comparable to that of a static electron field on a flat surface. The penumbra was defined as the physical distance between the 80% and 20% dose level for an orthogonal
beam profile. For example, a static 15x15 cm$^2$ field on a flat surface result in a penumbra of approximately 1.2 cm in the in-plane and cross-plane direction for energies of 6 and 9 MeV at the depth of maximum dose.

Experimental dosimetry using radiographic film (EDR2, Carestream Health Inc., Rochester, NY) was evaluated for static beam and DEAR plans on a 22 cm diameter cylindrical phantom (Delta4, Scandidos, Uppsala, Sweden) with 1.5 cm bolus (Superflab, CNMC Company Inc., Nashville, TN) wrapped radially around the phantom, as shown in Figure 13.

Figure 13. Set-up for the film irradiation: The radiographic film (in its jacket) was taped to the Delta4 and 1.5 cm of bolus were placed on top.
The static beam plan used a 16x10 cm$^2$ Cerrobend cut-out in a 20x20 cm$^2$ electron applicator (Figure 14).

![Figure 14. 16x10 cm$^2$ cut-out made for a 20x20 cm$^2$ applicator.](image)

The DEAR plan used a 3x10 cm$^2$ cut-out in a 15x15 cm$^2$ electron applicator. Both films were irradiated with 6 MeV electron beams delivering 350 MU at dose rates of 500 and 400 MU/min, respectively. The static beam plan was delivered at a gantry angle of 0° and the DEAR plan in an arc from gantry angle 315° to 45°. Subsequently, the film was processed and digitized using a VXR-16 Dosimetry Pro Scanner (VIDAR Systems Corporation, Herndon, VA). The film calibration (dose vs. optical density) was performed according to TG-25 recommendations (Khan et al 1991b, Gerbi et al 2009). The measured in-plane and cross-plane beam penumbra (20 – 80 %) were compared to the
calculated penumbra from the Eclipse TPS using the eMC algorithm v.10 at a depth of 1.5 cm. Because cross-plane penumbra was evaluated at the target depth i.e. a given radius in the cylindrical phantom, the planned profiles in the cross-plane direction must be “unwrapped” from the cylindrical geometry (Apostol and Mnatsakanian 2007) prior to analysis, such that it could be compared to the film. The unwrapping of the calculated dose matrix was achieved by re-gridding the data from a Cartesian coordinate system ($x, y$) to a polar coordinates ($r, \theta$) in the axial plane.

Since film and calculation results agreed well and to expand our understanding of the dosimetric qualities of DEAR, more calculations were performed for various static, abutting fields, and DEAR plans for two electron beam energies 6 and 9 MeV. Plans were calculated for a 32 cm diameter cylindrical phantom and were designed to irradiate the same target area on the phantom surface. The target dimension extended in a 90° arc from gantry angle 315° to 45° in the radial direction. The target depth was chosen to be at either 1 cm (6 MeV) or 2 cm (9 MeV). The target dose was set to 200 cGy. The irradiation set-ups are illustrated in Figure 15.
Figure 15. Phantom geometry and set-up for single-field, two-field, and DEAR (patient view) 9 MeV plans, as viewed in the depth and cross-plane view, with the in-plane direction perpendicular to the page. A representative axial dose distribution for each set-up is shown below. The dashed line represents the target depth for dosimetry analysis.

The single-field plan used a 23x10 cm² cut-out and a 25x25 cm² electron applicator at a gantry angle of 0°. The two-field plan used two 13.5x10 cm² cut-outs for a 15x15 cm² electron applicator at a gantry angle of 335° and 25°. The two fields abutted on the surface, as is typically done for clinical treatment of superficial tumors. The
DEAR plans were designed for equal and optimal beam weightings and two cut-out sizes (1x10 cm² and 3x10 cm²) and a 15x15 cm² electron applicator.

In-plane and cross-plane penumbra, dose homogeneity, and dose area histograms (DAH) were evaluated at target depths for the specified target regions, i.e. for the “unwrapped” 2D dose distribution at target depth, the DAH was found for the region bounded by the gantry angles from 315° to 45° and the in-plane direction (inferior-superior) of 10 cm. Dose homogeneity was defined as the percentage dose variation for the region encompassing 80% of the field size. The field size for a given profile was defined as the full width at half maximum (FWHM).

2.3. Results

2.3.1. Deliverability

2.3.1.1. Analysis of trajectory log file

Trajectory log analysis of expected and actual axes values for DEAR plans had small RMSE (<0.1 MU, <0.1° gantry, <0.1 cm couch positions). Figure 16 shows a trajectory log plot of expected and actual axes values for a single arc DEAR delivery. During this DEAR delivery, the gantry rotated 90° from 315° to 45° at a constant rate over the delivered MU, while the couch moved vertically from 0 cm, down to -5 cm, and back to 0 cm and laterally from 0 cm to -23 cm.
Figure 16. Trajectory log plots of expected and actual axes values for gantry angle, couch vertical, and couch lateral (a) Position versus MU, and (b) Velocity versus time.

For gantry rotation, the speed (vs. time) is constant except at the beginning and the end of the arc where the inertia of the gantry causes less smooth motion. It is important to note that the relation of position vs. MU specified in the XML file is maintained, however, not necessarily the speed as a function of the time. The intended trajectory agrees very well with the expected and was not shown in the plot.

Further, realistic delivery time can be estimated by the following. For a patient body of 30 cm diameter and a target area spanning over a 90° arc, the target length is approximately 25 cm. Using a 3x10 cm² slit cut-out, this would require an equivalent of 8
static gantry positions. With a prescription dose of 200 cGy per fraction, the total MU for the arc is approximately 1600 MU. At the highest dose rate of 1000 MU/min, the treatment time would be 1.6 minutes or 100 seconds. The gantry rotation speed is then approximately $90°/100\,s = 0.9\,°/s$, couch speed in lateral and vertical are 0.2 and 0.05 cm/s, respectively, to give a total couch speed of approximately 0.22 cm/s, which is well below its maximum speed shown in the next section.

2.3.1.2. Maximum velocities for couch, gantry, and collimator axis

Table 2 compares manufacturer specifications to measured maximum velocities. Measured maximum velocities agreed with manufacturer specifications and were 4.0, 2.0, and 8.0 cm/s for couch lateral, vertical, and longitudinal, respectively; and 3.0, 6.0, and 15.0 °/s for couch rotational, gantry, and collimator, respectively. Uncertainties in the measured velocities were on the order of 0.1 cm/s and 0.1°/s. Velocities in all of our delivered DEAR plans ranged from 0.1 – 1.1 cm/s, 0.1 – 1.7 cm/s, and 0.3 – 6 °/s for couch lateral, vertical, and gantry, respectively.
<table>
<thead>
<tr>
<th>Motion Axes</th>
<th>Maximum Velocity (Manufacturer Spec)</th>
<th>Maximum Velocity (Measured)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Couch Lateral (cm/s)</td>
<td>4.0</td>
<td>4.0 ± 0.1</td>
</tr>
<tr>
<td>Couch Vertical (cm/s)</td>
<td>2.0</td>
<td>2.0 ± 0.1</td>
</tr>
<tr>
<td>Couch Longitudinal (cm/s)</td>
<td>8.0</td>
<td>8.0 ± 0.1</td>
</tr>
<tr>
<td>Couch Rotation (°/s)</td>
<td>3.0</td>
<td>3.0 ± 0.1</td>
</tr>
<tr>
<td>Gantry Rotation (°/s)</td>
<td>6.0</td>
<td>6.0 ± 0.1</td>
</tr>
<tr>
<td>Collimator Rotation (°/s)</td>
<td>15.0</td>
<td>15.0 ± 0.1</td>
</tr>
</tbody>
</table>

2.3.1.3. Dose rate modulation

Dose rate modulation results are shown in Figure 17. The actual dose rate agreed very well with the intended ones at all dose rate levels, with some fluctuations occurring especially during the transition between dose rates. Larger variations were found for electron beam than photon beams, which is clearly shown in the zoomed-in portions of the plots for the dose rate level of 200 MU/min. For photon beams, the percent standard deviation was between 0.01 to 0.28% for dose rates of 600 to 100 MU/min. For electron beams, the percent standard deviation was between 0.26 to 1.4% for dose rates of 600 to 100 MU/min. The average standard deviation over all dose rates was found to be 1.6 and 0.2 MU/min for the electron and photon dose rate, respectively. This demonstrates that dose rate modulation is feasible during DEAR delivery. The larger variation for
electrons than photons may be due to the fact that the linac is usually tuned more precisely for photons. It is important to note that the current maximum dose rate for electron beams on the TrueBeam is 1000 MU/min. A dose rate of 600 MU/min was chosen in this example as a comparison with flattened photon beam whose maximum dose rate is 600 MU/min.

Figure 17. Intended (red) and actual (blue) dose rate for (a) 6 MeV electron and (b) 6 MV photon beam. The inserts are the zoomed section for dose rate of 200 MU/min.

2.3.1.4. Gantry velocity modulation

Gantry rotation modulation is shown in Figure 18, where changes in gantry rotation velocity were requested during delivery.
2.3.1.5. Additional axes motion

Further, the complexity of DEAR plans can be increased by including motions in multiple axes. For example, a scanning-type beam was delivered as shown in Figure 19. The scanning pattern is realized by the delivery of two symmetric arcs (arc 1: 315° to 45°, arc 2: 45° to 315°) with a longitudinal (inferior-superior direction) couch shift by the size of aperture between arc 1 and 2. Therefore, before the delivery of the second arc, the couch moves longitudinally while the beam is held off. This shows that a beam hold can be used to accommodate axis motion if necessary.
Figure 19. Trajectory logs for expected and actual dual arc DEAR delivery showing (a) gantry angle and (b) lateral (blue dotted), vertical (black dotted), and longitudinal (red dashed) couch position versus time. (c) Zoomed-in plots of couch longitudinal (red dashed) and MU (black dotted) versus time. The end of the first arc occurs at about 250 s. At this time, the couch moves longitudinally while the beam is held. The blue lines indicate the time span for which the beam is held. During this time, there is no beam delivery (i.e. no accumulation of dose). After the couch has moved into position, the delivery of the second arc starts.

2.3.2. Dosimetry

Figures 20 and 21 show dosimetric comparisons of the calculated and measured dose distributions for a static plan and a DEAR using a 3x10 cm² cut-out plans for 6 MeV
at the target depth of 1.5 cm depth. Both in-plane and cross-plane profiles showed good agreement for the penumbras (within 1 mm) between calculation and measurement. The in-plane penumbras for the static and DEAR delivery are similar, while the cross-plane penumbra for the static delivery is about 4.3 cm compared to 2.1 cm for the DEAR delivery. This suggests possible tighter margins for the target region as well as better dose uniformity in DEAR. The target dose homogeneity was within ± 2% for the DEAR plan, while the static plan displayed variations of dose of up to 30%.
Figure 20. Calculated and measured dose distributions at a depth of 1.5 cm for (a) the static and (b) the DEAR delivery with a 3x10 cm$^2$ cut-out. An energy of 6 MeV was used for both the static and DEAR delivery. The distributions are normalized to the maximum dose, respectively. Cross-plane and in-plane penumbra values are shown in cm. The in-plane penumbra for both plans are similar, but the cross-plane penumbra is markedly narrower for the DEAR plan.

Figure 21. In-plane and cross-plane profiles for the (a) static and (b) DEAR delivery.
As can be seen in Figures 20 and 21, the film measurements are noisy due to the bolus and film jacket bunching up and trapping small air pockets around the film. The increase in dose at the edges of the cross-plane profile of the film measurement for DEAR delivery is an artifact of the static to dynamic beam conversion with limited angular resolution $\Delta \theta$ in the static beamlets of $5^\circ$. A higher angular resolution should reduce such effect.

Since film measurements and calculations agreed well, additional calculations were evaluated for 6 and 9 MeV single-field, two-field, and DEAR plans. Figure 22 and 23 show the dose distributions for static single-field, two-field, and DEAR plans with equal and optimal beam weighting at depths of 1 and 2 cm for electron beam energies of 6 and 9 MeV, respectively. The in-plane penumbra did not vary much between plans. The cross-plane penumbra for all plans is given in Table 3. For the DEAR plan the smaller cut-out size of $1 \times 10$ cm$^2$ gives narrower cross-plane penumbra than the $3 \times 10$ cm$^2$ cut-out.
Figure 22. Dose distributions for 6 MeV plans normalized to maximum dose. (a) Static delivery (single field and two-field plans), (b) the DEAR delivery with 1x10 cm² cut-out (equal and optimal beam weightings (BW)), and (c) the DEAR delivery with 3x10 cm² cut-out (equal and optimal BW). The in-plane and cross-plane penumbra are in cm.

Figure 23. Dose distributions for 9 MeV plans normalized to maximum dose. (a) Static delivery (single field and two-field plans), (b) the DEAR delivery with 1x10 cm² cut-out (equal and optimal beam weightings (BW)), and (c) the DEAR delivery with 3x10 cm² cut-out (equal and optimal BW). The in-plane and cross-plane penumbra are in cm.
Table 3. Cross-plane penumbra in cm for all calculated plans with beam energies of 6 and 9 MeV at 1 cm and 2 cm depth, respectively.

<table>
<thead>
<tr>
<th>Plan</th>
<th>6 MeV</th>
<th>9 MeV</th>
</tr>
</thead>
<tbody>
<tr>
<td>Single Field (23x10 cm²)</td>
<td>5.0</td>
<td>5.6</td>
</tr>
<tr>
<td>Two-Field (13.5x10 cm²)</td>
<td>1.5</td>
<td>1.7</td>
</tr>
<tr>
<td>DEAR (1x10 cm²)</td>
<td>1.4</td>
<td>1.6</td>
</tr>
<tr>
<td>DEAR (3x10 cm²)</td>
<td>2.0</td>
<td>2.1</td>
</tr>
</tbody>
</table>

As expected, the single-field displayed the worst penumbra (> 5 cm). While the two-field’s penumbra was better, it exhibited a hot spot of up to 60% higher than the target mean at the beam junction covering approximately 20% of the target. If the fields are matched on the surface, a hot spot will develop at deeper depths; however, if the matching occurs at a deeper depth e.g. target depth, the surface will develop cold spots. DEAR plans displayed homogenous dose distributions within ± 2%. Please note that these results were only for the dose distribution at target depth. The percent variation about the mean target dose is shown in Table 4.
Table 4. Percent variation in dose about the mean target dose at target depth of 1 or 2 cm for 6 or 9 MeV, respectively.

<table>
<thead>
<tr>
<th>Energy (MeV)</th>
<th>Two-Field</th>
<th>DEAR Equal BW</th>
<th>DEAR Optimal BW</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>± 60%</td>
<td>± 1% (1x10)</td>
<td>± 1% (1x10)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>± 1% (3x10)</td>
<td>± 1% (3x10)</td>
</tr>
<tr>
<td>9</td>
<td>± 57%</td>
<td>± 2% (1x10)</td>
<td>± 1% (1x10)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>± 1% (3x10)</td>
<td>± 1% (3x10)</td>
</tr>
</tbody>
</table>

The dose area histogram (DAH) for all plans is shown in Figure 24. Similar trends were observed for all plans. The coverage criterion was chosen as 90% of the target receiving at least 90% of the prescription dose. The single-field failed to achieve the coverage criterion. The two-field technique passed the coverage criterion, however, it displayed a high-dose tail (>100%) corresponding to the hot spots in the field junction. The DEAR plans passed the coverage criterion and produced optimal coverage without hot/or cold spots.
2.4. Discussion

In this study, the feasibility of DEAR on a Varian TrueBeam was demonstrated. DEAR represents a potential new electron beam therapy technique with some advantages over standard electron radiotherapy techniques. The smooth dose

Figure 24. Dose Area Histogram (DAH) for all plans: DEAR plans for 6 MeV electron beams with (a) 1x10 cm$^2$ and (b) 3x10 cm$^2$ cut-out, and DEAR plans for 9 MeV electron beams with (c) 1x10 cm$^2$ and (d) 3x10 cm$^2$ cut-out and their respective single and two-field plans.
distributions achieved on the cylindrical phantom with DEAR are better than those from static single and multiple-field electron plans in terms of penumbra and uniformity.

DEAR maintains narrow beam penumbra using readily available standard electron applicators and cut-outs as tertiary collimation, unlike EAT which requires the fabrication of patient-specific casts and bolus, which are time consuming and involve skillful craftsmanship (McNeely et al 1988). This may make it easier to implement for general clinical use. Compared to MERT, DEAR preserves normal direction to the skin, thereby producing smooth uniform dose distributions even for curved surfaces during delivery, which is not possible with MERT. Furthermore, DEAR offers capabilities beyond EAT and MERT. In DEAR all motion axis are controlled by the computer system with high temporal feedback control including triggering of beam holds if necessary. The recorded trajectory log files can be used to reconstruct treatment delivery and provide enhanced quality control.

The most innovative feature of DEAR is the simultaneous motion of multiple mechanical axes such as couch and gantry. Having the couch move during treatment may cause some concerns about safety and secondary patient motion; however, techniques such as dynamic radiosurgery (Podgorsak et al 1988), which involve synchronous motion of gantry and couch during delivery, have been implemented clinically for many years. Since the motion is predetermined by the trajectory model
according to patient geometry, no unpredictable couch motion will occur. Implementation of independent quality assurance (Woo and Kim 2002, Yang et al 2011) will ensure the reproducible and accurate couch motion during DEAR delivery. As such, we do not foresee dynamic couch motion during delivery being a burdensome issue. Of course, comprehensive pre-treatment QA and dry-run simulations must be carried out before clinical implementations.

Couch positioning accuracy is assessed monthly and yearly as part of routine quality assurance of the linac as outlined in TG-40 (Kutcher et al 1994) and TG-142 (Klein et al 2009a). Couch position tolerance should be within 2 mm/1° (Non-IMRT or IMRT) and 1 mm/0.5° (Stereotactic radiosurgery or stereotactic radiation therapy) for the translation (vertical, longitudinal, lateral) and rotation axes, respectively. Since DEAR utilizes concurrent motion of multiple couch axes during beam delivery, the accuracy of couch motion needs to be assessed as well. Hysteresis of the couch motion due to positive and negative motions (e.g. couch vertical up or down) and varying table loads must be assessed in future studies.

It is important to note that only the cylindrical phantom was used in our current study to verify the delivered dose with film, due to the available resources and simplicity. However, the DEAR design in the radial direction is not limited to cylindrical geometries, but can easily be generalized for arbitrary patient surface shapes. Figure 25
(a) shows an axial CT slice though the chest of a patient overlaid with the body contour extracted from the treatment planning system. DEAR imposes two constraints on the beam: 1) The beam is always at a constant distance from the surface (e.g. 100 cm), and 2) the beam is always perpendicular to the surface. The range of the arc is defined by a start and stop gantry angle that encompasses the target region. The static beamlets are positioned equidistant from one another. This defines the spatial coordinates of the beams. Figure 25 (a) shows the extent of an arc e.g. 90° and the static beams spaced by an angular resolution of 5°. Constraint 2) is imposed by finding the gantry angle at each beam location for which the beam is perpendicular to the surface. The final static beam set-up is shown in Figure 25 (b).

It is the combination of the couch position and gantry angle that can attain the constraints for the DEAR plan for any patient surface geometry. The motion axis trajectories and gantry position for such a plan are shown in Figure 25 (c) - (e). Note that the combined motion of the couch vertical and lateral (Figure. 25 (f)) follow the contour of the chest (Figure 25 (a)).
Figure 25. Example of DEAR for chest wall irradiation: (a) A CT slice through a chest with the contour of patient body (blue), DEAR static beam selection (red), and arc (white). (b) DEAR final beam arrangement. (c) Couch lateral position versus control point (CP). (d) Couch vertical position vs. CP. (e) Gantry angle vs. CP. (f) Couch vertical position vs. lateral position.
In this 2D study, surface variations in the inferior-superior direction were not investigated. There are several options to compensate for such variations. For example, a trapezoid cut-out can be used instead of rectangular slit field, as implemented in the EAT. Another option is to employ the combination of the couch longitudinal motion and smaller cut-out to realize a scanning beam operation. Alternatively, the cut-out size could be changed dynamically with the motorized electron collimators or multiple aperture in a single cut-out could be used. Investigation of multiple apertures in a single cut-out as a novel patient-independent collimation system is addressed in Chapter 3.

Higher electron energies (>16 MeV) for DEAR may have limited advantages for clinical use, due to their increased bremsstrahlung component. Reduction of the secondary collimator size may reduce the bremsstrahlung contamination (Kase and Bjarnard 1979). Further, bremsstrahlung from the Cerrobend cut-outs accounts for approximately 10 – 30% of the total contamination which decreases with increasing cut-out size (Zhu et al 2001). Preliminary analysis for cylindrical phantoms has been performed and comparison of two 6 MeV DEAR plans utilizing a 1x10 cm$^2$ and 3x10 cm$^2$ cut-out are shown in the Figure 26. While the 1x10 cm$^2$ cut-out produces a sharper penumbra at the boundary than the 3x10 cm$^2$ cut-out, as indicated by the results in Table 3, it also generates a higher bremsstrahlung contamination component than the 3x10 cm$^2$ cut-out for the same energy, target area, and target dose. For this example, the dose
calculated at the center of the phantom was approximately 5% of the target dose for the DEAR plan using a 1x10 cm$^2$ cut-out and 2% of the target dose for a DEAR plan using a 3x10 cm$^2$ cut-out.

The increase in the Bremsstrahlung component is due to a decrease in output for smaller cut-out sizes, which in turn requires a larger number of Monitor Units (MU) delivered for the same target dose and area. This suggests that a narrow cut-out may only be needed at the edge of the arc to sharpen the penumbra and should be avoided in the middle target region.

Figure 26. 2D dose distributions of equal beam weight 6 MeV DEAR plans using a (a) 1x10 cm$^2$ and (b) 3x10 cm$^2$ cut-out.
2.5. Conclusion

The concept of DEAR, a new modality for conformal electron beam therapy, was presented and investigated aspects for planning, delivery, and dose verification on cylindrical phantoms. The preliminary findings showed that DEAR can produce homogenous dose distributions over large and curved targets while maintaining narrow penumbra.
3. Development of accurate dose calculation algorithms for DEAR

3.1. Introduction

In this chapter, extensive validation of a Monte Carlo dose calculation framework for the TrueBeam linac is presented. First, the general framework is presented for all clinical field sizes and range of energies, then a small field study specifically for DEAR is presented. Finally, the development of a tool to improve Monte Carlo simulation efficiency is described.

3.1.1. Clinical treatment planning systems are insufficient for DEAR

DEAR planning cannot fully take place within the clinical treatment planning system. Modern treatment planning systems lack the treatment planning tools to properly plan dynamic electron therapies such as EAT, MERT, and DEAR and are not able to account for motions of multiple dynamic axes. Thus the previously detailed DEAR plan generation workflow (section 2.2.1.) was developed.

Further, dynamic radiation therapy techniques involving electron beams such as DEAR require accurate dose modeling of small field sizes, similar to the requirement of accurate modeling of small photon field for IMRT. Small field electron beam dosimetry differs considerably from standard field dosimetry. A previous study (Hu et al 2008) has shown inaccuracies in calculated dose distributions for small fields (<5x5 cm²) using the clinical treatment planning system dose calculation eMC algorithm. Thus, development
of accurate Monte Carlo simulations of small fields is necessary for dynamic electron beam delivery techniques.

Another motivation is the apparent observed discrepancy in the out-field dose in the orthogonal profile, as shown in Figure 27. The calculated distribution from the treatment planning system shows a “step” not visible in the measurement or Monte Carlo data. Since DEAR is comprised of the superposition of dose kernels, these inaccuracies are propagated into the in-field dose as well.

Figure 27. A step-and-shoot delivery of nine 16 MeV $1\times10$ cm$^2$ fields with no gap at the surface is calculated in the treatment planning system and measured with an ion chamber. (a) Calculated 2D dose distribution at depth of maximum dose. (b) The in-plane profile is plotted for the calculated (blue line) and measured (red scatter) dose distribution. A “step” is visible in the calculated dose distribution, but absent in the measured profile.
3.1.2. The Monte Carlo method

Modern treatment planning systems utilize either analytical or fast Monte Carlo radiation transport solutions. Pencil beam algorithms (PBA) based on the Fermi-Eyges scattering theory have been the foundation of electron beam dose calculation for decades (Hogstrom et al 1981). Subsequent advancements in analytical algorithms have improved the scattering theory; however, there are still limitations especially for small fields or tissue heterogeneities where the PBA generally fails. As such, calculation accuracies of ± 5% or ± 5 mm for electron fields in inhomogeneous materials are currently achievable. A more recent trend has been to use fast Monte Carlo algorithms based on pre-calculated kernels. The Varian treatment planning system Eclipse utilizes the Macro Monte Carlo implementation (Neuenschwander et al 1995) in its eMC algorithm.

The Monte Carlo (MC) method has been shown to be the most robust and accurate dose calculation tool for modeling electron beams in radiation therapy (Ma and Jiang 1999) and can provide an accurate representation of the radiation transport in the linac and patient if properly validated. Importantly, it can handle multiple electron scattering in the presence of heterogeneities (e.g. lung and rib cage) more accurately than analytical methods. Thus, the Monte Carlo method should be able to predict a dose distribution close to the measurement within a specified tolerance level that is clinically acceptable. A commonly accepted accuracy level is ± 2% or ± 2 mm as stated in the ICRU
Report 42 (Thwaites 1987). To achieve these accuracies, it is important that an accurate description of the linac head geometry, material composition, and source parameters are incorporated into the Monte Carlo simulations of electron beams for radiation therapy.

Contrary to previous linac models, Varian does not distribute geometric and material information upstream of the jaws for TrueBeam linacs for proprietary reasons (Parsons et al 2014). Instead International Atomic Energy Agency (IAEA) compliant phase space files for clinical electron energies are provided. These phase space files are generated using the Geant4 Monte Carlo code and computer aided design (CAD) software for the geometry (Constantin et al 2010, Constantin et al 2011). Further, the design of the TrueBeam linac differs from that of previous Varian linacs. TrueBeam linacs have different characteristics including reengineered electron modes with new scattering foil geometries (Glide-Hurst et al 2013). Therefore, Monte Carlo models developed for previous linac models such as Clinac or Trilogy may not be directly applicable to the TrueBeam linac and the dosimetric properties of photon and electron beams from TrueBeam linacs are not guaranteed to match those from Clinacs.
3.2. Validation of an electron Monte Carlo dose calculation framework using vendor provided phase space files on TrueBeam linacs

3.2.1. Introduction

Section 3 presented the feasibility of dynamic electron arc radiotherapy (DEAR), a novel conformal electron beam therapy technique, which utilizes the synchronous dynamic motion of multiple axes including gantry rotation and couch translation to produce highly conformal dose distributions, and is currently possible on Varian TrueBeam linacs in Developer Mode. As such, DEAR treatment planning must be able to handle dynamic motions of multiple axes in the dose calculation. Further, DEAR treatment planning requires accurate modeling of small field dose distributions as it utilizes the superposition of many small fields to generate desired dose distributions. However, algorithms capable of handling these requirements for DEAR are not yet available in current commercial treatment planning systems. Accurate small field fluence and dose distributions are also required of other advanced electron beam radiation therapy techniques such as modulated electron radiotherapy (MERT) that utilize energy and intensity modulation (Ma et al 2000, Hogstrom and Almond 2006, Klein et al 2009b).

Implementations of the Monte Carlo method for the calculation of dose distributions from standard electron beams are currently employed in modern clinical TPS. The electron Monte Carlo (eMC) algorithm available on the Eclipse TPS consists of
two models: the Initial Phase Space (IPS) model, (Janssen et al 2001) which describes the phase of particles emerging from the treatment head and the Macro Monte Carlo (MMC) method, (Neuenschwander et al 1995) which describes the transport of the electron in the patient anatomy. MMC calculates dose distributions from predetermined electron simulations generated various spherical volume elements (kugels) and a variety of materials. These transport probabilities stored in look-up tables. Since the transport through these kugels is predetermined, it is only necessary to sample from the distributions as the electrons exit the kugels – This leads to a large increase in simulation efficiency. While these implementations of Monte Carlo are available for clinical treatment planning, previous publications have shown that they may be inaccurate for small field dose calculations (Hu et al 2008, Xu et al 2009). Further, it is well known that the Monte Carlo method is advantageous in calculating accurate dose distributions at air-tissue interfaces when compared to pencil beam algorithms (Zhang et al 2013), and an accurate Monte Carlo model of the TrueBeam will thus be beneficial to all electron therapy techniques. Moreover, the current version of the Eclipse TPS does not allow planning of dynamic electron radiation therapy techniques including multiple mechanical axes motions such as gantry and couch. As such the TPS may not be adequate for planning of these advanced electron beam therapy techniques. Therefore, development of a treatment planning framework for DEAR, specifically, the dose
calculation algorithms capable of accurate small field dosimetry and dynamic motions, is warranted.

As stated in section 3.1.2., contrary to previous linac models, Varian does not distribute geometric and material information upstream of the jaws for TrueBeam linacs for proprietary reasons. Further, the design of the TrueBeam linac differs from previous models, and dosimetric equivalence cannot be guaranteed.

The purpose of this study was to develop and validate an accurate Monte Carlo dose calculation framework. As such, a comprehensive evaluation of Monte Carlo simulated dose distributions using vendor provided electron beam phase space files for TrueBeam linacs was performed and compared the simulated dose distributions with the measured dose distributions (Rodrigues et al 2015).

3.2.2. Materials and Methods

The following sections briefly describe vendor phase space generation and parameter tuning based on TrueBeam linac open field commissioning data. Then, the end-user Monte Carlo simulation framework developed by us under the EGSnrc environment is described, and validation of energy-specific phase space files for a subset of the energies 6, 9, 12, 16, and 20 MeV are presented, as these encompass the range of commissioned electron energies in many clinics. Both open (40×40 cm²) and collimated field with sizes ranging from 3×3, 4×4, 5×5, 6×6, 10×10, 15×15, 20×20, and 25×25 cm² were
assessed by Monte Carlo simulations and measurement data. Various metrics for assessing differences between the simulated and measured percent depth dose curves, diagonal and orthogonal profiles at depths $R_{100}$, $R_{50}$, $R_p$, and $R_{p+}$ and output factors are used.

### 3.2.2.1. Vendor phase space generation and parameter tuning

Phase space files were tuned by our collaborating author\(^1\) at Varian, which were validated using reference open field beam data. Vendor Monte Carlo simulations were performed using the GEANT4 code. Since this is beyond the scope of my work, for more details on the phase space generation and parameter tuning procedures, please see our publication (Rodrigues et al 2015).

### 3.2.2.2. End-user Monte Carlo simulations

Phase space files for the TrueBeam linac for a subset of the aforementioned electron energies (6, 9, 12, 16, and 20 MeV) were provided by the vendor in the IAEA compatible format just below the ion chamber and above the movable jaws, as depicted in Figure 28.

\[^{1}\text{Daren Sawkey, Research Scientist, Varian Medical Systems}\]
The EGSnrc code system (Kawrakow et al 2010) (v4 2.4.0) and user codes BEAMnrc (Rogers et al 1995) and DOSXYZnrc (Ma et al 1995) were used in this study for the Monte Carlo simulations downstream of the phase space plane. BEAMnrc was used to model the particle transport through the linear accelerator components based on the manufacturer’s specifications, while DOSXYZnrc was used to calculate dose deposition in a water phantom at a source to surface distance (SSD) of 100 cm. Simulations for
extended SSDs of 105, 110, 115, and 120 cm were also included to validate the model for common clinical treatment conditions.

Since a large amount of particles (1 billion particles correspond approximately to 20 GBs of storage for this study) were used in a single Monte Carlo simulation, multiple phase space files of smaller size (2 GBs) were provided and used. These individual files were then concatenated to one large phase space file. This energy-specific phase space file was then used for the downstream jaw and applicator simulation in BEAMnrc. After exiting the phase space plane, the particle passes through the secondary collimator modeled as the Y and X jaws and tertiary collimators modeled as the electron applicator. The scoring plane is located at the bottom of the electron applicator ($z = 5$ cm or 95 cm from the isocenter or target, respectively) and the resulting new phase space file is used as the input source for the subsequent phantom simulation in DOSXYZnrc.

Figure 29 depicts the downstream linac model schematically for field-specific simulations. The linac component modeling is simplified with the BEAMnrc-provided component modules (CM) (Rogers et al. 1995) for each specific part of the linac. For example, the Y and X jaws were modeled using the JAWS CM, while the applicator was modeled using the PYRAMIDS CM which can properly account for the beveled edges of the applicator opening.
Figure 29. Schematic representation of the TrueBeam linac model in BEAMnrc for field-specific simulations. Each component and its material in parentheses are depicted. While the applicator is made of a Zn alloy, the corresponding insert (gray) is made of Cerrobend. Geometry information was provided by the manufacturer. The scoring plane is located below the electron applicator at 95 cm.

The photon multi-leaf collimators (MLCs), which are located under the X jaws, were modeled for the open field simulation as a solid piece of tungsten with a 40×40 cm² opening using the PYRAMIDS CM. When a field-specific geometry was used, the MLCs are parked well behind the X jaws. The inclusion or omission of the MLCs, was tested
for all energies and applicator sizes, and differences were found to be within 1% and 1mm and thus were within the statistical uncertainty of the simulations. Resulting dose distributions for a 6 and 16 MeV electron beam and the largest field size 25x25 cm$^2$ are shown in Figures 30 and 31. Thus, in the field-specific simulations the modeling of the MLCs was omitted.

Figure 30. PDDs for a 6 and 16 MeV and 25x25 cm$^2$ field size electron beam Monte Carlo simulations modeled without (solid lines) and with (dashed lines) MLC. Differences were within 1%/1 mm.
The phase space file captures traversing particle history information including particle type \( p \), energy \( E \), position \( x, y, \) and \( z \), and direction cosines \( U, V, \) and \( W \) at a given plane. Three different particle types (photons, electrons, positrons) are recorded. The phase space is recorded at a plane with a constant \( z = 73.3 \) cm from the machine isocenter (or equivalently \( 26.7 \) cm from the target location) for a region of interest extending from \(-6.5 \) cm to \( 6.5 \) cm in the \( x \) and \( y \) direction. This scoring location is the same as the vendor-provided “version 2” photon phase space files, unlike the vendor provided photon phase space files presented by Constantin et al. (Constantin et al 2010), which were scored on a cylindrical surface. Only forward scattering particles, i.e., the
particles that are directed downstream, are recorded in the phase space file. \( W \) is thus not recorded in the phase space file.

In reality, the Y jaws move along a circular arc to keep the edge focused. Monte Carlo simulations with linear trajectory of the Y jaws and circular trajectory of the Y jaws was tested for all energies and field sizes. Differences were found to be within 1\% and 1 mm and thus were within the statistical uncertainty of the simulations. Resulting dose distributions for a 6 and 16 MeV electron beam and the largest field size 25x25 cm\(^2\) are shown in Figures 32 and 33. The largest field size is shown as the linear and circular trajectory lead to the largest discrepancy in the position of the Y jaws.

Figure 32. PDDs for a 6 and 16 MeV electron beam and 25x25 cm\(^2\) field size Monte Carlo simulations modeled with a Y jaw position along a linear (solid lines) and circular (dashed lines) path. Dose differences were within 1\%/1 mm.
Figure 33. Cross-plane profiles at three depths $R_{100}$, $R_{50}$, and $R_p$ (top to bottom) for a 6 and 16 MeV electron beam and 25x25 cm$^2$ field size Monte Carlo simulations modeled with a Y jaw position along a linear (solid lines) and circular (dashed lines) path. Dose differences were within 1%/1 mm. Profiles were normalized to the dose on the central axis and scaled to 100%, 50%, and 10%, respectively for the three depths.

Thus, when adjusting for a specific field size, the assumption of a linear trajectory for the Y jaws (moving in the X-Y plane for a constant Z) did not reduce the accuracy of the simulation, as long as the Y jaw face tilt remains focused on the source. The X jaws move along a linear trajectory at a constant z, such that the source-to-X-jaw distance increases as the field size increases. Further, the X jaws tilt such that the jaw faces remain focused on a 0.3×0.3 cm$^2$ square at the target location. Data for the material and geometry of the linac components was obtained from the TrueBeam Monte Carlo package version 1.1 available at MyVarian.com.
Dose per primary particle was scored in a 60×60×20 cm³ and 30×30×20 cm³ water phantom for open and collimated fields, respectively. For field sizes smaller than 6×6 cm², a voxel size of 0.25×0.25×0.2 cm³ in the X (cross-plane), Y (in-plane), and Z (depth) direction was used. For field sizes larger than 6×6 cm², the voxel size was increased to 0.5×0.5×0.2 cm³ to increase computational efficiency. Approximately 760 million particle histories from the phase space were required such that the statistical precision of the dose was less than 1% at the depth of maximum dose, d_max, for the chosen voxel size. No particle recycling was used. The statistical precision of the Monte Carlo simulation gives the lower limit of the accuracy of the calculated dose (Faddegon et al 2008). Most particle transport parameters were chosen to be the same as in previous electron beam Monte Carlo studies (Faddegon et al 2009). As per Kawrakow et al. (Kawrakow et al 2010), the default particle transport parameters for an electron beam within EGSnrc were used. The energy thresholds for generation of secondary electrons or Bremsstrahlung photons AE and AP were also set to 0.521 and 0.01 MeV, respectively. Thus the electron lower energy cutoff ECUT and photon lower energy cutoff PCUT were set to 0.521 and 0.01 MeV, respectively.

All other options were turned off. Electron range rejection was employed to speed up calculation time during the linac components simulation in BEAMnrc. The electron range rejection variance reduction technique was employed (option 1: “on with
varying ECUTTR”) with an ESAGE GLOBAL of 1 MeV (O’Shea et al 2008). This means that during the simulation, any electron with an energy below 1 MeV is evaluated to determine if their range is large enough to make it out of treatment head with an energy greater or equal to ECUT and thus contribute to the energy deposition in the phantom simulation. To validate the use of this variance reduction technique, simulations were performed both with and without electron range rejection for a high energy beam (16 MeV) and the smallest field size (6×6 cm²). The resulting PDDs showed no significant differences (i.e. less than the statistical precision of the dose of 0.7%), with dose differences of < 0.5% and range differences of < 0.1 mm.

3.2.2.3. End-user measurement data

Measurement data were acquired in a large 48×48×41 cm³ water phantom with a 3D scanning system (Blue Phantom², IBA Dosimetry, Schwarzenbruck, Germany). An electron field diode detector (EFD³G, IBA Dosimetry, Schwarzenbruck, Germany) placed vertically in the water tank was used. This diode has an active area of 2 mm in diameter and 0.06 mm thickness. Percent depth doses were measured for field sizes ranging from 6×6 to 25×25 cm², as well as orthogonal and diagonal profiles at multiple depths R₁₀₀, R₅₀, R_p, and R_p+ for electron energies of 6, 9, 12, 16, and 20 MeV for an SSD of 100 cm. Further, cone (6×6 to 25×25 cm²) and cut-out (3×3 to 5×5 cm² cut-out for a 6×6 cm² applicator) output factors were previously collected during commissioning.
Table 5 lists the depths defined for a 15×15 cm² electron applicator as a function of energy based on commissioning data. For the extended SSDs of 110 and 120 cm, PDDs and profiles were acquired for a 15×15 cm² field size at depths $R_{100}$ and $R_{50}$ for the same electron energies. Table 6 lists the standard jaw settings for each electron applicator and electron energy combination for the TrueBeam linac. It is worth noting that these differ slightly from Clinac or Trilogy linacs.
Table 5. Ranges (cm) as a function of electron beam energy (MeV). Ranges $R_{100}$, $R_{50}$, $R_p$, and $R_{p+}$ were defined for a 15×15 cm$^2$ electron applicator for each energy. $R_{100}$ is the depth of maximum dose, $R_{50}$ is the depth at the descending 50%. $R_p$ is the practical range. $R_{p+}$ was defined as the depth at $R_p + 2$ cm.

<table>
<thead>
<tr>
<th>Range (cm)</th>
<th>Energy (MeV)</th>
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<tr>
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<td>$R_p$</td>
<td>3.1</td>
</tr>
<tr>
<td>$R_{p+}$</td>
<td>5.1</td>
</tr>
</tbody>
</table>

Table 6. Jaw settings for electron applicator and energy combinations for TrueBeam and Clinac linacs. Since jaw settings for square field are symmetric for both the X and Y jaws, only the absolute value in cm is listed. If the jaw settings are the same for both TrueBeam and Clinac, only one value is shown; if they are different, the value for Clinac is shown in parentheses.

<table>
<thead>
<tr>
<th>Energy (MeV)</th>
<th>Applicator (cm$^2$)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
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</tr>
<tr>
<td>6</td>
<td>20</td>
</tr>
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<td>20</td>
</tr>
<tr>
<td>12</td>
<td>11</td>
</tr>
<tr>
<td>16</td>
<td>11</td>
</tr>
<tr>
<td>20</td>
<td>11</td>
</tr>
</tbody>
</table>

Measured PDD curves were compared to commissioning measurements acquired with a CC13 cylindrical ionization chamber to assess the precision of the diode
measurements and the reproducibility of the measurement data. Differences in $R_{50}$ were found to be well within 1 mm. Comparison of the PDDs matched at $R_{50}$ with a gamma index criteria of 2%/1 mm yielded pass rates of 100% for all energies. The accuracy of the diode measurements can thus be considered to be within 2%/1 mm.

3.2.2.4. Comparison metrics

3.2.2.4.1. Percent depth dose and profiles

Post processing for Monte Carlo simulated dose distributions included smoothing by averaging with neighboring voxels to improve the precision without simulating more histories. (Faddegon et al 2005) Post processing for measured dose distributions included smoothing PDDs with a least-square smoothing filter to preserved peak information, and smoothing profiles with a median smoothing filter to preserve field edges, as well as symmetrizing the profiles. PDDs and profiles were normalized such that a relative comparison between Monte Carlo simulation, measurements, and calculations could be done. PDDs were normalized to the maximum dose and profiles were normalized to the value on the central axis. For the PDDs, dose difference, distance-to-agreement, and gamma index were calculated for the build-up region (surface $\leq$ depths $\leq$ distal $R_{100}$), fall-off region (distal $R_{100} <$ depths $\leq R_p$), and Bremsstrahlung region (depths $> R_p$). Since dose difference and distance-to-agreement metrics are inherently sensitive to different aspects of the PDD curve, the gamma
index (Low et al. 1998) is used to evaluate the agreement criteria of 2%/1 mm simultaneously. For each region, an energy-specific pass rates encompassing all applicator combinations was calculated as the ratio of the number of voxels with gamma index ≤ 1 to the total number of voxels for that energy in that region.

For the orthogonal profiles, gamma analysis was also performed and energy-specific pass rates were calculated for depths $R_{100}$, $R_{50}$, and $R_{10}$. Further, field size, penumbra, and flatness were evaluated at the aforementioned depths. The field size was defined as the lateral distance between the 50% dose levels relative to the central axis dose at a given depth, while the penumbra was defined as the lateral distance between the relative 20% and 80% dose levels for a given profile. The flatness (Equation 3) was defined as the maximum variation observed within 80% of the field size, where $M$ and $m$ are the maximum and minimum dose values.

$$Flatness = \frac{M - m}{M + m} \cdot 100\%$$

Equation 3

3.2.2.4.2. Cone and cut-out output factors

Output factors (OF) were calculated from the Monte Carlo simulation and compared to the commissioning data. Equation 4 defines the cone output factor for a specific energy $E$ (and therefore depth) for an arbitrary field size $FS$ as the ratio of the
dose for that field size to the dose for a reference field size of 15×15 cm² at the d_{max}
delivered at an SSD of 100 cm and the same number of MU.

\[
OF(E, FS) = \frac{Dose(E, FS)}{Dose(E, 15 \times 15 cm^2)} \quad \text{Equation 4}
\]

The depths of comparison were 1.5, 2.2, 2.7, 3.2, and 2.8 cm for 6, 9, 12, 16, and 20
MeV, respectively. Further, the cut-out output factors for square Cerrobend cut-outs of
3×3, 4×4, and 5×5 cm² in a 6×6 cm² applicator were also calculated and compared to
commissioning data.

To ascertain that the Monte Carlo simulation can accurately model output at
extended SSDs, output at SSDs of 105, 110, 115, and 120 cm (with corresponding gap of
5, 10, 15, and 20 cm) were acquired for both the measurement and simulations. The
output factors were calculated by normalizing the output at a given energy, field size,
and SSD to the output for that energy and field size at a SSD of 100 cm.

3.2.3. Results

3.2.3.1. End-user open field simulations

End-user simulations for open field geometries for energies of 6, 9, 12, 16, and 20
MeV showed similar levels of agreement to the commissioning data as the vendor
simulations. Simulated and measured results for the PDDs and cross-plane profiles are
shown in Figure 34. Excluding the surface voxel, the agreement in the PDDs is within
2% for 12, 16, and 20 MeV and generally within 2% for the 6 and 9 MeV PDDs. While the lower energies fail the 2% criteria at 2.3 cm and 3.5 cm, respectively, they are still within the 1 mm criteria, thus still fulfilling the gamma criteria. Similar to the vendor simulations, the cross-plane profiles agreed well within 2% for the higher energies, but failed for the 6 MeV by almost 3% towards the field edges. In the end-user simulation, the 9 MeV also failed toward the field edges by 2.4%, while the agreement for the higher energies 12, 16, and 20 MeV were all within 2%.
Figure 34. Simulated vendor (stars) and end-user (circles) and measured (lines) (a) PDDs and (b) in-air cross-plane profiles for the subset of energies 6, 9, 12, 16, and 20 MeV. Data were normalized to the same $d_{\text{max}}$ for the PDD and the same central axis value for the profiles the same energy, but arbitrary for different energies for clearer presentation (normalization increases with increasing energy).
3.2.3.2. End-user collimated field simulations

3.2.3.2.1. Percent depth dose curves

Measured (diode) and simulated (Monte Carlo) PDD curves for all combinations of electron energies 6, 9, 12, 16, and 20 MeV and applicators 6×6, 10×10, 15×15, 20×20, and 25×25 cm² at SSD of 100 cm were compared. Select PDDs for all energies and applicator 6×6, 15×15, and 25×25 cm² are shown in Figure 35. Dose differences between measurement and simulation were mostly within 2%. Minimum, maximum, and the average of the absolute dose differences for the build-up (defined from the surface to \( R_{100} \)), fall-off (from \( R_{100} \) to \( R_{p} \)), and Bremsstrahlung (> \( R_{p} \)) region are shown in Table 7. Dose differences in the build-up region were within 2% for depths larger than 0.5 cm, except in the first voxel adjacent to the surface which displayed dose differences of up to 4.5%. In the fall-off region, defined from the distal \( R_{100} \) to \( R_{p} \), disagreement was mostly within 2%. The agreement in the Bremsstrahlung region, which contributes to energy deposition past the practical range \( R_{p} \), was within 2% for all cases with most differences within 1%. PDDs for the higher energies of 16 and 20 MeV and all applicators exhibited differences within 2% for the entire PDD, excluding the first 0.5 cm.

The distance-to-agreement in the fall-off region from the distal \( R_{100} \) to \( R_{p} \) was within 1 mm for all cases. The gamma index with a passing criteria for the dose
difference of 2% and distance-to-agreement of 1 mm was calculated for all regions. The pass rate (ie. the percentage of voxels displaying a gamma index of less than or equal to 1) in the build-up region was 94%, 88%, 81%, 91%, and 90% for energies 6, 9, 12, 16, and 20 MeV, respectively, while the fall-off and Bremsstrahlung region displayed pass rates of 100% for all energies. Excluding the shallowest 0.5 cm would eliminate almost all the failing voxels. Alternatively, relaxing the distance criteria to 2%/2 mm for the build-up region leads to a pass rate of 100% for all energies and field sizes.
Figure 35. Select PDD curves for (a) 6×6, (b) 15×15, and (c) 25×25 cm² applicators and 6, 9, 12, 16, and 20 MeV electron beams (right to left) shown up to depths $R_{p+}$ for the respective energies. Measurements (lines) and Monte Carlo simulations (points) are compared. The curves are normalized to 100%, 110%, 120%, 130%, and 140% at $d_{\text{max}}$ for each energy, respectively.
Table 7. Minimum and maximum dose differences and average absolute dose differences (MC – measurement) for the build-up (< R_{100}), fall-off (R_{100} to R_p), and Bremsstrahlung (>R_p) region for all energy and applicator combinations. The dose values were normalized to 100% at d_{max}. Values that are larger than ± 2% are shown in bold.

<table>
<thead>
<tr>
<th>Applicator (cm^2)</th>
<th>6×6</th>
<th>10×10</th>
<th>15×15</th>
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<tbody>
<tr>
<td>E</td>
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<td>max</td>
<td>avg</td>
<td>min</td>
<td>max</td>
</tr>
<tr>
<td>6</td>
<td>-3.8</td>
<td>-1.9</td>
<td>-0.4</td>
<td>-4.2</td>
<td>-1.1</td>
</tr>
<tr>
<td>9</td>
<td>0.0</td>
<td>2.2</td>
<td>2.0</td>
<td>0.0</td>
<td>2.8</td>
</tr>
<tr>
<td>12</td>
<td>1.3</td>
<td>1.3</td>
<td>0.3</td>
<td>1.7</td>
<td>1.3</td>
</tr>
<tr>
<td>16</td>
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<td>-1.9</td>
<td>-0.4</td>
<td>-4.1</td>
<td>-0.9</td>
</tr>
<tr>
<td>20</td>
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<td>1.4</td>
<td>0.1</td>
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<tr>
<td>avg</td>
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<td>1.2</td>
<td>0.3</td>
<td>1.4</td>
<td>0.7</td>
</tr>
</tbody>
</table>

3.2.3.2.2. Orthogonal and diagonal profiles

Measured and simulated cross-plane profiles for all energies and field sizes at depths R_{100}, R_{50}, R_p, and R_{p^+} were compared. Select measured and simulated cross-plane profiles at depths of R_{100}, R_{50}, R_p for the 6 MeV and 20 MeV electron beams for field size
6×6, 15×15, and 25×25 cm$^2$ are shown in Figure 36. Further, cross- and in-plane profiles at R$_{p}$ for the same energies and largest field size 25×25 cm$^2$ are shown in Figure 37. Dose differences at R$_{100}$ were mostly within 2% with discrepancies of up to 3% occurring at the penumbra region. For a gamma index criteria of 2%/1 mm, the average pass rates for the orthogonal profiles were ≥ 99% at R$_{100}$ and 100% at R$_{50}$, R$_{p}$, and R$_{p}$ for all energies. To compare the range of discrepancies, field size, penumbra, and flatness differences for all energies at R$_{100}$ and R$_{p}$ are shown in Table 8. Orthogonal (in-plane and cross-plane) profiles agreed well at R$_{100}$ with differences in field size within 1 mm, penumbra within 1 mm, and flatness within 2%. For depths R$_{50}$ and R$_{p}$, field size and penumbra mostly agreed to within 1 mm as well. Agreement in flatness at depth R$_{50}$ and R$_{p}$ was mostly within 2%, with a maximum difference of 3.8% observed for a 6 MeV beam and a 15×15 cm$^2$ applicator. Additionally, selected measured and simulated diagonal profiles at R$_{100}$ for 6×6, 15×15, and 25×25 cm$^2$ field sizes and 9, 12, and 16 MeV are shown in Figure 38. While for the 6 and 20 MeV beam the agreement in the shoulder region is good, for 9, 12, and 16 MeV the simulation results in a higher dose in the shoulder of the profile than the measurement.

Measured and simulated cross-plane profiles at depths R$_{100}$ and R$_{50}$ for extended SSDs of 110 and 120 cm for a 15×15 cm$^2$ field size for all energies agreed within 2%/1 mm and field size, penumbra, and flatness were within 1 mm, 1 mm, and 2%, respectively.
Representative cross-plane profiles at SSDs of 100, 110, and 120 cm for the 6 MeV electron beam and a 15×15 cm² field size at depth $R_{100}$ are shown in Figure 39.

Figure 36. Select cross-plane profiles for measurements (lines) and Monte Carlo simulations (points) for 6 and 20 MeV electron beams and field sizes of 6×6, 15×15, and 25×25 cm². Each subplot contains measured and simulated profiles at $R_{100}$, $R_{50}$, and $R_v$ (top to bottom) which are arbitrarily normalized to 100%, 50%, and 10% at the central axis for easy viewing, respectively.
Figure 37. Cross-plane (measurement: blue solid line, MC: circles) and in-plane (measurement: red dashed line, MC: triangles) profiles at $R_p^+$ for the (a) 6 and (b) 20 MeV electron beam for the largest applicator of $25\times25$ cm$^2$. The measured profiles were consistently higher by 0.5% than the simulated profiles.
Table 8. Average absolute (in-plane and cross-plane) difference in field size (FS) (mm), penumbra (mm), and flatness (%) shown for each energy (6, 9, 12, 16, and 20 MeV) and applicator (25×25, 20×20, 15×15, 10×10, and 6×6 cm²) combination. Each quantity was calculated at depths R₁₀₀, and Rₚ, corresponding to the values listed in each cell of the table.

<table>
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<th>Energy (MeV)</th>
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<th>15×15</th>
<th>10×10</th>
<th>6×6</th>
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</thead>
<tbody>
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<td>0.3, 0.6</td>
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<tr>
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<td>0.5, 1.4</td>
</tr>
<tr>
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<td>Δflatness (%)</td>
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<td>1.8, 2.7</td>
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<td>0.2, 0.4</td>
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</tr>
<tr>
<td>9</td>
<td>Δpenumbra (mm)</td>
<td>1.3, 0.5</td>
<td>0.6, 0.6</td>
<td>0.8, 0.4</td>
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<td>0.2, 1.2</td>
<td>0.6, 1.4</td>
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<td>0.3, 1.6</td>
<td>0.6, 0.7</td>
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<td>0.8, 0.4</td>
<td>0.7, 0.6</td>
<td>1.1, 1.1</td>
<td>1.2, 0.4</td>
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<tr>
<td></td>
<td>Δflatness (%)</td>
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<td>0.8, 1.5</td>
<td>0.6, 3.2</td>
<td>1.7, 1.6</td>
<td>1.9, 1.0</td>
</tr>
<tr>
<td></td>
<td>ΔFS (mm)</td>
<td>0.7, 1.3</td>
<td>0.5, 1.4</td>
<td>0.3, 1.1</td>
<td>0.1, 0.1</td>
<td>0.4, 0.4</td>
</tr>
<tr>
<td>20</td>
<td>Δpenumbra (mm)</td>
<td>0.0, 1.3</td>
<td>0.4, 0.9</td>
<td>0.4, 0.9</td>
<td>0.7, 2.0</td>
<td>0.5, 2.4</td>
</tr>
<tr>
<td></td>
<td>Δflatness (%)</td>
<td>1.2, 2.8</td>
<td>1.3, 0.6</td>
<td>1.3, 1.3</td>
<td>1.1, 3.4</td>
<td>0.4, 2.5</td>
</tr>
</tbody>
</table>
Figure 38. Select diagonal profiles at R₁₀₀ for 6×6, 15×15, and 25×25 cm² applicators (left to right) and (a) 9, (b) 12, and (c) 16 MeV electron beams, shown due to the increasing discrepancy in the shoulder region with energy. Measurements (lines) and Monte Carlo simulations (points) are compared. Due to symmetry of the orthogonal profiles, only the positive off-axis positions are shown.

Figure 39. Measured (lines) and Monte Carlo simulations (points) for 15×15 cm² field size for 6 MeV beam at SSDs of 100, 110, and 120 cm. Due to symmetry of the orthogonal profiles, only the positive off-axis positions of the cross-plane profiles at R₁₀₀ are shown.
3.2.3.2.3. Output factors

Cone (6×6 to 25×25 cm²) and cut-out (3×3, 4×4, and 5×5 cm²) output factors from measurement and Monte Carlo simulation are shown in Table 9. Output factors derived from the Monte Carlo simulation agreed well with the commissioning data, with most differences within 2% and the worst at 2.5% observed for 6 MeV and a 25×25 cm² field size.
Table 9. Average commissioning (Measurement) and Monte Carlo cone and cut-out output factors for all energy and applicator combinations. Cut-outs (3×3, 4×4, 5×5 cm²) for a 6×6 cm² applicator are shown. Differences larger than 2% are highlighted in bold.

<table>
<thead>
<tr>
<th>Energy (MeV)</th>
<th>Cone &amp; cut-out size (cm²)</th>
<th>Measurement</th>
<th>MC</th>
<th>Difference (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>3×3</td>
<td>0.867</td>
<td>0.879 ± 0.009</td>
<td>1.2</td>
</tr>
<tr>
<td></td>
<td>4×4</td>
<td>0.936</td>
<td>0.953 ± 0.010</td>
<td>1.7</td>
</tr>
<tr>
<td></td>
<td>5×5</td>
<td>0.949</td>
<td>0.966 ± 0.010</td>
<td>1.7</td>
</tr>
<tr>
<td></td>
<td>6×6</td>
<td>0.945</td>
<td>0.957 ± 0.008</td>
<td>1.2</td>
</tr>
<tr>
<td>9</td>
<td>6×6</td>
<td>0.988</td>
<td>0.990 ± 0.008</td>
<td>0.2</td>
</tr>
<tr>
<td></td>
<td>10×10</td>
<td>1.000</td>
<td>0.998 ± 0.009</td>
<td>-0.2</td>
</tr>
<tr>
<td></td>
<td>15×15</td>
<td>1.000</td>
<td>1.000 ± 0.009</td>
<td>0.0</td>
</tr>
<tr>
<td></td>
<td>20×20</td>
<td>1.045</td>
<td>1.028 ± 0.009</td>
<td>-1.7</td>
</tr>
<tr>
<td></td>
<td>25×25</td>
<td>1.061</td>
<td>1.036 ± 0.009</td>
<td>-2.5</td>
</tr>
<tr>
<td>12</td>
<td>3×3</td>
<td>0.860</td>
<td>0.859 ± 0.008</td>
<td>-0.1</td>
</tr>
<tr>
<td></td>
<td>4×4</td>
<td>0.922</td>
<td>0.961 ± 0.008</td>
<td>0.3</td>
</tr>
<tr>
<td></td>
<td>5×5</td>
<td>0.947</td>
<td>0.950 ± 0.008</td>
<td>0.3</td>
</tr>
<tr>
<td></td>
<td>6×6</td>
<td>0.952</td>
<td>0.957 ± 0.007</td>
<td>0.5</td>
</tr>
<tr>
<td>Size</td>
<td>Value 1</td>
<td>Value 2 ± Error</td>
<td>Error 1</td>
<td></td>
</tr>
<tr>
<td>--------</td>
<td>---------</td>
<td>-----------------</td>
<td>---------</td>
<td></td>
</tr>
<tr>
<td>10x10</td>
<td>0.996</td>
<td>0.997 ± 0.007</td>
<td>0.1</td>
<td></td>
</tr>
<tr>
<td>15x15</td>
<td>1.000</td>
<td>1.000 ± 0.007</td>
<td>0.0</td>
<td></td>
</tr>
<tr>
<td>20x20</td>
<td>1.000</td>
<td>0.995 ± 0.007</td>
<td>-0.5</td>
<td></td>
</tr>
<tr>
<td>25x25</td>
<td>0.986</td>
<td>0.979 ± 0.007</td>
<td>-0.7</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Size</th>
<th>Value 1</th>
<th>Value 2 ± Error</th>
<th>Error 1</th>
</tr>
</thead>
<tbody>
<tr>
<td>3x3</td>
<td>0.912</td>
<td>0.909 ± 0.008</td>
<td>-0.3</td>
</tr>
<tr>
<td>4x4</td>
<td>0.960</td>
<td>0.961 ± 0.008</td>
<td>0.1</td>
</tr>
<tr>
<td>5x5</td>
<td>0.980</td>
<td>0.981 ± 0.008</td>
<td>0.1</td>
</tr>
<tr>
<td>6x6</td>
<td>0.982</td>
<td>0.991 ± 0.007</td>
<td>0.9</td>
</tr>
<tr>
<td>10x10</td>
<td>1.008</td>
<td>1.008 ± 0.007</td>
<td>0.0</td>
</tr>
<tr>
<td>15x15</td>
<td>1.000</td>
<td>1.000 ± 0.007</td>
<td>0.0</td>
</tr>
<tr>
<td>20x20</td>
<td>0.995</td>
<td>0.990 ± 0.007</td>
<td>-0.5</td>
</tr>
<tr>
<td>25x25</td>
<td>0.980</td>
<td>0.976 ± 0.007</td>
<td>-0.4</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Size</th>
<th>Value 1</th>
<th>Value 2 ± Error</th>
<th>Error 1</th>
</tr>
</thead>
<tbody>
<tr>
<td>3x3</td>
<td>1.003</td>
<td>1.003 ± 0.010</td>
<td>0.0</td>
</tr>
<tr>
<td>4x4</td>
<td>1.017</td>
<td>1.018 ± 0.010</td>
<td>0.1</td>
</tr>
<tr>
<td>5x5</td>
<td>1.022</td>
<td>1.020 ± 0.010</td>
<td>-0.2</td>
</tr>
<tr>
<td>6x6</td>
<td>1.022</td>
<td>1.023 ± 0.008</td>
<td>0.2</td>
</tr>
<tr>
<td>10x10</td>
<td>1.021</td>
<td>1.025 ± 0.008</td>
<td>0.4</td>
</tr>
<tr>
<td>15x15</td>
<td>1.000</td>
<td>1.000 ± 0.007</td>
<td>0.0</td>
</tr>
<tr>
<td>20x20</td>
<td>0.991</td>
<td>0.993 ± 0.007</td>
<td>0.2</td>
</tr>
<tr>
<td>25x25</td>
<td>0.974</td>
<td>0.976 ± 0.007</td>
<td>0.2</td>
</tr>
</tbody>
</table>
A common clinical scenario is to treat an extended SSD, thus increasing the air gap. The output at extended SSDs of 105, 110, 115, and 120 cm were evaluated for measured (average from commissioning data from three linacs) and Monte Carlo simulations for all energies 6, 9, 12, 16, and 20 MeV and field sizes of 6×6, 10×10, 15×15, 20×20, and 25×25 cm². The inter-linac variations of the output at extended SSDs were within 1.3%. Normalizing the output at a given SSD to the output at a SSD of 100 cm, measured and simulated output for a given energy and field size as a function of extended SSDs showed differences mostly within 1% for all energies and field sizes, except for the output of a 6 MeV and a 6×6 cm² field size, which showed differences of up to 2.5% for an extended SSD of 120 cm (Figure 40). While the higher energies generally did not show a significant trend toward higher or lower ratios (i.e. the ratio fluctuated about 1) as a function of SSD, the ratio for the 6 MeV and 6×6 cm² exhibited a trend towards higher ratios as the SSD increased.
Figure 40. Ratio between measured and simulated output factors for the 6 MeV electron energy for field sizes 6×6, 10×10, 15×15, 20×20, and 25×25 cm² as a function of SSD. While most differences are within 1%, the ratio for the 6x6 cm² at the largest SSD of 120 cm displays differences of up to 2.5%.

3.2.4. Discussion

3.2.4.1. End-user validation

Compared to previous validation studies on older generation linacs, this study achieved a similar level of disagreement of 2%/1mm. Antolak et al. (Antolak et al 2002) had previously achieved Monte Carlo models for a Varian Clinac 2100C that were accurate to within 2%/1 mm for almost the entire range of clinical data. Generally, this
agreement is difficult to achieve if accurate source parameters such as focal spot size, source angle, angular divergence, and energy distribution are not used (Huang et al 2005). Previous sensitivity studies have noted that due to the number of parameters, the selection of appropriate parameters may vary and multiple sets of values may give a good fit (Faddegon et al 2005, Faddegon et al 2008). While Varian does provide information about the entire treatment head for other linac models, it was left up to the user to determine the source parameters through iterative beam tuning by comparing measured and simulated dose distributions. This could lead to parameter selections that did not work over the entire clinical range. The framework described in this paper relies on the parameter selection based on the sensitivity study performed for open-field dose distributions, which are independent of the downstream jaw and applicator settings. Using the phase space file approach may thus result in better agreement over the entire clinical range.

3.2.4.2. Open and collimated field simulations

Comprehensive investigation of Monte Carlo simulated dose distributions from electron beam phase space files for TrueBeam linacs was presented. The measurement uncertainty was found to be less than 1% for repeat measurements. Generally, percent depth dose curves obtained with diodes and ion chambers match well (Ten Haken et al 1987). However, diodes have been shown to over respond to x-rays, (Faddegon et al
which may contribute to the difference between simulation and measurement in the Bremsstrahlung region. Systematic measurement set-up errors in depth were reduced by checking tank alignment and diode placement after every scan to compensate for water evaporation. Further, PDDs were compared to commissioning data (acquired with a CC13 ion chamber) and were found to be in agreement for a gamma criteria of 2%/1 mm. The diode was found to over-respond in the Bremsstrahlung region by about 0.6% for all energies and field sizes.

The simulations were run until an uncertainty of less than 1% was achieved at the depth of maximum dose. The simulated uncertainty remains at about 1% until $R_p$. At depths beyond the practical range, the simulated uncertainty rises to over 5%.

The Gamma index pass rates for the PDD indicated that agreement was within 2%/1 mm for the fall-off and the Bremsstrahlung region, while the voxels at depths <5 mm in the build-up region failed the 2%/1 mm criteria. A possible explanation for the discrepancy in the first 5 mm of the build-up region could be due to a missing low energy scatter component. Similar results had been previously reported by Faddegon et al. (Faddegon et al 2009). In the current framework, the photon Multi-leaf Collimator (MLC) component module was not included in the Monte Carlo simulation of the collimated fields as it is completely retracted during electron beam therapy. Including it increases the open field build-up region PDD by 0.5% on average, which slightly
improves the agreement with the measurement, however, does not change the collimated field distributions significantly. Further, the disagreement at the surface between measurement and simulation, both by the vendor and end-user, can be seen both in the open and collimated field simulation.

Orthogonal profiles agreement is almost always within the 2%/1 mm criteria for all depths. Field size, penumbra, and flatness for orthogonal profiles at depths $R_{100}$, $R_{50}$, and $R_p$ agreed well between measurements and simulations. Agreement in the shoulder region at different depths, suggests the correct modeling of beam-line components and source. Diagonal profiles seem to show a trend towards an increase in the simulated versus the measured dose in the shoulder region for energies 9, 12, and 16 MeV of up to 2.5%, however, 6 and 20 MeV do not show this trend. Orthogonal profiles for extended SSDs of 110 and 120 cm for all energies and a 15×15 cm$^2$ also agreed well with measurement data within 2%/1 mm. Cone and cut-out factors for a range of clinically relevant field sizes agreed within 1% for all applicators and energies above 6 MeV. For the lowest energy of 6 MeV, the output factors agree within 2.5%, with the largest discrepancy observed for the 25×25 cm$^2$ applicator. Output factors at extended SSDs mostly agreed to within 1%, with lower energies of 6 MeV showing deviations of up to 2.5% at SSDs of 120 cm. Generally, the 6 MeV simulations tended to show the worst
agreement in both percent depth dose curves and output factors than the higher energies.

The output factors at a given energy and SSD shows a trend as a function of field size as shown in Figure 41 for the measured and simulated of a 6 MeV beam at an SSD of 100 cm. This trend is most prominent for the lower energies and mostly falls within the uncertainty of the simulations for the 16 and 20 MeV output factors.

![Figure 41. Measured (dashed) and simulated (solid) output factor for a 6 MeV beam at an SSD of 100 cm as a function of field size. For small field sizes the simulation underestimates the output factor, while for larger field sizes, the simulation overestimates the output factor.](image-url)
One explanation for this discrepancy is that the simulation does not take into account the backscattered fraction, which contributes dose to the monitor chamber. As the jaw size decreases, one expects the ratio of backscattered dose to the monitor chamber to increase, thus, the delivered MU would be reached more quickly in the monitor chamber, and less dose would be measured by an ion chamber located at isocenter. Neglecting the backscatter fraction would lead to a larger simulated output factor for smaller field sizes than measured and vice versa. A previous study by Verhaegen et al. (Verhaegen et al 2000) on a Varian Clinac 2100C linac had noted that the measured and simulated backscatter contribution to the monitor chamber for standard field sizes increased by 2-3% for a 6 MeV when field sizes were decreased from a 40×40 to a 0.5×0.5 cm² field size. This study also found that the backscatter contribution was less important for higher energy electron beams. This trend was also observed in our data. Because the TrueBeam Monte Carlo package does not provide information about the monitor chamber, backscattered dose to the monitor chamber will thus not be accounted for and the absolute dose calibration formalism as provided by Popescu et al. (Popescu et al 2005) is not possible. A solution is to use the vendor provided Monte Carlo simulation (VirtuaLinac), which can be used to calculate monitor chamber does as a function of jaw position.
3.2.4.3. Comparison to eMC

As a comparison to the clinically used eMC algorithm v11, calculations for combinations of electron beam energies (6, 9, 12, 16, and 20 MeV) and field size (6x6, 10x10, 15x15, 20x20, and 25x25 cm²) were performed. As per previous studies (Popple et al 2006a, Hu et al 2008) the recommended calculation grid size (1 and 1.5 mm for 6 and 9 MeV and 2 mm for 12-20 MeV), uncertainty (< 1%), and smoothing (Medium level 3D Gaussian smoothing) were used. Comparison of calculated dose distributions using the eMC algorithm generally agreed well with diode measurements for standard field sizes. Percent depth dose curves and orthogonal profiles agreement was generally within the 2%/1 mm criteria. It is worth pointing out that good agreement between measured and eMC calculation derived output factors was expected since cone-specific output factors were required for the eMC during the algorithm configuration.

3.2.4.4. Future Work

The resulting phase space files created from realistic linac head geometry and electron beam parameters tuned from open field measurements can be used in the simulation of dose distributions of TrueBeam linacs for multiple energy and field size combinations, as have been demonstrated in this study. Electron beam Monte Carlo simulation of TrueBeam linacs requires the use of phase space files. While the use of phase space files produced accurate dose distributions, there are also limitations posed
to the end user. Phase space files can be large and thus rather cumbersome to distribute and work with. Further, since the phase space file is scored just prior to the jaws, variations of an individual linac from the standard configuration cannot be incorporated by the end user. One approach to overcome these limitations is to parameterize the phase space file. This reduces the computer storage requirement and allows users to generate as many histories as necessary for higher statistical precision. Further, parameterization may also provide the users with the opportunity to tune the beam model to better match a specific linac while still allowing simulations to be run in the presented framework. An alternative approach could be to use the VirtuaLinac, which allows users to run Monte Carlo simulations remotely and gives users the ability to change parameters of the model such as incident electron source parameters (Parsons et al 2014).

The presented Monte Carlo framework provides a basis for simulating advanced dynamic electron beam therapy techniques such as DEAR, which currently cannot be fully planned with the commercial clinical treatment planning system. The dose deposition in this validation study was done using a simple cuboid water phantom. For DEAR applications, more complex geometries and materials are likely needed. Since this framework was implemented in the EGSnrc BEAMnrc/DOSXYZnrc Monte Carlo environment, DOSXYZnrc will be used to model complex voxelized phantoms. Since
DEAR is a dynamic technique, full Monte Carlo simulations would require a way to parameterize the delivery as a function of time. This could be implemented in DOSXYZnrc using sources that allow for continuously variable beam configurations such as the sources developed by Lobo et al. (Lobo and Popescu 2010).

3.2.5. Conclusion

A Monte Carlo simulation framework for electron beam dose calculations for Varian TrueBeam linacs was presented. Instead of relying on knowledge of accurate geometric information and material composition of the linac head components, users can use the phase space files provided by the vendor as the source input to the simulation in the EGSnrc Monte Carlo environment. Electron beam energies from 6 to 20 MeV for open and collimated field sizes from 3×3 to 25×25 cm² were studied and results were compared to the measurement and commissioning data with excellent agreement in central axis depth dose, both orthogonal and diagonal profiles at different depths, cone and cut-out output factors, and for extended SSD dosimetry. This validates that the phase space files provided by the vendor can be used in the accurate simulation of dose distributions from TrueBeam linacs, as well as establishing this framework as the platform to perform advanced electron beam related treatment planning research, such as DEAR.
3.3. Evaluation of small field electron Monte Carlo for DEAR on TrueBeam linacs

3.3.1. Introduction

DEAR employees the superposition of many small fields to obtain sharp beam penumbras over curved surfaces and dose homogeneity. DEAR requires the knowledge of accurate small field dose distributions to produce accurate plans. Previous studies (Hu et al 2008, Xu et al 2009) have shown that the eMC algorithm can provide dose distributions for standard field sizes that are accurate to within 3% and 2 mm. For field sizes smaller than 3x3 cm$^2$ Hu et al. showed that disagreement in output factors can reach 8%. Since clinical electron beam therapy rarely used field sizes smaller than 3x3 cm$^2$, there is no impetus to improve upon this current limitation of the eMC algorithm as it has been optimized for large field dose calculation.

The accuracy of dose distributions in DEAR is a direct consequence of the accuracy of the small field dose kernels used. Thus the purpose of this study was to develop and validate a method to accurately model small field electron beam Monte Carlo for TrueBeam, explore the range of limitations of the application of small field dose distributions, and provide preliminary aperture and path optimizations for DEAR.

3.3.2. Materials and Methods

Small square (1x1, 2x2, and 3x3 cm$^2$), circular (1, 2, and 3 cm diameter), and rectangular (1x10 and 3x10 cm$^2$) field sizes were investigated for electron energies of 6, 9,
12, and 16 MeV on a Varian TrueBeam linac. Cerrobend cut-outs were made in-house. The square and circular cut-outs were made for a 6x6 cm$^2$ electron applicator (Figure 42) and the rectangular cut-outs were made for a 15x15 cm$^2$ applicator (Figure 9). Measurements of the physical dimensions of the cut-outs using calipers was done, which are shown in Table 10. These values were used in the subsequent simulations and calculations to ensure accurate modeling of the cut-out component. Differences in the cut-out size to the nominal dimensions were within 1 mm at SSD.

![Figure 42. Square (1x1, 2x2, and 3x3 cm$^2$) and circular (1, 2, and 3 cm diameter) Cerrobend cut-outs produced for a 6x6 cm$^2$ applicator.](image-url)
Table 10. Size for the nominal and actual cut-out size measured at 95 cm. Cut-outs are defined as the cross-plane x in-plane dimension. Almost all cut-out sizes displayed differences in cut-out within 1 mm at SSD to the nominal field size, except the 3x10 cm² rectangular cut-out, where the in-plane dimension was 5 mm larger than nominal. These differences were taken into account for the subsequent simulations and calculations.

<table>
<thead>
<tr>
<th>Cut-out Type</th>
<th>Field size</th>
<th>Nominal</th>
<th>Actual</th>
</tr>
</thead>
<tbody>
<tr>
<td>Square (cm²)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1x1</td>
<td>0.95x0.95</td>
<td>0.84x0.78</td>
<td></td>
</tr>
<tr>
<td>2x2</td>
<td>1.90x1.90</td>
<td>1.72x1.82</td>
<td></td>
</tr>
<tr>
<td>3x3</td>
<td>2.85x2.85</td>
<td>2.74x2.67</td>
<td></td>
</tr>
<tr>
<td>Circular (cm diameter)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>0.95</td>
<td>1.07</td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>1.90</td>
<td>1.85</td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>2.85</td>
<td>2.72</td>
<td></td>
</tr>
<tr>
<td>Rectangular (cm²)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1x10</td>
<td>0.95x9.50</td>
<td>0.82x9.42</td>
<td></td>
</tr>
<tr>
<td>3x10</td>
<td>2.85x9.50</td>
<td>2.9x9.95</td>
<td></td>
</tr>
</tbody>
</table>

Measurements, Monte Carlo simulations, and calculations from the treatment planning system were acquired. Previous validation of an end-user Monte Carlo simulation framework for the Varian TrueBeam Linac using vendor provided phase space files was performed for standard field sizes 6x6, 10x10, 15x15, 20x20, and 25x25 cm² for energies 6, 9, 12, 16, and 20 MeV with good agreement mostly within 2%/1mm. The same framework and agreement criteria was extended to the small field study for both the simulation and calculation, which is described in section 3.2.2.

3.3.2.1. Measurements

Cerrobend cut-outs were manufactured in-house for the specified nominal field sizes. Measurement percent depth dose curves and orthogonal profiles were acquired in
a 48x48x41 cm³ water phantom with a 3D scanning system (Blue Phantom2, IBA Dosimetry, Schwarzenbruck, Germany). Output factors were acquired in a smaller tabletop water phantom 1D water phantom. A detector with a small active area was preferred as to minimize convolution effects along steep dose gradients. Thus an electron field diode detector (EFD3G, IBA Dosimetry, Schwarzenbruck, Germany) with an active area of 2 mm diameter and thickness of 0.06 mm was chosen. It was expected that the dosimetry and output of the electron beam would be significantly degraded due to the lack of lateral scatter equilibrium in small field sizes when compared to standard dosimetry.

3.3.2.2. Monte Carlo simulations

Dose per primary particle was scored in a 30×30×20 cm³ water phantom with a voxel size of 0.25×0.25×0.2 cm³ in the X (cross-plane), Y (in-plane), and Z (depth) dimension. Approximately 760 million particle histories from the phase space were required such that the statistical precision of the dose was less than 1% at the depth of maximum dose $d_{\text{max}}$ for the chosen voxel size.

3.3.2.3. eMC algorithm calculations

Dose distributions for energies 6, 9, 12, and 16 MeV and field sizes as previously listed in Table 10 were calculated using the clinically available dose calculation algorithm eMC v.11 with a calculation grid size of 1 mm, uncertainty of <1%, and
medium level (i.e. kernel size is equal to the grid size, in this case 1 mm) 3D Gaussian-weighted smoothing.

3.3.2.4. Comparison metrics

3.3.2.4.1. Percent depth dose and profiles

Central axis percent depth dose and orthogonal in-plane and cross-plane profiles at depths $R_{100}$, $R_{50}$, $R_p$, and $R_{p^+}$ were measured and gathered from the simulation and calculation volumes. The energy-specific range depths were defined for a standard 15x15 cm² applicator and are shown in Table 5.

3.3.2.4.2. Cut-out output factors

Cut-out output factors were calculated as the ratio of the dose at reference depth $R_{100}$ for a specific energy and field size to the dose at reference depth $R_{100}$ for the same energy and a reference field size of 6x6 or 15x15 cm². Dose from measurements and calculations were assessed for a fixed number of monitor units, while simulation used the dose per incident particle instead of the dose per monitor unit (Zhang et al 1999).

Uncertainties in the simulations and calculations was within 1%, while measured charge values varied within 0.5%. Output factor measurements for small fields are known to be susceptible to setup errors due to the much reduced or nonexistent flat region on the central axis (Turian et al 2004). Thus uncertainties due to measurement setup errors were assessed. Additionally, a sensitivity analysis of the precision of the cut-
out size on the output factor was undertaken. A similar methodology had been previously used by Charles et al. for very small photon fields (Charles et al 2014).

3.3.2.5. Applications: Advantages and limitations for DEAR

The previous section aimed to validate the small field dosimetric accuracy of the Monte Carlo simulations. Therefore, it was assumed that field sizes in between those (e.g. 4x4 and 5x5 cm$^2$) would be accurately modeled in the Monte Carlo simulation as well.

Dynamic electron radiation therapies such as DEAR utilize the superposition of small fields to provide target conformity and fluence modulation. In the feasibility study described in section 2, surface variations in the inferior-superior direction were not investigated. There are several options to compensate for such variations. A trapezoid cut-out could be used instead of a rectangular slit-field, as implemented in EAT. Another option is to employ the combination of couch longitudinal motion and smaller cut-outs to realize a virtual scanning beam. This section aims to demonstrate the feasibility of virtual scanning mode using small fields and provide evaluation of the results with respect to the applications for DEAR.

3.3.2.5.1. Virtual scanning beam

Monte Carlo simulations were performed using validated Varian TrueBeam phase space files for electron beam energies of 6, 9, 12, and 16 MeV and the square and
circular cut-outs ranging from 1x1 cm$^2$ or 1 cm diameter to 5x5 cm$^2$ or 5 cm diameter in increments of 1 cm. Fluctuations in the 3D dose distributions (kernels) resulting from the uncertainty in the Monte Carlo simulation The kernels were smoothed radially with an averaging kernel in each 2D plane. The kernels were then used for subsequent dose calculations outside the Monte Carlo environment, following the convolution procedure outlined in section 2.2.1. Comparison of the reference 10x10 cm$^2$ dose distributions and with those resulting from composite square fields with static step and shoot and scanning beam deliveries for square fields realized as the convolution of kernels and scanning pattern. Preliminary beam weight and pattern optimization were also performed. The goal of the beam weight optimization was to obtain the same dose distribution as the reference. Preliminary scanning trajectory pattern was chosen to be a row scanning trajectory along the cross-plane direction.

### 3.3.2.5.1.2. Composite fields

Small field dose kernels were used to generate a composite dose distribution covering a 10x10 cm$^2$ field size by using either a static step-and-shoot or a dynamic convolution approach (Figure 43). In the dynamic approach, the preliminary scanning pattern was linear scanning in the cross-plane direction and the beams were chosen such there was no overlap on the surface (SSD = 100 cm). Therefore, only composite fields using the 1x1, 2x2, and 5x5 cm$^2$ kernels were investigated. Both equal and optimized
beam weights were investigated for the convolution approach. Resulting dosimetry including PDDs and profiles were compared to that of the reference 10x10 cm² field.

Figure 43. Composite small fields (white squares) are used to obtain the reference 10x10 cm² (blue square). Left: static step and shoot mode. Right: dynamic delivery (scanning mode) with a fixed trajectory (row scanning). Both equal and optimized beam weights were investigated.

3.3.2.5.1.3. Optimization of overlap dose

Due to the divergence of the electron beam, abutting fields will produce heterogeneous dose distributions at the interface of the two beams resulting in hotter and colder regions as a function of depth. Since the composite fields still suffer from abutment at the interface of the scanning trajectories, the linear scanning pattern for multiple rows was investigated for optimal overlap as a function of cut-out size and energy at the depth of R₁₀₀ (Figure 44). Similar optimization had been previously used in MERT to optimize gap separation (Eldib et al 2010). Optimization of the overlap dose
was investigated at the depth of $R_{100}$, where the overlap dose on the central axis in the orthogonal profile was targeted to be within 2\% of the reference level.

![Figure 44. Small circular field scanning without (left) and with (right) overlap.](image)

### 3.3.3. Results

#### 3.3.3.1. Small field dosimetry

**3.3.3.1.1. Percent depth dose curves**

The measured, simulated, and calculated central axis depth dose curve for all energies and cut-out sizes are shown in Figures 45 – 47. Compared to the PDDs from a standard field 15x15 cm$^2$, the $R_{100}$ moves towards the surface as the field size decreases, leading to a shallower fall-off slope. This due to the decrease in lateral electronic equilibrium as the field size decreases (Podgorsak and IAEA 2005). $R_{100}$, $R_{50}$, and $R_F$ for the measured square small field PDDs and all energies are summarized in Table 11. The shift for $R_{100}$ is greater than for $R_{50}$ and $R_F$ as compared to the values listed in Table 5 and thus the change in PDD is non-linear, and deeper depths display smaller shifts resulting in the shallower fall-off region.
Agreement in the build-up region (depths $\leq R_{100}$) was generally better for the Monte Carlo simulations than the calculations using the eMC algorithm, with the simulations underestimating the surface dose on average by 2%. The surface dose tended to be overestimated by the calculations, however, they were almost all within 2%, except for the rectangular field size, where differences of up to 7% were observed at the surface. Excluding voxels at depths $\leq 0.5$ cm lead to differences within 2% across almost all cases for both simulations and calculations in the build-up region. In the fall-off region ($R_{100} \leq$ depths $\leq R_p$), simulations agreed with measurements mostly within 2%, while the difference with the calculations were mostly with 5%, with a maximum difference of 12% at a depth of 1.5 cm observed for the 6 MeV circular 1 cm diameter case. In the fall-off region, agreement was within 2%, except for the 6 MeV 3x10 and 3 cm diameter, which displayed differences of up to 4.5% for both simulations and calculations. Generally, simulations and calculations for all field sizes at 6 MeV displayed the worst agreement with the measurement. Distance-to-agreement in the fall-off region for the simulations were well within 0.1 cm, while calculations agreed to within 0.2 cm. For a gamma index with a 2% and 0.1 cm criteria, the following energy-specific pass rates were observed in the build-up region: 55 and 95% (6 MeV), 79 and 91% (9 MeV), 89 and 95% (12 MeV), and 95 and 95% (16 MeV) for calculations and simulations, respectively. In the fall-off region, the following pass rates were observed:
55 and 98% (6 MeV), 64 and 97% (9 MeV), 58 and 98% (12 MeV), and 63 and 99% (16 MeV) for calculations and simulations, respectively. Pass rates in the Bremsstrahlung region were 100% for both simulations and calculations.

Figure 45. Central axis percent depth dose curves for all energies 6, 9, 12, and 16 MeV (top left to bottom right) and square cut-outs 1x1, 2x2, and 3x3 in a 6x6 cm² applicator. Measurements (solid), Monte Carlo simulations (circles), and calculations (dashed) are compared. Each PDD is normalized to its respective R_{100}, and for a given energy and field combination the measurement, Monte Carlo, and calculation PDDs are scaled to the same value.
Figure 46. Central axis percent depth dose curves for all energies 6, 9, 12, and 16 MeV (top left to bottom right) and circular cut-outs with 1, 2, and 3 cm diameter in a 6x6 cm² applicator. Measurements (solid), Monte Carlo simulations (circles), and calculations (dashed) are compared. Each PDD is normalized to its respective $R_{100}$, and for a given energy and field combination the measurement, Monte Carlo, and calculation PDDs are scaled to the same value.
Figure 47. Central axis percent depth dose curves for all energies 6, 9, 12, and 16 MeV (top left to bottom right) and rectangular cut-outs 1x10 and 3x10 cm$^2$ in a 15x15 cm$^2$ applicator. Measurements (solid), Monte Carlo simulations (circles), and calculations (dashed) are compared. Each PDD is normalized to its respective $R_{100}$, and for a given energy and field combination the measurement, Monte Carlo, and calculation PDDs are scaled to the same value.
Table 11. Select ranges $R_{100}$, $R_{50}$, and $R_p$ from measurement data for square small fields 1x1, 2x2, and 3x3 cm$^2$.

<table>
<thead>
<tr>
<th>Energy (MeV)</th>
<th>Field size (cm$^2$)</th>
<th>$R_{100}$ (cm)</th>
<th>$R_{50}$ (cm)</th>
<th>$R_p$ (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>1x1 cm$^2$</td>
<td>0.1</td>
<td>1.5</td>
<td>2.6</td>
</tr>
<tr>
<td></td>
<td>2x2 cm$^2$</td>
<td>0.7</td>
<td>2.2</td>
<td>2.9</td>
</tr>
<tr>
<td></td>
<td>3x3 cm$^2$</td>
<td>1.1</td>
<td>2.3</td>
<td>3.0</td>
</tr>
<tr>
<td>9</td>
<td>1x1 cm$^2$</td>
<td>0.1</td>
<td>1.8</td>
<td>3.0</td>
</tr>
<tr>
<td></td>
<td>2x2 cm$^2$</td>
<td>0.9</td>
<td>2.9</td>
<td>4.3</td>
</tr>
<tr>
<td></td>
<td>3x3 cm$^2$</td>
<td>1.5</td>
<td>3.4</td>
<td>4.4</td>
</tr>
<tr>
<td>12</td>
<td>1x1 cm$^2$</td>
<td>0.3</td>
<td>2.1</td>
<td>3.6</td>
</tr>
<tr>
<td></td>
<td>2x2 cm$^2$</td>
<td>1.1</td>
<td>3.7</td>
<td>5.5</td>
</tr>
<tr>
<td></td>
<td>3x3 cm$^2$</td>
<td>1.7</td>
<td>4.5</td>
<td>6.1</td>
</tr>
<tr>
<td>16</td>
<td>1x1 cm$^2$</td>
<td>0.5</td>
<td>2.6</td>
<td>4.3</td>
</tr>
<tr>
<td></td>
<td>2x2 cm$^2$</td>
<td>1.1</td>
<td>4.4</td>
<td>6.5</td>
</tr>
<tr>
<td></td>
<td>3x3 cm$^2$</td>
<td>1.7</td>
<td>5.5</td>
<td>7.9</td>
</tr>
</tbody>
</table>

3.3.3.1.2. Orthogonal profiles

Figures 48 – 51 show the cross-plane and in-plane profiles for the square, circular, and rectangular cut-outs. Field sizes at $R_{100}$ agreed on average to within 1.0 mm and 1.9 mm for the simulations and calculations, respectively. Penumbra agreed at $R_{100}$ agreed on average to within 1.2 and 1.3 mm for the simulation and calculation, respectively. The smallest field sizes produced the largest disagreements of up to 2 mm. Profiles generally agreed well, with similar gamma pass rates at $R_{100}$ for both simulations and calculations. Gamma pass rates were greater than 93% (calculations) and 95% (simulations) for field sizes 2x2 and 3x3 cm$^2$, 2 cm and 3 cm diameter, and 3x10 cm$^2$. 

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Field sizes 1x1 cm², 1 cm diameter, and 1x10 cm² displayed pass rates of 84, 67, and 81% for the calculations and 92, 66, and 80% for the simulations.

Figure 48. Cross-plane profiles at R_{100}, R_{50}, and R_p for all energies 6, 9, 12, and 16 MeV (rows) and square cut-outs 1x1, 2x2, and 3x3 cm² in a 6x6 cm² applicator (columns). Measurements (solid), Monte Carlo simulations (circles), and calculations (dashed) are compared. Profiles at R_{100}, R_{50}, and R_p are normalized to 100%, 50%, and 10% at the value on the central axis, respectively.
Figure 49. Cross-plane profiles at $R_{100}$, $R_{50}$, and $R_p$ for all energies 6, 9, 12, and 16 MeV (rows) and circular cut-outs 1, 2, and 3 cm diameter in a 6x6 cm² applicator (columns). Measurements (solid), Monte Carlo simulations (circles), and calculations (dashed) are compared. Profiles at $R_{100}$, $R_{50}$, and $R_p$ are normalized to 100%, 50%, and 10% at the value on the central axis, respectively.
Figure 50. Cross-plane profiles at $R_{100}$, $R_{50}$, and $R_p$ for all energies 6, 9, 12, and 16 MeV (rows) and rectangular cut-outs 1x10 and 3x10 cm$^2$ in a 15x15 cm$^2$ applicator (columns). Measurements (solid), Monte Carlo simulations (circles), and calculations (dashed) are compared. Profiles at $R_{100}$, $R_{50}$, and $R_p$ are normalized to 100%, 50%, and 10% at the value on the central axis, respectively.
3.3.3.1.3. Cut-out output factors

Measurement, Monte Carlo, and calculation Cut-out output factors were calculated according to equation and are shown in Table 12. Uncertainties in the Monte
Carlo simulation, and calculations were both within 1%. Measurement uncertainties stemming from statistical noise in the charge measurement were found to be within 0.5%.

Table 12. Summary of output factors for measurement, Monte Carlo, and calculation. Square and circular field size output was normalized to that from a 6x6 cm$^2$ field size, while rectangular field sizes were normalized to the output from a 15x15 cm$^2$ field size. Absolute differences between measurement and Monte Carlo and measurement and calculation are also shown. Differences larger than 2% are bolded.

<table>
<thead>
<tr>
<th>Energy (MeV)</th>
<th>Field size (cm$^2$)</th>
<th>Measurement</th>
<th>MC</th>
<th>Calculation</th>
<th>Difference Meas – MC (%)</th>
<th>Difference Meas – Calc (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>1x1</td>
<td>0.285</td>
<td>0.280</td>
<td>0.305</td>
<td>-0.5</td>
<td>2.0</td>
</tr>
<tr>
<td></td>
<td>2x2</td>
<td>0.738</td>
<td>0.732</td>
<td>0.743</td>
<td>-0.6</td>
<td>0.5</td>
</tr>
<tr>
<td></td>
<td>3x3</td>
<td>0.939</td>
<td>0.936</td>
<td>0.931</td>
<td>-0.3</td>
<td>-0.8</td>
</tr>
<tr>
<td></td>
<td>2 dia</td>
<td>0.714</td>
<td>0.699</td>
<td>0.725</td>
<td>-1.5</td>
<td>1.0</td>
</tr>
<tr>
<td></td>
<td>3 dia</td>
<td>0.894</td>
<td>0.888</td>
<td>0.893</td>
<td>-0.6</td>
<td>-0.1</td>
</tr>
<tr>
<td></td>
<td>3x10</td>
<td>0.975</td>
<td>0.963</td>
<td>0.981</td>
<td>-1.2</td>
<td>0.6</td>
</tr>
<tr>
<td>9</td>
<td>1x1</td>
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<td>0.221</td>
<td>0.254</td>
<td>0.8</td>
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<tr>
<td></td>
<td>2x2</td>
<td>0.646</td>
<td>0.632</td>
<td>0.630</td>
<td>-1.5</td>
<td>-1.6</td>
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<tr>
<td></td>
<td>3x3</td>
<td>0.873</td>
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<td>-1.3</td>
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<tr>
<td></td>
<td>2 dia</td>
<td>0.616</td>
<td>0.603</td>
<td>0.592</td>
<td>-1.3</td>
<td>-2.4</td>
</tr>
<tr>
<td></td>
<td>3 dia</td>
<td>0.819</td>
<td>0.802</td>
<td>0.804</td>
<td>-1.7</td>
<td>-1.5</td>
</tr>
<tr>
<td></td>
<td>3x10</td>
<td>0.946</td>
<td>0.935</td>
<td>0.937</td>
<td>-1.0</td>
<td>-0.9</td>
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<tr>
<td>12</td>
<td>1x1</td>
<td>0.243</td>
<td>0.238</td>
<td>0.258</td>
<td>-0.5</td>
<td>1.5</td>
</tr>
<tr>
<td></td>
<td>2x2</td>
<td>0.658</td>
<td>0.647</td>
<td>0.645</td>
<td>-1.2</td>
<td>-1.3</td>
</tr>
<tr>
<td></td>
<td>3x3</td>
<td>0.875</td>
<td>0.857</td>
<td>0.859</td>
<td>-1.8</td>
<td>-1.6</td>
</tr>
<tr>
<td></td>
<td>2 dia</td>
<td>0.628</td>
<td>0.609</td>
<td>0.609</td>
<td>-1.9</td>
<td>-1.9</td>
</tr>
<tr>
<td></td>
<td>3 dia</td>
<td>0.824</td>
<td>0.808</td>
<td>0.806</td>
<td>-1.6</td>
<td>-1.8</td>
</tr>
<tr>
<td></td>
<td>3x10</td>
<td>0.938</td>
<td>0.923</td>
<td>0.936</td>
<td>-1.5</td>
<td>-0.2</td>
</tr>
</tbody>
</table>
Measured and simulated cut-out output factor agreed to within 1.9% for all cut-outs, while calculated cut-out output factors agreed to within 2.4%.

### 3.3.3.1.3.1. Effect of measurement set-up on cut-out output factor

Additionally, measurement set-up induced errors were assessed: Output factor measurements were repeated with the addition of 1 and 2 mm shifts in either x or y dimension. Figure 52 shows the variation of the output factor as a function of x and y positioning error for a 1x10 cm² field size (i.e. x = along the short axis of the cut-out, 1 cm; x = along the long axis of the cut-out, 10 cm) for 6 and 16 MeV.

<table>
<thead>
<tr>
<th></th>
<th>1x1</th>
<th>0.323</th>
<th>0.320</th>
<th>0.338</th>
<th>-0.3</th>
<th>1.5</th>
</tr>
</thead>
<tbody>
<tr>
<td>2x2</td>
<td>0.755</td>
<td>0.746</td>
<td>0.744</td>
<td>-0.9</td>
<td>-1.1</td>
<td></td>
</tr>
<tr>
<td>3x3</td>
<td>0.924</td>
<td>0.916</td>
<td>0.916</td>
<td>-0.8</td>
<td>-0.8</td>
<td></td>
</tr>
<tr>
<td>2 dia</td>
<td>0.727</td>
<td>0.723</td>
<td>0.722</td>
<td>-0.4</td>
<td>-0.5</td>
<td></td>
</tr>
<tr>
<td>3 dia</td>
<td>0.891</td>
<td>0.881</td>
<td>0.882</td>
<td>-0.9</td>
<td>-0.9</td>
<td></td>
</tr>
<tr>
<td>3x10</td>
<td>0.965</td>
<td>0.959</td>
<td>0.958</td>
<td>-0.6</td>
<td>-0.7</td>
<td></td>
</tr>
</tbody>
</table>
Figure 52. Output factor as a function of (a) x and (b) y positioning error of the diode for energies 6 and 16 MeV. Differences in output factor of 3% can be observed for an error of -2 mm in the x-direction.

Introduction of an x or y positioning error leads to variations in output factor of up to 3% or 1%, respectively. Similar variations were observed for the 1 cm diameter field size in both the x and y directions, with variations in output factor of up to 7% for an error of 2 mm.

3.3.3.1.3.2. Effect of aperture size on cut-out output factor

Generally, patient specific cut-outs or non-standard small field cut-outs are made in-house. Cerrobend cut-outs are manufactured by the process of using a Styrofoam mold around which Cerrobend is poured. There is variability in this process and accurate cut-out size must be verified through measurement and corrections made if
necessary. The precision of the cut-out may be potentially significant for smaller fields: as the cut-out size decreases, even a small variation in the cut-out size of ±1 mm can lead to a larger change in the area of the cut-out, e.g., decreasing the field size of a 2x2 cm² by 1 mm on each side leads to a decrease in area of approximately 10%. Therefore, the output factor is expected to vary much more for smaller cut-outs than for larger cut-outs.

To ascertain the impact of the precision of the cut-outs on output, additional cut-outs were generated for nominal 1x1, 2x2, and 3x3 cm² field sizes, with cut-outs ±2 mm from the nominal size. At an SSD of 95 cm, the nominal cut-out size is 0.95x0.95, 1.9x1.9, and 2.85x2.85 cm², respectively. Varying the cut-out size by ±1 mm leads to variation on average of -10 to 4%, -4 to 3%, and -2 to 1% for the 1x1, 2x2, and 3x3 cm² cut-outs. As can be observed, the variation in output is not symmetric for a variation in cut-out size of +1 and -1 mm. This means that reducing the cut-out size can drastically reduce the output to the nominal output.

A tolerance of ±2% for the nominal output factor was targeted. Diode measurements for the cut-out output factors and the aforementioned depths were acquired. Figure 53 shows the relationship between the output factor and field size for all tested cut-outs.
Figure 53. Cut-out output factor as a function of cut-out size for cut-outs ±1-2 mm at SSD = 100 cm about the nominal field size for energies 6, 9, 12, and 16 MeV. Small variations in the cut-out size can have a substantial impact on the OF. Varying the cut-out size by ± 1 mm leads to variation on average of -10 to 4%, -4 to 3%, and -2 to 1% for the 1x1, 2x2, and 3x3 cm$^2$ cut-outs.

Through polynomial fitting, the range of cut-out sizes was determined such that the resulting cut-out output factor was within the stated tolerance of ±2%. To preserve an output factor within ±2%, cut-outs must be made within the range of -0.3 to 0.5 mm (1x1 cm$^2$), -0.4 to 0.7 mm (2x2 cm$^2$), -0.6 to 1.7 mm (3x3 cm$^2$) from the nominal cut-out size for all energies.
3.3.3.2. Virtual scanning beam

3.3.3.2.1. Composite fields

Monte Carlo simulations for energies 6, 9, 12, and 16 MeV for square and circular cut-outs (1x1, 2x2, 3x3, 4x4, and 5x5 cm² and 1, 2, 3, 4, and 5 cm diameter) in a 6x6 cm² applicator were generated. The single kernel PDD and cross-plane profiles at d_{max} for electron energies 6, 9, 12, and 16 MeV for square cut-outs are shown in Figure 54. As the field size decreases, the PDD shifts substantially towards the surface, and the fall-off slope becomes less steep.
Figure 54. PDDs and cross-plane profiles for square 1x1 to 5x5 cm² kernels for 6, 9, 12, and 16 MeV (top to bottom).
A reference 10x10 cm² field was compared to composite static step and shoot and dynamic irradiations with equal and optimized beam weightings. As an example, a 2D dose distribution at \( d_{\text{max}} \) for a 6 MeV beam using a 1x1 cm² kernel to cover the reference field are shown in Figure 55. As the fields diverge, contributions from neighboring fields influence dose in a given field. Further, field abutments are visible on either side of the kernel in the static step and shoot case, while in the dynamic case abutment is only seen where the rows meet.
Figure 55. 2D dose distribution at $d_{\text{max}}$ comparing the composite small fields using a 6 MeV 1x1 cm$^2$ kernel (a) – (c) to the (d) reference 10x10 cm$^2$ field.

PDDs along the central axis for the composite cases for all kernels (1x1, 2x2, and 5x5 cm$^2$) and energies were compared to that of the reference case and are shown in Figure 56. Composite fields tended to restore the depth of maximum dose to that of the reference field. Static step and shoot and dynamic equal beam weight composite fields
displayed similar PDDs, while the dynamic optimal beam weight composite field tended to display a much degraded Bremsstrahlung region. The 5x5 cm$^2$ kernel showed the least amount of degradation to the PDD especially in the build-up and Bremsstrahlung region, with the largest deviation seen at the surface for the 16 MeV energy and 5x5 cm$^2$ kernel, where the composite fields reduced the surface dose by about 4%. As the kernel size decreases, the build-up region becomes flatter and the surface dose increases. Further, the Bremsstrahlung region increases by up to 40% for the 6 MeV energy 1x1 cm$^2$ composite field cases. The increase in Bremsstrahlung tail is probably due to the increased contribution of the scatter from the face of the aperture as the cut-out size decreases relative to the primary radiation.
Figure 56. PDDs along the central axis of composite fields comprised of fields generated with small field cut-outs using static step-and-shoot, and dynamic scanning with equal beam weight, and optimized beam weight compared to a reference 10x10 cm$^2$ for the same energy.
On one hand, the use of the 1x1 cm² kernel degraded the central axis PDD, on the other hand, it provides a penumbra which is comparable to the reference for most cases, as shown in Figures 57 and 58 for both the cross-plane and in-plane profiles at d_{max}. The reference beam penumbra was 1.1, 1.2, 1.3, and 1.1 mm for energies 6, 9, 12, and 16 MeV respectively. The static step and shoot and equal beam weight dynamic composite fields displayed similar penumbras, except for the 5x5 cm², while the optimal beam weight was able to improve the penumbra for the 1x1 and 2x2 cm² while regaining the uniformity i.e. flatness of the reference 10x10 cm² field. The static-step and shoot homogeneity displays the worst homogeneity as the 4 field abutment on the central axis causes a substantial hot spot.

Static step-and-shoot composite fields suffer from field abutments both in the cross-plane and in-plane direction, while the dynamic scanning composite fields suffer from field abutments only in the in-plane direction. While optimization improves the dose homogeneity across the field, variations of up to 4% can be observed. Further, the in-plane penumbra for the dynamic delivery is degraded due to the convolution of the kernels.
Figure 57. Cross-plane profiles: Composite fields comprised of fields generated with small field cut-outs using static step and shoot, and dynamic scanning with equal beam weight, and optimized beam weight compared to a reference 10x10 cm² for the same energy.
Figure 58. In-plane profiles: Composite fields comprised of fields generated with small field cut-outs using static step and shoot, and dynamic scanning with equal beam weight, and optimized beam weight compared to a reference 10x10 cm² for the same energy.
3.3.3.2.2. Optimization for overlap dose

Two linear scans of 10 cm length with and without overlap for circular and square fields were assessed. Squares generally yielded better penumbra, but due to the radial symmetry of circular field sizes they may give better dosimetric results for complex scanning trajectories.

The effect of overlap is shown in Figure 59 for a 6 MeV 5 cm diameter cut-out. Interpolation between discrete overlaps provides an optimal overlap of 0.8 cm. This would provide, for this case, the most homogenous dose distribution at depth of maximum dose.

![Figure 59. Right: 2D dose distribution for two 6 MeV 5 cm diameter linear scans of 10 cm length with 0 and 1 cm overlap at the surface. Middle: The corresponding cross-plane profile for overlaps ranging from 0 to 2.5 cm in increments of 0.5 cm. Left: Plot of overlap dose at an off-axis distance of 0 cm as a function of overlap.](image)
The optimal overlap for each energy as a function of field size, where the dose on the central axis is 100% is shown in Table 13.

<table>
<thead>
<tr>
<th>Circular cut-out (cm diameter)</th>
<th>Optimal overlap (cm) for circular cut-outs</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>6 MeV</td>
</tr>
<tr>
<td>1</td>
<td>0</td>
</tr>
<tr>
<td>2</td>
<td>0.2</td>
</tr>
<tr>
<td>3</td>
<td>0.4</td>
</tr>
<tr>
<td>4</td>
<td>0.6</td>
</tr>
<tr>
<td>5</td>
<td>0.8</td>
</tr>
</tbody>
</table>

### 3.3.4. Discussion

#### 3.3.4.1. Small field dosimetry

Monte Carlo simulations were in better agreement with the measurement data than with the eMC algorithm. Measurements for the smallest field size of 1 cm diameter and 1x10 cm² are known to be challenging (Zhang et al 2013) as even small deviations in cut-out size and measurement set-up can impact the dose distribution characteristics. A framework utilizing the Monte Carlo generated small field dose kernels can be used in subsequent DEAR planning.
3.3.4.2. Composite fields dose for DEAR

Composite fields regained reference depth dose profiles for most energies and fields within 5%. Smaller kernels and higher energies increased dose in the build-up and Bremsstrahlung region (30%, 16 MeV and 1x1 cm$^2$), while reference $d_{\text{max}}$ was maintained for all energies in composite fields. Composite fields utilizing smaller kernels (<2x2 cm$^2$) maintained penumbra and field size within 0.2 cm, and flatness within 2 and 4% in the cross-plane and in-plane direction, respectively. Deterioration of penumbra for larger kernels (5x5 cm$^2$) was observed for the dynamic scanning modes, while the static step and shoot maintained the same penumbra as the reference. Balancing desirable dosimetry and efficiencies suggests that smaller kernels can be used at edges and larger kernels in the center of the target. Beam weight optimization improves cross-plane penumbra (0.2 cm) and increases the field size (0.4 cm) on average. In-plane penumbra and field size remain unchanged. Overlap depends on kernel size and optimal overlap results in flatness ±2%.

A smaller field size could be utilized at target edge while larger field sizes can be used in the center to balance advantages and disadvantages in dosimetry and delivery efficiency. Further, flatness of the profile at $d_{\text{max}}$ could be improved by optimizing the overlap or gap between kernels.

Virtual scanning is possible on Varian TrueBeam linacs in Research Mode. Unlike previous scanning electron methods discussed in section 1.4.3.3., the robotic
feedback control of the linac allows for beam holds if the actual trajectory does not agree with the expected trajectory, as shown in section 2. Additionally, the virtual scanning mode with small fields could utilize scattering foil free beams for DEAR. The feasibility of using scattering foil free beams clinically has been recently reported for applications in MERT (Connell et al 2012, Eldib et al 2014).

3.3.5. Conclusion

We were able to characterize the small field dosimetry using Monte Carlo with good agreement with the measurement data, and thus offers the opportunity for treatment planning of DEAR. Dynamic electron beam therapy in virtual scanning mode was found to be feasible by employing small fields to achieve desired dose distributions and acceptable efficiencies.
3.4. Characterization of electron Monte Carlo phase space files

3.4.1. Introduction

The phase space file captures all traversing particle’s information including particle type $p$, energy $E$, position $x$, $y$, and $z$, and direction cosines $U$, $V$, and $W$ at a given plane in space, which is used as the input the Monte Carlo simulation. Essentially the phase space file is a list of data containing information on all the particles that cross a defined scoring region. Thus, disadvantages of using the phase space files include their large size, often on the order of 20 GB to achieve reasonable uncertainties in the dose scoring simulation. This makes them cumbersome to distribute. The end-user is thus also limited to the number of particles given in that particular phase space file: Recycling particles from the same phase space file may introduce bias, thus it is more conservative to have independent particles. Further, files may still contain statistical noise and do not allow for small variations in linac geometry.

Previous research has shown that parameterization of phase space data i.e. finding the statistical description of all attributes, can reduce the disk space requirement and shorten computation time (Ma et al 1997). Ma et al. used a multiple-source model, breaking down the distributions both from their primary and scattered origin. The idea is that particles coming from different components of the linac could be treated as if they are coming from different sources and thus have similar particle attribute distribution.
The following sections will detail a different methodology to characterize TrueBeam phase space files. The particles scored in the phase space file were not subdivided into primary and scattered sources, rather the source of the phase space file was treated as a “black box” and an acceptable parameterization of the phase space files was sought. Further, the described methodology was evaluated in terms of a direct comparison between original and pseudo phase space file and also by comparing their resulting dose distributions for energies 6, 9, 12, and 16 MeV and all standard field sizes 6x6, 10x10, 15x15, 20x20, and 25x25 cm$^2$.

### 3.4.2. Materials and Methods

MATLAB code was implemented to read the original IAEA phase space file format as well as perform the characterization and subsequent validation.

#### 3.4.2.1. Varian TrueBeam IAEA phase space file

The TrueBeam phase space file is scored on a flat plane at a constant $z$ of 26.7 cm. The phase space is scored on the plane extending from -6.5 to 6.5 cm in both x and y direction.

The binary IAEA phase space files for 6, 9, 12, and 16 MeV were downloaded from MyVarian.com. Endianness is little endian and each particle is described in 21 bytes: 1 byte (int8) for particle type (either 1 for electron, 2 for photon, or 3 for positron), 4 bytes (real*4) for energy in MeV, 4 bytes (real*4) for x position in cm, 4 bytes (real*4) for y position in cm. 4 bytes (real*4) for z position in cm, and 4 bytes (real*4) for weight.
for y position in cm, 4 bytes (real*4) for x direction cosine \( U \), and 4 bytes (real*4) for y direction cosine \( V \). The sign of the z direction cosine \( W \) is not recorded, because only forward directed particles are counted i.e. the sign of \( W \) is always positive. The direction cosines represent cosine of the angle the direction vector makes with either the x, y, or z basis vector. The direction cosines are related by the following Equation 5:

\[
U^2 + V^2 + W^2 = 1
\]

Equation 5

3.4.2.2. Characterization

The probability of a specific particle with a specific set of attributes can be described by the multivariate probability distribution \( P(\text{Type}, E, x, y, U, V) \). Sampling a particle from such a distribution can be computationally intensive, especially if a simple accept-reject algorithm for the random sampling is chosen. Reducing the dimensionality and employing a decision tree sampling scheme is therefore advantageous.

Therefore, the first step was to determine the dependencies between the particle attributes \( \text{Type}, E, x, y, U, V \). This was done by performing pairwise correlation analysis for all attributes. The multivariate probability distribution was equivalently described as \( P(\text{Type}, E, r, \theta, U, V) \). Further, this probability distribution can be broken down for each particle type, thus, particle type \( \text{Type} \) can be separated from the multivariate distribution. Pairwise correlation analysis was thus performed for the remaining attributes \( E, r, \theta, U, V \).
for each particle type. The results from the correlation analysis thus enabled us to separate the multivariate probability distribution into multiple conditional probabilities. Piece-wise parameterized probability distributions were then generated. For this study, the distributions were piece-wise to ensure accurate generation of the phase space files. Finally, a random sampling scheme was developed to generate the pseudo phase space files.

### 3.4.2.3. Validation of pseudo phase space files

Pseudo phase space files were generated by random sampling of the piece-wise parameterized distributions. Two validation approaches were used: (1) Direct comparison of the original and pseudo phase space histograms for all particle attributes was performed using the correlation coefficient as the metric of similarity; (2) Comparison of the original and pseudo phase space files generated dose distributions in terms of their percent depth dose and orthogonal profiles at depths $R_{100}$, $R_{50}$, and $R_p$ was also performed. For the PDD, dose differences and range differences were assessed, and for the profiles field size, penumbra, and flatness were assessed. A gamma index criteria of 2% and 1 mm was used to assess agreement between original and pseudo phase space file generated dose distributions. Varian TrueBeam phase space files of energy 6, 9, 12, and 16 MeV and field sizes of 6x6, 10x10, 15x15, 20x20, and 25x25 cm$^2$ were tested.
3.4.3. Results

3.4.3.1. Correlation analysis

For visualization purposes, Figure 60 shows a 2D fluence plot scored in pixel sizes of 0.1x0.1 cm$^2$ for both electrons, photons, and positrons for a subset of particles in a 6 MeV phase space file. As expected, the electron fluence is spatially broad and uniform, an artifact of the dual scattering foil. The photon fluence is forward peaked, while the positron fluence lacks any visible trend.
Figure 60. 2D fluence plots (# of particles/0.1x0.1 cm$^2$) for electrons, photons, and positrons.
The energy histogram evaluated at the center of the scoring plane (i.e. $x = 0$ cm and $y = 0$ cm) shows that the electron energy distribution can be approximated by a Gaussian with a mean energy approximate to the nominal energy and a smeared out low energy tail. At the edge of the scoring plane (i.e. $x = 6$ cm and $y = 6$ cm) the energy spectrum shifts toward the lower energies. The direction histogram of the direction cosines show that particles scored at the center of the scoring plane diverge minimally from the central axis, while particles at the edge of the field tended to have a larger divergence.

The multivariate probability distribution can be equivalently described as $P(\text{Type}, E, r, \theta, U, V)$. 2D positions were transformed to polar coordinates and correlation analysis was performed. Pairwise correlation analysis of all particle attributes, $E, r, \theta, U, V$ was evaluated. Energy, and directions cosines $U$ and $V$ were correlated to the radius. Energy was weakly correlated to $\theta$. $U$ and $V$ were correlated to $\theta$. To remove the dependence of $U$ and $V$ with $\theta$, the direction cosines were transformed to a particle specific coordinate system. Figure 61 depicts the particle and its direction in both the room and transformed coordinate systems. This primed coordinated system aligns its $z$-axis along a vector pointing from the virtual source (at $z = 0$ cm) to the particle’s position in the phase space plane ($z = 26.7$ cm), making it essentially a rotating coordinate system that is independent of $\theta$ and specific to each particle. In other words, $U$ and $V$
transformed to $U'$ and $V'$ were then uncorrelated to $\theta$. Thus, the energy and $U'$ and $V'$ distributions were found to be uncorrelated for a given radius.

Figure 61. Description of the primed coordinate system. A particle in the phase space plane (blue point) and the virtual source (red point) are depicted in the room coordinate system ($x$, $y$, and $z$) and the primed coordinate system ($x''$, $y''$, and $z''$). The line depicts the particles direction described by the direction cosines $U$, $V$, and $W$. The $z''$-axis aligns itself with the line that connects the virtual source with the particle position.

While it is clearly visible that the phase space file is not perfectly radially symmetric, the use of radial symmetry is also motivated by the following: The largest
field size (open field) is 40x40 cm$^2$. Backprojected to the x-jaw, the orthogonal and
diagonal physical jaw opening is approximately 10.7 and 15.1 cm (or equivalently, a
radius of 5.3 and 7.6 cm), respectively. Only 5% of photons and electrons are at radii >
5.3 cm. At these distances off-axis, the particles directions are generally titled away from
the central axis. This means there is a higher probability that these particles will not exit
the linac, but rather scatter and interact with the x and y jaws. Further, since the largest
standard jaw setting is 22x2 cm$^2$ for a 6 MeV electron beam, aforementioned effects may
not substantial impact collimated fields, as particles outside the physical jaw size are
mostly absorbed or blocked by the collimators, however, due to the highly scattering
nature of electrons, this assumption will be investigated.

3.4.3.2. Parameterization

The correlation analysis guided the generation of the probabilities and the
structure of the subsequent sampling algorithm. Distributions for each particle attribute
will be discussed in detail herein. First, the multivariate probability distribution can be
equivalently described as using polar coordinates $P(\text{Type}, E, r, \theta, U, V)$. From the
correlation analysis, the assumption of radial symmetry i.e. for a given radius bin $r$, the
particle’s energy and direction did not vary as a function of polar angles $\theta$. Further, the
particles energy and direction for a given radius were uncorrelated.
3.4.3.2.1. Type

The simplest distribution is that of particle type: Electrons, photons, and positrons were tallied and the resulting distribution for all energies is shown in Figure 62. Electrons dominate the particle type distributions with a frequency of about 70-80%, while photons account for about 20-30% over the range of energies. Positrons account for just a very small fraction and were mostly ignored.

![Particle Type PDF](image)

**Figure 62. Particle type distribution.**

3.4.3.2.2. Position

There are three particle position probability distributions as a function of radius. The radius histograms were binned in increments of 0.1 cm. As an example, the particle
position distribution for a 16 MeV phase space file is shown in Figure 63. Electron position frequency is almost constant up to approximately a radius of 3.5 cm, while photon position frequency is highest in the center of the phase space.

Figure 63. Particle position distributions for a 6, 9, 12, and 16 MeV phase space.
3.4.3.2.3. Energy

Three energy distributions (electron, photon, and positron) were generated in increments of 0.1 MeV at each radius bin of 0.1 cm. Figure 64 and 65 show the energy distributions for electrons and photons for 6, 9, 12, and 16 MeV phase space files at the center (x = 0 cm and y = 0 cm) of the scoring plane, respectively.

![Electron Spectrum](image)

Figure 64. Electron energy distributions for 6, 9, 12, and 16 MeV phase space files at the center of the scoring plane. Normalization is arbitrary.
Figure 65. Photon energy distributions for 6, 9, 12, and 16 MeV phase space files at the center of the scoring plane shown as (a) linear-linear and (b) log-linear plots. Normalization is arbitrary.

3.4.3.2.4. Direction

Three direction distributions per radius were generated in increments of 0.1 for $U$ and $V$. As the correlation analysis showed, the distributions of the direction cosines $U$ and $V$ are correlated to $\theta$ at a given radius, thus they must be first transformed to the particle’s own coordinate system. $U'$ and $V'$ are thus uncorrelated to the $\theta$ for a given radius. $U'$ and $V'$ are however correlated and must be sampled at the same time. As such the particle direction distributions are 2D. The direction of the electrons and photons in the center of the scoring plane is mostly forward peaked i.e. the 2D particle direction distribution has the highest frequency where $U$ and $V$ are almost equal to 0 and thus $W$ is almost equal to 1.
3.4.3.3. Generation of pseudo phase space files

Random sampling from non-uniform distributions was accomplished by implementation of the accept-reject algorithm. Generation of the pseudo phase space files from the conditional probability tree is shown schematically in Figure 66.

![Diagram of pseudo phase space generation sampling scheme](image)

**Figure 66.** Pseudo phase space generation sampling scheme. First, a particle type is randomly sampled from the non-uniform distribution. The appropriate particle position distributions is then used to sample the radius. The particle type and radius then dictate which particle energy and direction distribution to sample from. Finally, the polar angle is randomly sample from a uniform distribution (not shown).

First, the particle type is randomly sampled from the non-uniform particle type distribution. Then, depending on the particle type, the appropriate fluence distribution is chosen from which the radius \( r \) is sampled. Because of the radial symmetry, the polar angle \( \theta \) can be sampled uniformly from a range of \(-\pi\) to \(\pi\). For a given particle type and
radius, the particle energy and direction distributions can then be sampled. The sampled radius \( r \) then determines the energy and direction probability distribution from which the particle’s energy and direction is sampled from. The direction is sampled from a 2D direction distribution in the particle’s own coordinate system. The sampled radius \( r \) and polar angle \( \theta \) are then converted back to Cartesian coordinates.

For evaluation of the parameterization, the aforementioned sampling scheme is not implemented in real-time during Monte Carlo simulation runtime, rather, the pseudo phase generation is performed offline. The pseudo phase space and header follow that of the original IAEA formatted phase space files.

3.4.3.4. Validation

3.4.3.4.1. Direct comparison of phase space files

Figure 67 shows the direct comparison of the histograms of all particle attributes for both the original and pseudo phase space file for an energy of 6 MeV for electron and photons.
<table>
<thead>
<tr>
<th></th>
<th>Electrons</th>
<th></th>
<th>Photons</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Original</td>
<td>Pseudo</td>
<td>Original</td>
<td>Pseudo</td>
</tr>
<tr>
<td><strong>Energy</strong></td>
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<td><img src="image11" alt="Graph" /></td>
<td><img src="image12" alt="Graph" /></td>
</tr>
<tr>
<td><strong>Direction U</strong></td>
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<td><img src="image14" alt="Graph" /></td>
<td><img src="image15" alt="Graph" /></td>
<td><img src="image16" alt="Graph" /></td>
</tr>
<tr>
<td><strong>Direction V</strong></td>
<td><img src="image17" alt="Graph" /></td>
<td><img src="image18" alt="Graph" /></td>
<td><img src="image19" alt="Graph" /></td>
<td><img src="image20" alt="Graph" /></td>
</tr>
</tbody>
</table>
Figure 67. Histograms for a 6 MeV original (blue) and pseudo (red) phase space for all electron and photon attributes. The correlation coefficient is >0.98 for all attributes.

Differences in the histograms can be observed at the edge of the field: e.g. the original histogram of the electron x position, shows a sudden decrease in the original phase space when compared to the pseudo phase space. However, as previously mentioned and will be shown the next section, particles at these distances of axis rarely contribute to the dose distribution on the phantom as those particle histories are mostly terminated prior to exiting the linac treatment head. The original and pseudo phase space show a high level of similarity with correlation coefficients upwards of 0.98 for all particle attributes (Type, E, x, y, U, V) for all energies 6, 9, 12, and 16 MeV. This suggests that the parameterization is capable of accurately generating phase space data.

3.4.3.4.2. Comparison of resulting dose distributions

The resulting dose distributions for energies 6, 9, 12, and 16 MeV and field sizes 6x6, 10x10, 15x15, 20x20, and 25x25 cm² were compared to those from the original dose distribution. PDDs for energies 6, 9, 12, and 16 MeV for field sizes 6x6 and 25x25 cm² are shown in Figure 68. The gamma index was within 2%/1mm over the entire range of the PDD for all energies and field sizes.
Figure 68. PDDs originating from original (solid) and pseudo (dashed) phase space files. Energies 6, 9, 12, and 16 MeV (left to right with arbitrary normalization at depth of maximum dose for each energy) are shown for (a) 6x6 and (b) 25x25 cm² field size.

Pseudo orthogonal profiles at $R_{100}$, $R_{50}$, and $R_p$ were in agreement with those derived from the original phase space files (within a gamma index of 2% and 1 mm) for all field sizes 6x6 to 25x25 cm². Profiles at $R_{100}$, $R_{50}$, and $R_p$ for 6, 9, 12, and 16 MeV beams and subset of field sizes 6x6, 15x15, and 25x25 cm² are shown in Figure 69.
Figure 69. Profiles at $R_{100}$, $R_{50}$, and $R_p$ originating from original (solid) and pseudo (dashed) phase space files. Each profile is normalized to 100%, 50%, or 10% at the central axis value, respectively. Energies 6, 9, 12, and 16 MeV are shown for 6x6, 15x15, and 25x25 cm$^2$ field size.
3.4.4. Discussion

Histograms of the original and pseudo phase space file agree very well with correlation coefficients greater than 0.98 for all particle attributes. Dosimetric comparison between original and pseudo dose distributions yielded agreement within 2%/1mm for PDDs and profiles at all depths for all field sizes 6x6, 10x10, 15x15, 20x20, and 25x25 cm$^2$ and energies 6, 9, 12, and 16 MeV.

The current parameterization method uses piece-wise probability distributions of the particles attributes. This translates to a storage requirement of a few KB in comparison to the original phase space file which can occupy tens of GB, while still maintaining an accurate description of the phase space. Continuous fits will be investigated as this will further reduce the disk space requirement, reduce statistical noise in the phase space file, and improve computational efficiency.

Currently, the sampling scheme used to generate the pseudo phase space is not implemented in real-time during the MC simulation. One of the factors that can be improved is the sampling algorithm used: The accept-reject algorithm is well known to be inefficient when the function from which the sample are being taken from is sparse but contains a large peak at some location i.e. these distributions have long tails with low probabilities. Especially in the 2D sampling of the direction cosines $U'$ and $V'$ does this effect becomes even more prominent, as many rejections must take place before an acceptable sample is generated. The probability of rejection increases exponentially as a
function of the number of dimensions. Thus, implementation of adaptive rejection sampling will speed up the sampling process.

Another motivation to pursue parameterization is the idea of generating virtual energies. Currently, electron energies are only available in stepwise increments. While the parameterization itself has multiple benefits which impact the clinical utility of Monte Carlo treatment planning, an interesting and novel application would be to investigate virtual energies (Hogstrom and Almond 2006). By fully parameterizing the phase space files with continuous functions, interpolations of the probability distributions between energies (e.g. 12 and 16 MeV could give 14 MeV) may allow for investigation of optimal energies for DEAR and generally for optimal energy modulation in electron beam therapy.

3.4.5. Conclusion

In conclusion, Varian TrueBeam electron beam phase space files can be parameterized by distributions for particle type, position, energy, and direction. The parameterization can generate accurate pseudo phase space files that are equivalent to the original phase space files for a range of energies and field sizes and future work will refine the parameterization for robustness. The benefits of parameterization include: Occupation of less disk space (~1 MB) as well as generation of as many particles as the user needs and reduced statistical uncertainties.
4. Design of new hardware for improved DEAR capability: Multiple apertures in a single cut-out

4.1. Introduction

In the current clinical workflow for electron beam therapy, a fixed patient specific cut-out must be manufactured such that the beam will laterally conform to the target and shield normal tissue. While effective, it is also labor-intensive and inefficient as it requires the physicist or therapist to prepare a patient-specific cut-out using a positive mold method: First, Styrofoam must be cut to a specified shape, and then placed in an applicator ring, where Cerrobend is poured. The Cerrobend takes up to 24 hours to harden, and must still be checked for consistency. This procedure must be followed for each treatment field, making it an inefficient technique. Further, some target shapes require the use of multiple patient-specific cut-outs, thus the therapist must enter the room multiple times to change cut-outs during treatment e.g. a chest wall scar boost might require the use of two static abutting beams, due to the curvature of the patient and the extent of the target. In the DEAR feasibility study, a 1x10 or 3x10 cm² aperture in a cut-out was utilized. While this worked for a simple geometry, more complex shapes may require, variable cut-out sizes.

A patient-independent tertiary collimation system in conjunction with dynamic electron beam therapy is thus more desirable, as it would reduce cut-out fabrication
time, and improve efficiency. Additionally, higher resolution fluence modulation over the target, which is not feasible with a single patient specific cut-out, is also desirable.

Previous developments in dynamic electron beam therapies, such as MERT, propose to achieve the lateral dose constriction and fluence modulation by using a few leaf collimator (FLEC) or an electron multi-leaf collimator (eMLC). The use of the photon MLC as a tertiary collimation device has been previously investigated (Klein et al 2009b). However, for most linacs, intensity modulation using the photon MLC without modification of the treatment head does not produce useful beam penumbra because of the large distance between the downstream edge of the MLC to the isocenter and the large extent of the virtual source due to the dual scattering foil (Ma et al 2000). The large thickness of the photon MLC and the rounded edge design also contributes significantly to the scatter downstream and useful intensity modulation can only be achieved when the SSD is much reduced. However, that is not desirable for many clinical situations. Extensive MC simulations, design, and implementation studies for eMLCs and FLECs for MERT have been performed to investigate its characteristics (Lee et al 2000, Ma et al 2000, Lee et al 2001b, Deng et al 2002, Ma et al 2003, Al-Yahya et al 2005, Al-Yahya et al 2007, Vatanen et al 2009, Alexander et al 2011). More recent studies have reported on prototype eMLCs and tested their feasibilities (Eldib et al 2013, Jin et al 2014). The use of
intensity and energy modulation is beneficial for conforming the beam to complex target shapes and reducing scattered dose and dose to normal tissue.

While FLECs and eMLCs are a promising technique for achieving these goals, the manufacturing and implementation of eMLCs or FLECs has not become commercially available and has in turn impeded the wide-spread clinical implementation of MERT. Further, Cerrobend block cutting rooms are still widespread in the clinic and cut-outs are also available through companies that produce electron cut-outs. Thus the aim of this study was to explore possible improvements that could be implemented with readily available resources. Multiple small apertures in a single cut-out with variable jaw sizes were investigated to see if they could be utilized in a single dynamic delivery. In the following simulation study, the dosimetric characteristics of such an arrangement were investigated to understand what cut-out design would be optimal.

4.2. Materials and Methods

Figure 70 describes the idea behind using multiple apertures in a single cut-out. In Figure 70 (a) a single aperture cut-out provides the lateral beam constriction. While a single aperture is effective for a target on a flat surface and a limited target extent, curved and large targets require the use of multiple cut-outs. This adds additional treatment time and degradation of the dose distribution at the field abutment. In Figure
the eMLC is similar to the photon MLC, in that it has two banks of leaves composed of a high Z material such as lead or tungsten. Not only can an eMLC provide lateral beam constriction, but as depicted with the dashed and solid leaves, fluence modulation can be achieved by leaf sequencing. In Figure 70(c) multiple apertures in a single cut-out could achieve lateral beam constriction and fluence modulation limited to the size of the smallest aperture. Multiple apertures could be used in DEAR to dynamically deliver small fields along a scanning beam trajectory: The larger apertures could be used to cover the interior of the target, while the smallest apertures could be used on the edges to improve penumbra. This represents a trade-off in delivery efficiency and desired dose distribution that was discussed in section 3.3.

Figure 70. Schematic describing approaches to provide lateral beam construction and fluence modulation. (a) Single patient-specific cut-out, (b) eMLC, and (c) multiple apertures in a single cut-out e.g. 1x1, 2x2, 3x3, and 4x4 cm². In this setup, the 4x4 cm² aperture is open, with the other apertures shielded by the Y and X jaws.
Jaw settings for standard field sizes are set larger than the field sizes to minimize extra-focal scatter stemming from the secondary collimators, thus providing a large, flat field i.e. a large electron field with equal fluence. For small field sizes (defined here as a field size smaller than 6x6 cm²) such as 1x1, 2x2, 3x3, 4x4, and 5x5 cm², these jaw settings may be superfluous and smaller jaw sizes might still yield acceptable dosimetry, similar to the clinical standard jaw size. This would open up the possibility of placing multiple apertures on a single cut-out and using the jaws to control which aperture is irradiated, while blocking the electron beam from entering the unused aperture openings. The beam will not be fully blocked, as electron beam tend to scatter at large angles. Thus scattered dose to the shielded apertures must be assessed, to see if it within a tolerable limit.

This study was performed as a MC simulation study in the EGSnrc environment and the BEAMnrc and DOSXYnrc user codes utilizing Varian TrueBeam phase space files which had been validated in section 3.2. The linac head components were modeled using the PYRAMID, APPLICAT, and BLOCK CM for the jaws, applicator, and cut-out with single or multiple apertures. Unlike in the previous study, the BLOCK CM, is more flexible than the PYRAMIDS CM as it allows for definition of multiple apertures. All MC simulations resulted in a statistical uncertainty of <1.5% at depth of maximum dose d_{max} for a cuboid water phantom with a dose scoring resolution of 0.2x0.2x0.2 cm³ in the x, y,
and z dimension. Investigated quantities included energy, jaw size, applicator/cut-out size, number of apertures, aperture size, and aperture placement and are summarized in Table 14.

Table 14. Investigated quantities to assess the dosimetry of multiple apertures on a single cut-out.

<p>| | |</p>
<table>
<thead>
<tr>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Energy (MeV)</strong></td>
<td>6, 16</td>
</tr>
<tr>
<td><strong>Jaw size (cm²)</strong></td>
<td>Standard: <strong>18x18 (16 MeV), 22x22 (6 MeV)</strong></td>
</tr>
<tr>
<td></td>
<td>Reduced: 1x1, 2x2, 3x3, 4x4, 5x5, 7x7, 8x8, 10x10, 15x15</td>
</tr>
<tr>
<td><strong>Applicator size (cm²)</strong></td>
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<tr>
<td><strong># of apertures</strong></td>
<td>single, multiple</td>
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<tr>
<td><strong>Aperture size (cm²)</strong></td>
<td>1x1, 2x2, 3x3, 4x4, 5x5</td>
</tr>
<tr>
<td><strong>Aperture placement</strong></td>
<td>On-axis (1 position), Off-axis (3 positions)</td>
</tr>
</tbody>
</table>

For the electron beam energy both a low (6 MeV) and high (16 MeV) energy were investigated for a 15x15 cm² applicator/cut-out combination. A 15x15 cm² applicator was selected as it is a commonly used applicator size that will permit the placement of multiple apertures as well as allow for collision avoidance during delivery. The standard jaw setting for this applicator is 22x22 and 18x18 cm² for the 6 and 16 MeV, respectively. Jaw sizes from the standard down to the smallest aperture size were investigated. Both single and multiple apertures on a cut-out were studied, with apertures sizes ranging from 1x1 to 5x5 cm². Apertures were placed both on and off-axis.
Three configurations were assessed and are illustrated in Figure 71:

(a) The standard configuration of a single aperture on-axis (x and y coordinates 0 and 0 cm) on a cut-out with standard jaw setting (blue dashed line). Additionally the jaw sizes were reduced (black dashed line) symmetrically around the apertures. All aperture sizes were investigated.

(b) A single aperture on a cut-out placed off-axis with both standard and reduced jaw setting symmetric to the aperture opening. Three different aperture placements were investigated. A 15x15 cm² cut-out has the dimensions of 14.25x14.25 cm² at 95 cm (the bottom of the cut-out). The three aperture placements off-axis (positions 1, 2, and 3) are thus defined at the following coordinate locations: (1) 1.76 cm and 1.76 cm, (2) 3.52 cm and 3.52 cm, (3) 5.28 and 5.28 cm, and is shown graphically in Figure 72. Configuration (b) was only tested in quadrant I. All aperture sizes were investigated.

(c) Multiple apertures on a single cut-out with reduced jaw settings centered on aperture. Aperture sizes 1x1, 2x2, 3x3, and 4x4 cm² were spaced optimally, i.e. edges of each aperture were placed furthest away from each other, as shown in Figure 71 (c). The ordering of the apertures was chosen as follows 4x4, 1x1, 3x3, and 2x2 cm² for quadrant I – IV, to maximize the distances between apertures. Off-axis placement using positions (2) and (3) were investigated as well.
Figure 71. Configuration for the single aperture (a) on- and (b) off-axis, and (c) multiple apertures on a single cut-out. The dashed lines represents the jaw size (blue: standard, black: reduced).

Figure 72. For off-axis aperture placement, three different positions were tested (1) \((x,y) = 1.76 \text{ cm}, 1.76 \text{ cm}\), (2) \((x,y) = 3.52 \text{ cm}, 3.52 \text{ cm}\), and (3) \((x,y) = 5.28 \text{ cm}, 5.28 \text{ cm}\) defined on the cut-out at an SSD of 95 cm.

Reference was configuration (a) with standard jaw size. Comparison metrics included depth dose and orthogonal profiles at depths \(R_{100}\), \(R_{50}\), and \(R_p\) defined for a standard 15x15 cm\(^2\) cone according to Table 5. For the PDDs, dose differences in the
build-up and Bremsstrahlung region were assessed, while in the fall-off region range difference at $R_{50}$ were assessed. The gamma index with a criteria of dose difference of 2% and distance to agreement of 1 mm was used. The output factors are measured as defined in section 3.2. For configuration (c), optimal aperture placement and jaw size were chosen such that reference dosimetry was maintained and leakage through unused apertures was <5%. The leakage dose was defined as the dose at the center of the unirradiated aperture at the depth $R_{100}$ of the irradiated aperture.

4.3. Results

4.3.1. Single aperture on-axis

Small field dosimetry for on-axis aperture placement (Figure 71 (a)) was characterized as a function of symmetric and centered jaw sizes starting from the standard jaw setting down to the respective aperture setting. The PDD, profiles at multiple depths, and output factor for the smallest (1x1 cm$^2$) and largest (5x5 cm$^2$) aperture are shown in Figures 73 and 74 for both the 6 and 16 MeV electron beam energy.
Figure 73. 6 MeV Profiles and PDDs for a 1x1 cm² (top) and 5x5 cm² (bottom) on-axis aperture as a function of jaw size from 22x22 cm² to the cut-out size.
Figure 74. 16 MeV Profiles and PDDs for a 1x1 cm\(^2\) (top) and 5x5 cm\(^2\) (bottom) on-axis aperture as a function of jaw size from the standard 18x18 cm\(^2\) to reduced jaw sizes.

For a given aperture, as the jaw sizes decrease, the field size decreases i.e. the flat region becomes smaller, and the penumbra becomes narrower. The cross-plane and in-plane profiles remain similar within a gamma criteria of 2%/1mm (i.e. 100% pass rate) for a range of reduced jaw sizes. Since only a handful of jaw setting were tested, Table 15 shows the jaw sizes for a given aperture and energy combination for which the gamma
criteria is met. For most aperture and energy combinations, the acceptable jaw size is
equal to or slightly larger than the aperture size itself. The largest discrepancies in the
gamma index are seen in the outside field dose i.e. past the penumbra region, where the
reduced field sizes actually results in a lower out-field dose.

Table 15. Jaw sizes (cm$^2$) which give a cross-plane or in-plane profile within a gamma
criteria of 2% and 1 mm when compared to the standard jaw size for all apertures and
ergy combinations.

<table>
<thead>
<tr>
<th>Energy (MeV)</th>
<th>Aperture size (cm$^2$)</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>1x1</td>
</tr>
<tr>
<td>6</td>
<td>2x2</td>
</tr>
<tr>
<td>16</td>
<td>2x2</td>
</tr>
</tbody>
</table>

As shown previously in section 3.3., for the same energy and varying cut-out
sizes, as the aperture decreases, the PDD shifts towards the surface, giving a shallower
depth of maximum dose $R_{100}$, while $R_p$ relatively maintains its depth. For a given
aperture size, the changes observed for the PDDs as a function of jaw size can be
described in three different regions: the build-up or surface dose, the fall-off, and
Bremsstrahlung region.

In the build-up region, the surface dose decreases as a function of jaw size: For
the apertures sizes >3x3 cm$^2$ the changes in surface dose from the standard jaw size to
the jaw size of the aperture is less than 1.5% for both energies. For 1x1 and 2x2 cm$^2$, surface doses can drop up to 7.7% (6 MeV) and 4.1% (16 MeV). Figure 75 shows the surface dose for all apertures for an energy of 6 MeV and various jaw sizes, as that energy displayed the largest differences for the surface dose.

Figure 75. PDDs in the build-up region for all apertures for a 6 MeV electron beam and reduced jaw sizes. As the aperture decreases the depth of maximum dose shifts towards the surface, shifting away from the a depth of 1.4 cm for a standard 15x15 cm$^2$ field size, to less than 0.5 cm for a 1x1 cm$^2$ aperture. For a given aperture, as the jaw size decreases the surface dose decreases. This effect is most prominent for the 1x1 cm$^2$ aperture.
In the fall-off region, it was observed that $R_{50}$ tended to shift progressively deeper as the jaw size decreases for a given aperture. Shifts in $R_{50}$ were within 1 mm for apertures 3x3, 4x4, and 5x5 cm$^2$, all jaw sizes, and both energies. For apertures 1x1 and 2x2 cm$^2$, shifts of up to 2 mm are observed for the 16 MeV 1x1 cm$^2$ aperture and 4x4 cm$^2$ jaw size.

In the region beyond $R_p$, reducing the jaw size leads to a higher bremsstrahlung tail for all apertures and both energies. The absolute differences for the 16 MeV and all apertures and jaw sizes are within 1%. Differences for 6 MeV and apertures 3x3, 4x4, and 5x5 cm$^2$ were within 3% for all jaw sizes, while apertures 1x1 and 2x2 cm$^2$ displayed differences of up to 9.5% and 7%, for the smallest jaw sizes 1x1 and 2x2 cm$^2$, respectively.

As for the gamma criteria of 2% and 1 mm, the jaw sizes for which the PDDs agree to the standard jaw size are the same as shown in Table 15. Thus, PDDs and profiles remain similar (within gamma index criteria of 2%/1mm) to the standard dosimetry (defined by the standard jaw setting and the respective aperture) for a range of reduced jaw sizes. For jaw sizes smaller than those shown in Table 15, a reduced surface dose, an increased Bremsstrahlung tail, and reduction in OF, especially for the 1x1 cm$^2$ aperture and 6 MeV electron beam energy case, were observed.
4.3.2. Single aperture off-axis

Off-axis placement of the apertures (Figure 71 (b)) was investigated as a function of the location on the cut-out along three positions along the diagonal. Simulations were performed by using standard and reduced jaw sizes symmetric about the center of the aperture. Figure 76 shows the results of a 6 Mev 4x4 cm\(^2\) irradiation at three off-axis positions for the standard jaw size of 22x22 cm\(^2\). For larger jaw sizes the PDD and profiles remain comparable to those on axis (differences within 2%/1mm). As the jaw size decreases to 5x5 cm\(^2\), larger apertures such as the 4x4 cm\(^2\) begin to show some asymmetries in the profiles, while the smaller apertures such as 1x1 cm\(^2\) remain comparable to those on-axis.
Figure 76. 6 MeV PDD for a 4x4 cm² for 3 off axis positions and standard jaw size (22x22 cm²). In comparison to the on-axis PDD, the dosimetry is comparable for all positions. Further, because the Bremsstrahlung contamination, which is mainly generated in the scattering foil, is forward directed, the off-axis dosimetry displays a reduction in the Bremsstrahlung tail by 0.5%.

4.3.3. Multiple apertures off-axis

Optimal jaw size was targeted with the goal of balancing leakage in the shielded apertures and similar dosimetry as the on-axis single aperture. The apertures were
placed equidistant as shown in the configuration in Figure 71 (c) and at two off-axis positions were tested. From the previous study in section 4.3.2, position 2 and 3 (3.52/3.52 and 5.28/5.28 cm) were chosen. Symmetric and centered jaw sizes from 10x10 to the size of the aperture were simulated. The placement of the apertures on the cut-out and jaw size influenced the leakage in the shielded apertures, as well as the dosimetric characteristics in the irradiated apertures.

To illustrate the effect of both aperture placement and jaw sizes on the resulting dosimetry, Figures 77 and 78 show 2D dose distributions at the depth of maximum dose for a 6 and 16 MeV beam for 1x1 and 4x4 cm² irradiations for both aperture placements Position 2 and Position 3 and the largest jaw size of 10x10 cm² and smallest jaw size of either 1x1 or 4x4 cm².
Table 1: 2D dose distributions at depth of maximum dose for a 1x1 cm$^2$ irradiation for electron energies of 6 MeV and 16 MeV for two different aperture placements (Position 2 and Position 3) and two different jaw settings (10x10 and 1x1 or 4x4 cm$^2$, respectively).

<table>
<thead>
<tr>
<th></th>
<th>Position 2</th>
<th>Position 3</th>
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<tbody>
<tr>
<td><strong>Jaw Size</strong></td>
<td></td>
<td></td>
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<tr>
<td>10x10 cm$^2$</td>
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<td><img src="image2.png" alt="Image" /></td>
</tr>
<tr>
<td>1x1 cm$^2$</td>
<td><img src="image3.png" alt="Image" /></td>
<td><img src="image4.png" alt="Image" /></td>
</tr>
<tr>
<td><strong>Energy</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>6 MeV</td>
<td><img src="image5.png" alt="Image" /></td>
<td><img src="image6.png" alt="Image" /></td>
</tr>
<tr>
<td>16 MeV</td>
<td><img src="image7.png" alt="Image" /></td>
<td><img src="image8.png" alt="Image" /></td>
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</table>

Figure 77. 2D dose distributions at depth of maximum dose for a 1x1 cm$^2$ irradiation for electron energies of 6 MeV and 16 MeV for two different aperture placements (Position 2 and Position 3) and two different jaw settings (10x10 and 1x1 or 4x4 cm$^2$, respectively).
Figure 78. 2D dose distributions at depth of maximum dose for a 4x4 cm$^2$ irradiation for electron energies of 6 MeV and 16 MeV for two different aperture placements (Position 2 and Position 3) and two different jaw settings (10x10 and 1x1 or 4x4 cm$^2$, respectively).

The following was observed for all energies, irradiated apertures, jaw sizes, and aperture placements: For a given aperture placement and irradiation, reducing the jaw size reduces the leakage in the shielded apertures. Further, for a given aperture irradiation and jaw size, increasing the distance between apertures reduced the leakage in the shielded apertures. In the example shown in Figure 77 and 78, the leakage dose to the shielded apertures 1x1, 2x2, and 3x3 cm$^2$ can be reduced to under 5% when shifting...
the aperture placement from Position 2 to Position 3. Symmetry in the orthogonal profiles can be observed in the 2D dose distributions: The smaller aperture sizes such as 1x1 and 2x2 cm$^2$ are less sensitive to degradation from the reduced jaw sizes, as they have a very small flat region. For larger aperture size of 3x3 and 4x4 cm$^2$, however, the degradation of the orthogonal profile’s symmetry due to aperture placement further off-axis and reduced jaw size is more substantial.

Figure 79 shows the PDDs through the center of all four apertures for a 6 MeV 1x1 cm$^2$ aperture irradiation for reduced jaw sizes. The other apertures (2x2, 3x3, and 4x4 cm$^2$) are thus shielded by the jaw sizes. Even though the “unwanted” apertures are shielded geometrically by the jaws, because of the nature of electron interaction with matter, leakage radiation is still present through the unirradiated apertures. Aperture placement substantially reduced leakage dose: For the largest jaw size of 10x10 cm$^2$, the leakage dose decreases from 85% to 25% (3x3 cm$^2$) from Position 2 to 3.
Figure 79. PDDs of all four apertures for the 6 MeV show a 1x1 cm² irradiation for 2 different apertures placements ‘Position 2’ and ‘Position 3’. PDDs are normalized to the maximum dose in the phantom. The black dashed line represents the 5% dose level. Placing the apertures further apart reduced leakage substantially for the same jaw size. A reduction of the bremsstrahlung tail was also observed for increasing distance between apertures.
For a given applicator size of 15x15 cm$^2$ and two energies 6 and 16 MeV, aperture placement in Position 3 and reduced jaw sizes listed in Table 16 yielded the best results when balancing for comparable dose distribution in the irradiated aperture and reducing leakage in adjacent shielded apertures to within mostly 5%.

Table 16. Optimal jaw sizes that balance comparable dose distributions and leakage dose for 6 and 16 MeV for all apertures.

<table>
<thead>
<tr>
<th>Energy (MeV)</th>
<th>Aperture size (cm$^2$)</th>
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<tr>
<td></td>
<td>1x1</td>
</tr>
<tr>
<td>16</td>
<td>7x7</td>
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Smaller aperture sizes 1x1 and 2x2 cm$^2$ were found to be more robust to asymmetric profiles due to off-axis aperture placement, than the larger 3x3 and 4x4 cm$^2$ apertures sizes. The higher 16 MeV energy led to a lower ratio of leakage radiation in adjacent apertures than the 6 MeV energy, thus larger field sizes can be used, which also beneficial in preserving the output factor.
4.4. Discussion

Comparable dosimetry can be achieved with multiple off-axis apertures (1x1, 2x2, 3x3, and 4x4 cm²) on a single cut-out with reduced jaw sizes when compared to reference on-axis dosimetry. An interesting effect can be seen in Figure 77 for the 1x1 cm² irradiation: It was observed that the 3x3 cm² had a higher leakage dose than the 4x4 cm², even though the edge of the 4x4 cm² is closer to the 1x1 cm² than the 3x3 cm². It was hypothesized that the scattering contributions from the Y and X jaws were not symmetric as the jaws do not reside on the same z plane, rather the Y jaw is further upstream than the X jaw. This was tested by two additional simulations and comparing to the original configuration (Figure 80): (1) By flipping the location of the 3x3 and 4x4 cm² apertures and (2) by replacing the 4x4 cm² with a 3x3 cm² apertures for a 1x1 cm² irradiation with a 5x5 cm² jaw size.

Figure 80. 2D Dose distributions normalized to the depth of maximum dose for a 6 MeV 1x1 cm² irradiation with a 5x5 cm² jaw size for the original (left), ‘flipped’ (middle), and the ‘replaced’ (right) configuration. The apertures 4x4, 1x1, 3x3, and 2x2 cm² are in the quadrants I, II, III, and IV, respectively.
Flipping the locations of the 3x3 and 4x4 cm² or replacing the 4x4 with the 3x3 cm² did not change the fact that the leakage dose was higher in quadrant III than I, even for the same aperture size. Study of the asymmetric jaw sizes may provide more insight to the effect of the jaw sizes on the leakage dose.

Future work should also include investigation for all energies in between 6 and 16 MeV e.g. 9 and 12 MeV, although, it is expected that these results will lie in between the stated results for 6 and 16 MeV. Optimizing aperture placement should also be investigated for other aperture shapes (e.g. circular) and sizes, aperture placements with different distances, and applicator sizes.

4.5. Conclusion

While prior publications have shown both the feasibility and advantages of FLEC or eMLC over patient-specific cut-outs, clinical implementation has been slow. This study has shown that multiple apertures in a single cut-out with variable jaw size can be used in a single dynamic electron beam delivery, thus providing a practical alternative to FLEC or eMLC. This technique could offer a low-cost solution in providing lateral beam conformity and fluence modulation for dynamic electron beam therapies such as DEAR.
5. **Summary and future research**

Developmental work for dynamic electron arc radiotherapy (DEAR) and tools and techniques to improve current clinical and research electron beam therapy were presented. The scope of this doctoral research project was to develop foundational knowledge for dynamic electron arc radiotherapy (DEAR), a novel electron therapy technique. An initial feasibility study investigated aspects of planning, delivery, and verification on cylindrical phantoms. The study showed that DEAR can produce homogenous dose distributions over large and curved surfaces while maintaining narrow penumbra. Subsequently, an electron Monte Carlo model for the Varian TrueBeam linac was validated. Comprehensive small field dosimetry as the foundation for the treatment planning for this novel conformal electron therapy technique was evaluated and virtually scanning mode employing small fields was assessed. Additionally, an accurate parameterization of the phase space file was investigated as a means to improve Monte Carlo simulation efficiency, occupation of less disk space, as well as generation of as many particles as the user requires. To improve efficiency and effectiveness of DEAR delivery, multiple apertures on a single cut-out were investigated as a technique that can offer a low-cost solution to providing lateral beam conformity and fluence modulation for DEAR.
Future research plans and long-term goals for the planning, delivery, and verification of DEAR are presented in the following sections.

5.1. Planning

As previously stated, current electron dose algorithms are not accurate enough nor can they handle dynamic planning for DEAR. With the Monte Carlo framework, dose calculations with high accuracy can be achieved.

5.1.1. DEAR dose calculation algorithm

Current implementation for the DEAR dose calculation algorithm has been shown for simple geometries and planning trajectories. The flowchart in Figure 81 depicts the full realization of the DEAR dose calculation algorithm.
Starting from the patient CT data, contours of all target (PTV) and normal structures are obtained. Combined with the planning objectives, this guides on one hand the aperture selection and energy selection and on the other hand the delivery parameters. Once the aperture and energy have been selected, the appropriate pre-calculated MC dose kernel in conjunction with the selected trajectory can be used for the subsequent beam weight optimization and the dose for the dynamic DEAR plan can then be calculated according to the superposition algorithm discussed in section 2.2.1.
Here “selection” can be supplanted by “optimization” once appropriate algorithms are implemented. Plan verification can be performed with a full MC simulations: To this extent, the DOSXYZnrc user code sources 20 and 21 could be used. These sources allow for the delivery of dynamic treatments to be simulated as a function of MU (Lobo and Popescu 2010). This would fully and accurately simulate a DEAR delivery.

5.1.2. Potential and limitations

5.1.2.1. Aperture selection

There are both benefits and drawbacks to using a smaller cut-out for coverage of a larger region: A 1x10 cm$^2$ leads to a narrower penumbra than the 3x10 cm$^2$ cut-out for the same target coverage. However, a smaller cut-out also translates to a decrease in output. This means that a larger number of MU must be delivered for the same target dose and area, which in turn reduces delivery efficiency. There is also potential for increased leakage dose to the normal tissues. This suggests that a narrow cut-out may only be needed at the edge of the delivery to sharpen the penumbra and can be avoided in the middle target region.

The advantage and limitations of using multiple apertures to cover a single target was shown in section 3.3. While dedicated aperture and fluence modulation hardware such as an eMLC or FLEC is desirable for delivery, a cost-effective technique
using readily available Cerrobend cut-outs with multiple apertures in conjunction with
dynamic jaws could be used to deliver the small fields, as was shown in section 4.

5.1.2.2. Energy selection

As discussed with MERT, energy modulation can provide conformity along the
distal part of the PTV. The selected energy’s $R_{80}$ should encompass the distal part of the
target. DEAR will benefit from energy modulation, and as such, investigation of optimal
ergies for DEAR are warranted. Beyond energy selection, optimal mixing of multiple
ergies with different fluence weighting can result in limited depth modulation such as
generating a range between two energies or increasing the build-up region to provide a
higher dose to the surface, depending on the requirements of the target volume
(Olofsson et al 2004). This is shown with the examples in Figure 82. Due to the design of
the linear accelerator, only a few discrete energies are available, thus mixture of energies
with variable weighting can produce variation in depth. A limitation of this technique is
that certain mixed energies lead to a gradient in the fall-off region that becomes
shallower as shown in the example in Figure 82 (b).
Figure 82. Effect of mixing energies on resulting PDDs. PDDs are shown for Monte Carlo simulations on Varian TrueBeam linacs for a 15x15 cm\(^2\) field size. (a) A 6 MeV beam with a fluence weighting of 0.05 is added to a 12 MeV beam with a fluence weighting of 1. The resulting PDD increases the dose in the surface and build-up region, while maintaining the same fall-off region slope as the 12 MeV beam. (b) A 12 MeV beam with a fluence weighting of 0.5 is added to a 16 MeV beam with a fluence weighting of 1. The resulting range for the PDD is between that of the 12 and 16 MeV. A limitation of this technique is that the gradient in the fall-off region may become shallower.

5.1.2.3. Trajectory selection

Optimization of the trajectory to cover the target region must be considered in terms of dosimetry and delivery. As shown in section 3.3.3.2., overlap/gap must be properly selected based on the field size used to maintain a homogenous dose distribution. Concurrently, delivery optimization in terms of delivery efficiency and trajectory possibility space, further constraint trajectory selection. This will be discussed in more detail in section 5.3.
5.1.3. Preliminary planning study

A standard tangential photon and DEAR plan were generated in the treatment planning system for a phantom simulating a post-mastectomy chest wall irradiation. For the photon plan, a pair of opposing tangent 6 MV photon beams were used. For the electron plan, a combination of 6, 9, and 12 MeV electron beams along the arc were utilized and the beam weights were manually optimized. Resulting dose distributions for the DEAR and tangential photon plan are shown in Figure 83.

Figure 83. Axial slice of the simulated mastectomy chest wall irradiation with a DEAR plane (left) and tangential photon beam plane (right). The yellow contour represents the 100% isodose line.
Both plans exhibit similar target coverage, however, for the tangential photon plan, the ipsilateral lung receives between 30 to 80% dose, while in the DEAR plan the lung dose is limited to below 30%.

While the lung dose in DEAR plan was within reasonable limits for this phantom case, it may be higher for actual patient with different chest wall thickness. This is due to the fact that adequate target coverage needs to be maintained to deeper depths with the concurrent requirement of adequate skin dose. This situation is currently challenging for DEAR as planning is limited by the discrete electron beam energies available on linacs, and addition of bolus will make the planning more complex.

**5.1.4. Planning for specific treatment sites**

Once the planning framework outlined in section 5.1.1. is fully established, investigation of DEAR for potential treatment sites will be explored. New radiation therapy techniques aim to improve the therapeutic ratio by either dose escalation to improve local tumor control to by reducing dose to normal tissue. Planning studies for sites such as chest wall, breast, scalp, and extremities will compare dosimetric results to that from standard clinical treatment techniques. Measures of normal tissue toxicities can then be modeled using the normal tissue complication probability (NTCP) model, which is based on normal tissue DVHs. While dosimetric advantages are to be expected with DEAR, many current techniques, especially for breast and chest wall may already
have an acceptable therapeutic ratio. However, these reasons should not impede research in DEAR as therapeutic ratio is one among several important factors to consider when determining the clinical value of a therapy technique. Additionally, it represents a novel treatment modality that seeks to improve and streamline in a cost-effective manner electron beam therapy treatment and is thus a worthy endeavor. As with any new treatment technique its clinical effectiveness should be assessed to maximize its clinical utility (van Loon et al 2012).

Potential patient subgroups that may benefit from DEAR include chest wall recurrence, where the lifetime tissue radiation tolerance has been met. DEAR may still produce optimal target coverage for these cases. As breast-conserving treatment gains popularity, improving quality of life and breast cosmesis may add additional demands to radiation therapy techniques. DEAR may be able to produce equivalent target coverage with a reduction in normal tissue complications. Further, dose escalation with DEAR may be viable for specific types of cancer, which due to their biology are more radiation resistant, and thus require a higher dose to reduce tumor cell survival.

DEAR would benefit from the use of more energies than are currently available on linear accelerators. Even with energy modulation, the discrete energies available may not result in optimal penetration depths and surface coverage. Challenging situations such as a thick target area need a high electron energy for adequate coverage. This may
result in unacceptable lung and heart dose. Bolus may be used to shift the PDD upstream, thus reducing the dose to the lung. However, the additional bolus introduces more complexity to the plan. Investigation of optimal energies, especially of lower energies, is therefore of interest to DEAR.

5.2. Verification

In the following sections 5.2.1. and 5.2.2., dosimetric and delivery verification of the DEAR plan are discussed in terms of pre-treatment verification. Additional discussion on actual delivery accuracy are discussed in section 5.3.

5.2.1. Dosimetric verification

For the work presented in this doctoral research, 1D (ionization chambers and silicon diodes) and 2D (radiographic film) dosimetry was performed. Ion chambers and are seen as the ‘gold standard’ for dosimetry and enjoy excellent dosimetric characteristics such as stable operation over time, linear response to dose, dose rate, and energy independence (Khan et al 1991a). Silicon diodes are advantageous because of their small size, high sensitivity, and thus high spatial resolution. Further, the stopping power ratio for silicon diodes to water is essentially constant over the range of energies encountered in radiation therapy (Gerbi et al 2009). As such diodes measure dose directly and do not require conversion from ionization to dose as is required by ionization chambers. Measurements acquired with diodes must be compared to ion
chamber measurements to ensure proper operation (Song et al 2006). Radiographic film provides high-resolution 2D dosimetry, and generally enjoys good dosimetric properties, if properly calibrated. However, film is tedious to use. Variations in overall film response depend on the film batch, processing conditions, and densitometer. Instability in the processing conditions alone can lead to variability in final dose distribution of up to ±30%, but can be reduced if corrections are used. For these reasons, investigation of additional 2D and 3D dosimetry methods is warranted.

2D dosimeters such as the electronic portal imaging device (EPID) or computed radiography (CR) systems may potentially be used for DEAR dosimetry. EPIDs consist of a phosphor layer atop of an array of aSi photodiodes and works on the principle of indirect conversion, which is optimal for detection of photons. As such, EPIDs are commonly employed for IMRT and VMAT treatment verification QA, due to their high spatial resolution and linear dose and dose rate response. Recent publications have shown the possibility of using EPIDs for electron beam quality control (Beck et al 2009) and verification of static treatments (Jarry and Verhaegen 2005, Chatelain et al 2013) while maintaining the aforementioned advantageous EPID dosimetric properties. CR systems, which also work on the principle of indirect conversion, have been investigated for energy consistency checks (Cai et al 2009), but research on verification of electron beam treatments has been limited.
For 3D dosimetry, PRESAGE™, a dosimetry system consisting of solid polyurethane doped with radiochromic leucodye has shown potential for accurate electron dosimetry (Guo et al 2006). PRESAGE™ displays linear response to dose and independence to dose rate, demonstrating itself as a good candidate for stable and sensitive 3D dosimetry for electron beams. Further, the dosimeter can be manufactured in any shape, which is appealing for DEAR applications. Recently, radiochromic film stacks have been used to create 3D dosimetry and have been shown to accurately characterize photon beams (McCaw et al 2014) presenting another, albeit involved, option for electron beam verification. Continued research in 3D dosimetry will not only affect development in DEAR, but in all of radiation therapy.

5.2.1.1. Preliminary investigation of CR-based electron beam dosimetry

In our clinic, CR is primarily used as a qualitative localization and verification technique for patient positioning in Total Body Irradiation therapy. However, as showed in a study by Olch, CR systems can be successfully implemented to obtain absolute photon beam dosimetry (Olch 2005). The same has not been extensively evaluated for electron beams.

In comparison to radiographic film, CR systems enjoy numerous benefits including non-chemical development of the images, an image quality that does not depend on processing conditions (but may depend on other variables such as time
between irradiation and scanning), and provides an immediate digital storage of the images. Further, CR plates are “film-like” in that the plates are flexible (if removed from the rigid cassette), can be placed inside a phantom – or more pertinently for the case of DEAR – wrapped around a cylinder (unlike the EPID) and thus can be independent from the linac orientation.

Multiple aspects of CR-based electron beam dosimetry need to be investigated to make sure it is a feasible option. Moreover, these aspects would need to be investigated for any novel electron dosimetry system. This includes: (1) irradiation without a cassette (As mentioned, CR plates are usually irradiated with a cassette, which sandwiches the plate between two phosphor-coated intensifying screens to improve the signal efficiency), uniformity across homogeneously irradiated CR plates, (2) dose rate dependence, (3) energy dependence, (4) field size dependence, (5) angular dependence of incident beam (e.g. parallel or perpendicular orientation, for profile or PDD measurements), and (6) saturation dose. Stability and reproducibility of the CR system such as blank scans, scanner uniformity, and inter-batch variation should also be assessed.

Testing of a single Agfa MD10 Imaging plate consisting of two layers (active layer: BaSrFBr:Eu^{2+} on flexible plastic layer), a Kodak EC-L lightweight cassette (regular), and a Kodak ACR 2000i scanner, yielded preliminary answers to aspects (1) – (4) and (6)
as well as some aspects of the CR system itself. Irradiation without a cassette was possible, and the CR plate was placed in a light-tight paper envelope instead, thus protecting the plate from ambient room light. All subsequent measurements were made inside a plastic water equivalent phantom with the film parallel to the beam. The CR plate was irradiated with varying dose of 1-250 MU, variable dose rate (400 and 1000 MU/min), electron energies 6 and 9 MeV, and field sizes (6x6 and 15x15 cm$^2$). The CR plate with the cassette was placed on the table with the SSD set to 100 cm. The naked CR plate was placed at depth of maximum dose for the respective energy in the phantom with the SSD set to 100 cm. Scanner bed variability and uniformity was checked with a blank scan on multiple days. While stable, the blank scan displayed a lower response at the edge of the scanner bed, and which was subsequently used as a non-uniformity correction (Figure 84).
Figure 84. Scanner blank scan response. Blank scan for the entire 14”x17” CR plate (left). A non-uniformity across the x-direction is visible, such that a non-uniformity correction using a polynomial fit can be derived as used to correct for scanner non-uniformities (right).

The amount of luminescence emitted by the CR plate and recorded as arbitrary scanner units was found to have a logarithmic response to dose over a range of MUs from 1 to 250 MU with $R^2 > 0.99$ (Figure 85). Energy dependence was found to be within ±2 MU over the range of 0 – 250 MU for the 6 and 9 MeV energy. Dose rate dependence between a 400 and 100 MU/min irradiation was found to result in a consistently lower response for the higher dose rate by about 2% in scanner units for dose irradiation of 10 to 100 MU and a 6 MeV beam. Field size dependence for a 6 MeV beam and a 6x6 and 15x15 cm² field size yielded differences within 1.5% in scanner units. Irradiation of the CR plate with or without the cassette, did not yield substantial differences in the dose response curve i.e. the logarithmic fit was still found to be within ±2 MU for a range of 1 – 250 MU.
Figure 85. Dose response curve for 6 and 9 MeV and a 15x15 cm² field size.

Metrics from orthogonal profiles of a single field 6 MeV 15x15 cm² irradiation of 100 MU at a dose rate of 400 MU/min at depth of maximum dose using a CR plate sandwiched inside a plastic water phantom was compared to that of the treatment planning system calculated using the eMC algorithm (Figure 86).
Figure 86. Comparison of calculated (blue solid) and measured (red dashed) orthogonal profiles for a 6 MeV irradiation at depth of maximum dose, 15x15 cm² field size, 100 MU, and a dose rate of 400 MU/min. The profiles metrics of penumbra and field size agree very well (within 1 mm), however the edge of the field is not sharp in the measured profiles.

CR-based electron dosimetry for DEAR represents an interesting approach as it is a clinically available and an efficient dosimetry technique and potentially effective at capturing absolute dose distributions.
5.2.2. Mechanical verification

An innovative feature of DEAR is the simultaneous motion of multiple mechanical axes such as couch and gantry. Having the couch move during treatment may cause some concerns about safety and secondary patient motion; however, since the motion is predetermined by the trajectory model according to the patient geometry, no unpredictable couch motion can occur, barring the malfunction of the system. Further, as the feasibility study showed, mechanical couch motions can be maintained well below the maximum mechanical velocities, such that patient motion due to inertia should not be of concern.

Verification of the mechanical axes for a treatment delivery is an important aspect. Both pre-treatment and post-treatment verification methods need to be explored. The completed work has shown that the trajectory log file can be used as a post-treatment verification tool. However, since the linac control system both delivers and records the treatment, it cannot be used as an independent verification tool. Thus, secondary and independent verification tools merit further investigation for comprehensive treatment verification to ensure reproducible and accurate mechanical axes motion during DEAR delivery (Woo and Kim 2002, Yang et al 2011, Yu et al 2014). Possible methods to independently verify couch motion could include triangulation of position using infrared cameras and markers. A commercially available option, used for respiration synchronized imaging and treatment is the Real-time Positioning
Management™ (RPM) System, which has the capability to provide real-time 3D information.

Another option would be to use the kV-MV imaging system and multiple radiopaque markers. This technique has been previously used to track internal markers during arc radiotherapy (Liu et al 2008). All these systems could use time stamps such that comparisons between the expected (trajectory model) and tracked couch motion can be made.

5.3. Delivery

Delivery efficiencies will need to be addressed. As cut-out sizes become smaller, and intensity modulation increases, the delivered MU increases. Realistic delivery times for DEAR can be estimated by the following example: For the simplified geometry used in the feasibility study, a cylinder with a diameter of 30 cm and a target area spanning 90°, the target length is approximately 25 cm. DEAR planning would require an equivalent of 8 static gantry positions to cover this area. With a prescription dose of 200 cGy per fraction for the target area, the total MU for the arc is approximately 1600 MU. At the highest dose rate of 1000 MU/min currently available clinically, the treatment time would be 1.6 min. The gantry rotation velocity is then by approximately 0.9 °/s, and the couch velocity is the lateral and vertical are 0.2 and 0.05 cm/s, respectively, which is well below the maximum velocities for these mechanical axes. It is important to note,
that the dose rate could be much higher, as it is not a technical but clinical limitation. Incorporation of high dose rates (e.g. 2400 MU/min) is feasible and could reduce delivery time.

A linac treatment delivery represents a single trajectory in a multi-dimensional space with all dependent axes such as gantry angle, couch vertical position, and even dose rate parameterized as a function of the independent variable MU (Figure 87).

![Figure 87. A delivery as it is defined in the XML file as an accumulation of discrete control points (CP). At each CP, variables such as gantry angle are functions of MU.](image)

Other parameters that may constrain this space may include patient-specific limitations e.g. the ability of a patient to extend their arms over their head for chest wall treatment may limit the range of the couch and gantry for DEAR planning. Similar efforts in 4pi radiotherapy, a non-coplanar photon therapy technique, have generated
patient-specific geometric solution spaces (Dong et al 2013). As such, plan resequencing and optimization of the deliveries needs to be investigated in order to understand the flexibility for DEAR delivery in this space as well as developing an algorithm to optimize towards delivery efficiency.

Additionally, safety issues such as collision with the patient and gantry can be addressed by constraining the space to a patient-specific geometric solution space. As discussed in section 2.4., implementation of independent quality assurance (Woo and Kim 2002, Yang et al 2011) will ensure reproducible and accurate couch motion during DEAR delivery and comprehensive pre-treatment QA and dry-run simulations must be carried out before clinical implementations. During delivery, a system that monitors the patient surface and linac in real-time could safeguard against collision.

5.3.1. Delivery accuracy

Additional constraints on delivery accuracy due to the clinical feasibilities must be investigated to ensure accurate DEAR delivery.

Patient motion during delivery will affect dose distributions, especially if the delivery is highly modulated both in energy and intensity as well as fractionated. Thus the intended dose distribution may vary from the actual dose distribution in the patient. Multiple factors contribute to the uncertainty in patient positioning including length of treatment time and comfort of the patient. Delivery efficiency could be improved by
using a higher dose rate thus reducing the overall treatment time. This in turn reduces
the likelihood of patient motion during delivery. Immobilization of the patient or target
is an important aspect that needs to be included in DEAR delivery. Further, online, real-
time tracking of the patient surface could be used to trigger a beam-hold if the patient
moves outside of a pre-determined tolerance.

For targets located on the chest, patient breathing changes SSD during DEAR
delivery thus affecting the resulting dose distributions. The displacement of the chest in
the posterior-anterior direction varies greatly between patients. Further, a given patient
can breathe differently for each treatment fraction. As shown in section 3.2.3.2.2., SSD
changes mostly affect the beam penumbra. To ensure that the intended dose distribution
is achieved, breath-hold or respiratory-gated treatments could be used. This issue is not
specific to DEAR and has been addressed in photon beam therapies treating lung
tumors, where breathing motion substantially affects the target position. The
incorporation of breathing-gated treatment for lung tumors can improve the accuracy of
photon beam delivery, since the motion of the tumor during delivery is monitored.
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Biography

Anna Elisabeth Rodrigues
Born October 13th, 1985 in São Paulo, Brazil

EDUCATION

<table>
<thead>
<tr>
<th>Degree</th>
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<tr>
<td>Ph.D., Medical Physics</td>
<td>Duke University, Durham, NC</td>
<td>2012 – Expected 2015</td>
</tr>
<tr>
<td>M.S., Medical Physics</td>
<td>Duke University, Durham, NC</td>
<td>2010 – 2012</td>
</tr>
<tr>
<td>B.S., Physics</td>
<td>University of Maryland, College Park, MD</td>
<td>2008 – 2009</td>
</tr>
<tr>
<td>B.Sc., Technical Physics</td>
<td>Vienna University of Technology, Vienna, Austria</td>
<td>2005 – 2008</td>
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PUBLICATIONS


**PRESENTATIONS**


**INVITED TALKS**


**MODERATOR**

**LEADERSHIP**

**American Association of Physicists in Medicine**

*Working Group on Medical Physics Graduate Education Program Curriculum, Member*

*American Association of Physicists in Medicine Committee & Meeting Chair on Students and Trainees Subcommittee*

- Organized and moderated student meeting at 56th annual AAPM meeting with over 200 attendees
- Planned interview workshop for students and trainees at 56th annual AAPM meeting

**Duke University Graduate Medical Physics Program**

*Contributed to the Duke medical physics graduate program through student lead initiatives, leading recruitment efforts, and contributing to decisions on the administrative level.*

- Annual Medical Physics Retreat Coordinator
- Recruitment Committee member
- Student Representative on Medical Physics Administrative Council
- Chair on the Student Advisory Board

**HONORS & AWARDS**

- Duke Medical Physics Director’s Award for Exemplary Service
- Duke Radiation Oncology and Imaging Program Annual Scientific Retreat Winner
- NC Health Physics Annual Meeting Student Paper Winner
- Duke Medical Physics Master’s Scholarship
- Duke Medical Physics Summer Research Scholarship
- University of Maryland Associate Dean’s & Dean’s List

*2010 – present*

*2013 – present*

*2015 – present*

*2012, 2013, 2014*

*2014*

*2012*

*2010 – 2012*

*2011*

*2008, 2009*