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<b>13. ABSTRACT (Maximum 200 Words)</b>  In this project, we investigate energy- and intensity-modulated radiotherapy (EIMRT) for breast cancer to deliver dose distributions that closely match the target volume and minimize the dose to critical normal structures. We have worked on the following tasks: (1) to characterize electron beams from Helium-filled accelerators for EIMRT, (2) to develop optimization algorithms for EIMRT using these electron beams, (3) to verify these optimized dose distributions using the Monte Carlo simulation technique, and (4) to compare the optimized dose plans obtained by EIMRT with conventional treatment plans and those obtained by photon intensity-modulated radiotherapy (IMRT). During the three-year research, we have performed accurate Monte Carlo simulations of the electron beams in He-filled accelerators and also investigated the effect of magnetic field modulation. Our results demonstrated that electron beams could be modulated more effectively using an electron MLC to deliver superior dose distributions for EIMRT. We have tested different algorithms for "Inverse treatment planning" to optimize breast treatment plans for EIMRT. The results confirmed that EIMRT is superior to photon IMRT and much more effective than conventional tangential photon treatments. Further studies were performed to verify the dose plans for realistic patients. The results confirm that EIMRT offers a significant advantage over conventional photon/electron treatment and over photon IMRT.				
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## **Introduction**

This project is aimed at exploring energy- and intensity-modulated electron beams for breast cancer treatment to deliver optimized conformal radiotherapy dose distributions that closely match the target volume and minimize the dose to critical normal structures. We have proposed to work on the following tasks: (1) to characterize electron beams from Helium-filled accelerators, (2) to develop optimization algorithms for energy- and intensity-modulated radiotherapy (EIMRT) using these electron beams, (3) to verify these optimized dose distributions using the Monte Carlo simulation technique, and (4) to compare the optimized dose plans obtained by EIMRT with conventional treatment plans and those obtained by photon intensity-modulated radiotherapy (IMRT).

## **Body**

Although photon beams have been an effective modality for breast cancer treatment in radiation therapy the following problems (or potential areas of improvement) remain: (1) the inclusion of the lung and sometimes of a small volume of the heart in the high-dose volume due to tumor location, patient size or in the case of chest-wall treatments; (2) lower dose near the skin surface due to lack of electron build-up in a photon beam; and (3) high exit or scatter dose to the normal structures such as the lung and heart, and more importantly the contralateral breast, which may be a major cause for the occurrence of secondary cancer in the contralateral breast for women under the age of 45.

Recent development of computer-controlled medical linear accelerators along with improved treatment planning techniques, may provide new solutions in the delivery and control of external beam radiation through beam-intensity modulation<sup>1-9</sup>. It is expected that using photon IMRT, the problem (1) above may be significantly improved but (2) will remain and (3) may become more serious as treatment time increases with the number of fields used (increased leakage or scattering dose). With modulated electron beams<sup>10-15</sup>, on the other hand, problem (1) may also be significantly improved and problems (2) and (3) can be completely eliminated due to the nature of electron beams.

In this project we proposed to study He-filled accelerators equipped with computer-controlled multi-leaf collimators and the Monte Carlo treatment planning technique for energy- and intensity-modulated electron beams for breast cancer to significantly improve the dose uniformity in the target volume, to exclude the lung and heart from the high or moderate dose volume and to eliminate the scatter dose to the lung, the heart and the contralateral breast to reduce complications and late effects associated with breast cancer radiotherapy. The purpose of this study is to determine the degree of feasibility of using electron EIMRT to improve breast cancer treatment.

This project has 4 specific aims: (1) To perform Monte Carlo beam simulation; (2) To develop an optimization algorithm for EIMCRT; (3) To perform Monte Carlo dose verifications; and (4) To evaluate the optimized treatment plans.

We report on the research accomplishments associated with the tasks outlined in the approved "Statement of Work" below for our research between Oct. 1, 1998 and Sept. 30, 2001. Since our results for the first two years have been reported in detail in our earlier reports we only provide detailed information (with figures) for the results in the third year.

## **Task 1. Perform Monte Carlo beam simulations:**

### Simulation of realistic electron beams from He-filled clinical accelerators

The EGS4<sup>16</sup> BEAM<sup>17</sup> system has been used in this work to investigate the electron beams from clinical linear accelerators. The BEAM code produces a phase-space output of the beam (i.e., the energy, charge, position, direction, and a tag called LATCH to record the particle history) at any specified plane in the simulation geometry. The simulation geometry may consist of a series of individual component modules (CMs) positioned perpendicularly to the beam axis. The CMs used in this work to simulate the clinical linear accelerators included SLABS and CONESTAK for electron scattering foils, CONESTAK for photon target and primary collimators, and shielding rings, FLATFILT for photon flattening filters, CHAMBER for the monitor chamber, and dose simulation phantoms, APPLICAT for electron applicators, JAWS for the secondary photon collimators, MLC for the multileaf collimator, MIRROR for the light mirror and BLOCK for electron cutouts. The actual dimensions and materials used for each of the components were handled by an input data file, which also contained the parameters required for the Monte Carlo transport simulations such as the energy cutoffs for electron (ECUT) and photon (PCUT) transport, minimum energy for creation of knock on electrons (AE) or bremsstrahlung photons (AP), maximum electron step length (SMAX), maximum energy loss per electron step (ESTEPE), and the incident beam parameters. The simulated phase-space data can be used as source input for further BEAM simulations or dose calculations using the code DOSXYZ (see below) or analyzed using various software such as BEAMDP (BEAM Data Processor) to derive particle energy spectra, fluence and angular distributions.

The EGS4/DOSXYZ code system was designed for dose calculations in a 3D rectilinear voxel geometry<sup>17,20</sup>. Voxel dimensions are completely variable in all three directions. Every voxel (volume element) can be assigned to a different material. The cross-section data for the materials used are available in a pre-processed PEGS4 cross-section data file. The density of the material defaults to that in the PEGS4 data file but can be varied in a DOSXYZ calculation for use with the patient's CT data although the density effect corrections for the stopping powers of the material remain unchanged<sup>18,24</sup>. The phase-space data obtained from a BEAM simulation can also be used as a source input to imitate all possible beam positions and directions from the linear accelerator. Dose contributions from different beam components were selectively calculated based on the particle charge or the LATCH settings specified in the BEAM simulation. DOSXYZ produced a data file that contained geometry specifications such as the number of voxels in all the three directions and their boundaries as well as the dose values and the associated ( $1\sigma$ ) statistical uncertainties in the individual voxels.

We have simulated 6 – 20 MeV electron beams from Varian Clinac 2100C and 2300CD clinical accelerators. We have replaced the air in the treatment head and in the gap between the treatment head and the patient with Helium to study the effect of air scattering in air and the potential improvement with He-filled accelerators. The calculated planar fluence distributions at a source-detector distance of 90 cm for 10.08 MeV electrons from the Clinac 2300. The thick solid histogram represents standard geometry (case 1) and the thin solid histogram represents helium atmosphere (case 2). The fluence distributions can be significantly improved by replacing the air in the linac head with Helium. This is especially important for lower incident electron energies (below 10 MeV) and large gaps between the treatment head and the patient<sup>19,21-24</sup>.

We have also investigated the effect of the MLC leaf shape on the beam profiles. The Varian MLC has rounded leaf ends while the MLCs from other manufacturers have double focused leaf shape. The calculated dose distributions at phantom surface for a 20 MeV electrons from the Clinac 2100CD

accelerator (see Appendix). The blue histogram represents a focused photon MLC and the red histogram represents unfocused photon MLC (straight edge). For comparison, a specially designed electron MLC with straight edge was also simulated with 7 cm air gap (i.e., at the electron applicator level). The photon MLC simulations were performed for a 20 cm He gap. It is clear that focused MLC will result in significant improvement in the dose profile for modulated electron beams. Unfocused MLC will degrade the dose profiles near the beam penumbra and the effects can be seen at different depths. This means the Varian MLC rounded leaf shape has to be modified if it is used to deliver modulated electron beams.

Although this work is designed to investigate the feasibility of He-filled accelerators for energy and intensity modulated electron beam therapy. The results have been presented to the accelerator manufacturers for evaluation. Possibilities have been discussed regarding fill the accelerator with He by sealing the treatment head or installing a He-filled balloon. Further studies will be made when the sealing (or balloon) materials and thicknesses are known.

#### Commission simulated beam data by comparisons with measurements

In order to commission the simulated the clinical beams, we have compared the dose distributions simulated by the Monte Carlo method and those measured using ion chambers and diodes. The calculations were performed using the DOSXYZ user code. The simulation parameters were modified to improve the simulated beam data to match the measured data until the two sets of data agreed with each other to within 2% of the maximum dose. We have also compared the dose distributions in heterogeneous phantoms. The heterogeneity correction factors for layered lung or layered bone phantoms irradiated by 12 MeV electron beams (see Appendix). They are calculated as ratios of the doses in a heterogeneous phantom to the doses in a water phantom calculated by Monte Carlo, by FOCUS (3D pencil beam) and by measurements. Our results clearly demonstrate that the Monte Carlo method can faithfully reproduce the dose distributions in these phantoms while the conventional dose calculations may significantly underestimate or overestimate the dose in or near heterogeneous regions. These findings are consistent with the previous results<sup>22-26</sup>.

#### Study the characteristics of intensity-modulated electron beams

We have studied the characteristics of small field electron beams collimated by blocks and an electron MLC<sup>29</sup>. We have studied the beam profiles from different MLC leaf shapes. It is found that the MLC shape will affect the dose profile near the phantom surface but have little effect on the dose profiles at depth beyond the depth of maximum dose. This supports the idea of using a photon MLC or an electron MLC to deliver small field electron beams to form intensity modulated electron fields. The dose distributions for a 2 cm x 2 cm field and a 4 cm x 4 cm field collimated by a photon MLC and an electron MLC (see Appendix). The difference between these dose distributions is insignificant clinically. The dose distributions formed by four 1 cm x 1 cm fields and those formed by one 1 cm x 4 cm field. The dose distributions are identical at depths beyond the depth of maximum dose but are slightly different on the surface – the four small fields give more fluctuations. However, the leakage dose outside the treatment is more significant for the multiple small fields because of the longer delivery time. This leakage can be reduced by increasing the MLC leaf thickness.

#### Characterize the simulated beams using simplified source models

A multiple source model was developed by Ma et al for electron beam Monte Carlo treatment planning<sup>19</sup>. The algorithm has been further improved by Jiang et al<sup>26</sup> for beam commissioning in

clinical routine dose calculations. A hybrid approach for commissioning electron beam Monte Carlo treatment planning systems is studied. The approach is based on the assumption that accelerators with similar designs have similar electron beam characteristics. For one type of accelerators, a reference machine is selected and simulated with the Monte Carlo method. A beam model is built for this type of accelerators using the Monte Carlo simulated phase space information for the reference machine. When commissioning another accelerator of the same type, the appropriate parameters in the beam model are tuned according to the standard measured data such as output factors, depth-dose and dose profiles. A Varian Clinac 2100C accelerator is chosen as the reference machine and simulated using the EGS4/BEAM code. A four-source beam model is established based on the simulated beam information to reconstruct electron phase space down to the last applicator scraper. The model includes a point electron source for direct electrons and electrons scattered from the primary collimator and jaws, a point photon source for all contaminant bremsstrahlung photons, and two square ring electron sources representing electrons scattered from the two scrapers above the last scraper. A Varian Clinac 2300C/D machine, which is similar in design to the reference machine, is commissioned using this beam model. By tuning the appropriate parameters in the model, accurate dose calculation is achieved using the model, compared to the corresponding measurement.

In this work, we have characterized the Monte Carlo simulated beams using this improved source model. We have compared the dose distributions calculated using the source model and the original simulated phase space data and achieved good agreement (within 2% of maximum dose). The dose distributions calculated by the source model and measured by an ion chamber for a 12 MeV electron beam from a Varian Clinac 2300CD accelerator are compared. The agreement was within 1% for both cases. Similar results have been obtained for other beam energies and field sizes (as reported in previous report and in Jiang et al 1999 of peer-reviewed papers). This demonstrated the validity of the source model for beam commissioning for this study.

#### Install and test MCDOSE for photon calculations

We have further developed a Monte Carlo EGS4 user code MCDOSE for photon/electron beamlet and treatment plan dose calculations. Good agreement was achieved between the MCDOSE results and measurements. The MCDOSE code has been implemented on a 32-PC network for Monte Carlo dose calculations. Features of MCDOSE include a multiple-source model to reconstruct the beam phase space, inclusion of beam modifiers such as jaws, wedges, blocks, compensators and electron cutouts in the patient simulation, the implementation of several variance reduction techniques, and suitable for both conventional and intensity modulated radiation therapy (IMRT) treatment planning. Before MCDOSE is used reliably for dose calculation in clinic, it must be properly validated. The clinical validations for beam modifiers and dose calculation are presented. A comparison of the dose distribution with 45-degree wedge in a water phantom made between MCDOSE and EGS4/BEAM/DOSXYZ<sup>17,20</sup> demonstrates that MCDOSE can give accurate result with wedge. The dose distributions for a blocked 10x10 15MV photon beam in a water phantom are also compared between them. They agree very well. The comparison of Electron cutout factors between MCDOSE and measurements show a good agreement too. After applying variance reduction techniques in MCDOSE, the agreement of dose distributions in specifically designed inhomogeneous phantoms between MCDOSE and DOSXYZ is within the statistical uncertainty of 0.5%. All these results demonstrate that MCDOSE is accurate for routine dose calculation in radiotherapy treatment planning. The heterogeneity correction factors calculated by MCDOSE for layered-lung or layered-bone phantoms were consistent with results from measurement to within 1%. Due to the elegant variance reduction techniques, MCDOSE is also faster than EGS4/DOSXYZ<sup>20</sup> dose calculation by a factor of

up to 30. A nine field IMRT planing can be done in 1-4 hours on a personal computer, including pre- and post-optimization dose calculation.

To test the accuracy of dose calculation by the MCDOSE code, comparison calculations were performed with DOSXYZ, which is an EGS4 usercode developed by the NRCC group and well benchmarked by the radiotherapy investigators. Each example has been calculation with source surface distance (SSD) of 100cm and field size of 6cm x 6cm or 10 cm x 10cm. The global electron cut off energy (ECUT) was set to 0.7MeV and photon cut off energy (PCUT) was set to 0.01MeV in MCDOSE and DOSXYZ respectively. The fractional energy loss per step (ESTEPE) was limited to 40% for DOSXYZ and 20% for MCDOSE. The maximum step size (SMAX) is set to default 5cm for DOSXYZ and 2mm for MCDOSE. Voxel size in the phantom is set to 4mm x 4mm x 5mm. The material compositions and densities are taken from ICRU Report No. 46. Mono-energetic point sources and phase space files, multiple source models for Clinac 2300C/D linear accelerator are used as input to do simulation. Since dose per incident fluence is calculated, absolute dose distributions produced by MCDOSE and DOSXYZ are compared. Excellent agreement has been achieved as reported in our 1999 annual report and published in a paper by Li et al (2000, see peer-reviewed papers).

Initial schedule included the implementation of the PEREGRINE code system for this study. This was not done because of the delay of the commercial availability of the PEREGRINE system and the complication of the commercialization of the PEREGRINE system. The system has been exclusively licensed to the NOMOS, Corp. and it is no longer available as free software. Instead, we have developed the MCDOSE system, which is based on the EGS4 system, which is free for research and education use. The MCDOSE code has been validated against the DOSXYZ code and proved to be equally accurate but about 20 times faster in computation speed. This is therefore our suggestion to replace the PEREGRINE software with MCDOSE for this study.

## **Task 2. Develop optimization algorithms for EIMCRT:**

### Evaluate dose calculation algorithms for optimization calculation

We have compared the dose distributions calculated by the conventional dose calculation algorithms (the pencil beam algorithm in a commercial treatment planning system, the FOCUS system by Computerized Medical System, St. Luis, MO) and the Monte Carlo method (with MCDOSE). It was found the dose distributions predicted by the pencil beam were very uncertain under the conditions of extended source surface distance (SSD), oblique incidence and for heterogeneous phantoms<sup>15,18,21,22</sup>. We have calculated dose distributions for different field sizes, incident beam energies, SSDs and phantom geometries. The pencil beam dose distributions calculated by the FOCUS system and the Monte Carlo system for a 20 MeV electron beam (see Appendix). Monte Carlo correctly predicted the dose variations near the interface between tissue and air and between tissue and bone while the FOCUS calculation shows almost symmetrical distributions in these cases. It is concluded that Monte Carlo is needed for the electron beamlet calculation to ensure the accuracy of the optimization process. In our future studies, we have scheduled to compare dose distributions using these algorithms and the optimization algorithms for realistic patient treatment plans.

### Investigate suitable objective functions for electron beam optimization

We have installed several objective functions for the optimizer used for electron beam EIMRT. We have selected a dose based objective function with soft constraints using dose volume histogram

information. With this final selection decided, the optimization can be performed by using a dose based penalty function method and the center-of-mass method to minimize the augmented objective function. An improved conjugate vector method can also be used. For the target area, a quadratic form of objective function is specified. In addition, two target dose-uniformity constraints are used to ensure a uniform target dose distribution and to distinguish the clinical importance of cold and hot spots. For the critical structures, maximum-dose constraint and several levels of dose-volume constraints are assigned to each structure. For each objective function and constraint, an importance weight relative to the target objective function is assigned. All the constraints are mathematically transformed to the penalty functions of quadratic forms. The augmented objective function, which should be minimized, is a combination of the original objective functions and all penalty functions. The detailed methods have been reported in our publications<sup>24</sup> and the applications have been presented.

### Develop fast iterative optimization algorithms

A fast iterative optimization algorithm has been developed for electron beam optimization. The algorithm has been tested for both electron treatment planning optimization and photon beam optimization. The dose calculation can be cast as the matrix equation,

$$\mathbf{D} = \mathbf{M} \cdot \mathbf{I}$$

where  $\mathbf{D}$  is the vector whose elements are the dose deposited in the voxels in the calculation volume,  $\mathbf{I}$  is the vector whose elements are the beamlet weights, and  $\mathbf{M}$  is the matrix relating the weights of each beamlet to the dose deposited in each voxel. Using the Monte Carlo method, the individual beamlet distributions that make up the matrix elements of  $\mathbf{M}$  will be calculated separately taking into considerations of the location of the beamlet in the field, patient contour, inhomogeneous patient anatomy, organ motion and patient setup uncertainty. The beamlet dimensions will be specified through the RTP module, based on the MLC leaf width. For the Varian MLC, the beamlet size will be variable between 10mm x 1mm and 10mm x 10mm. The beamlet profile will be different depending on the location of the beamlet (its slight spatial dependence on the accelerator head scatter, beam flatness and SSD variation). This will be automatically included when using the simulated phase-space data as source input. The MCDOSE code has been modified for the dose calculations in the patient. The beam incident angles and patient CT data will be obtained from the RTP system. Each beamlet will require 1 – 10 million phase space photons depending on the dimensions of the beamlet. For a typical IMRT case with 9 gantry angles, the beamlet calculation (for 1 – 10 thousand beamlets) will be completed in a few minutes on the PC network after the variance reduction techniques are implemented.

Our optimization procedure developed to calculate the optimal intensity profiles consists of the following two stages. First, the planner inputs the patient geometry and defines the treatment setup, such as the beam energy, number and orientations of beams, etc. The target volume and the critical structures are defined by the clinician. The planner also determines the size of the beamlets and number of dose or constraint points placed inside the target or critical structures, according to the patient anatomy, the required computation accuracy and the available computation time. Each broad beam is divided into beamlets and the dose and constraint points are uniformly and randomly placed inside the corresponding area. Then a reference monitor unit is assigned to each open rectangular beam and the dose deposition coefficients, which is defined as the dose contribution from a beamlet to a point, are calculated using MCDOSE and a conventional dose calculation model for comparison.

Second, using the calculated dose deposition coefficients as input, the optimal intensity profile for each beam is achieved by using the Zangwill's penalty function method and the center-of-mass method to minimize the augmented objective function. An improved conjugate vector method will also be

investigated. For the target area, a quadratic form of objective function is specified. In addition, two target dose-uniformity constraints are used to ensure a uniform target dose distribution and to distinguish the clinical importance of cold and hot spots. For the critical structures, maximum-dose constraint and several levels of dose-volume constraints are assigned to each structure. For each objective function and constraint, an importance weight relative to the target objective function is assigned. All the constraints are mathematically transformed to the penalty functions of quadratic forms. The augmented objective function, which should be minimized, is a combination of the original objective functions and all penalty functions. The results of the optimization process are the intensity profiles for the individual gantry angles (photon fields).

After the optimization calculation, beam intensity maps at each gantry angle (the elements of  $\mathbf{I}$ ) will be generated and leaf-setting sequences will be computed using a leaf sequencing algorithm. The final dose calculation will be performed again using MCDOSE. The effect associated with leaf and jaw movement will be accounted for in these calculations.

The monitor units  $MU_k$  required for the  $k^{\text{th}}$  field can be calculated using the following equation:

$$MU_k = D_p w_k F_k / (N S_p)$$

The weight assigned to the  $k^{\text{th}}$  field,  $w_k$ , will be the maximum weight for the beamlets included in field  $k$ . The plan normalization factor,  $N$ , will be the ratio of the Monte Carlo calculated maximum dose in the patient to that in water under the reference conditions. The modulation scaling factor,  $F_k$ , will be the ratio of the total MUs required to deliver all the beamlets in field  $k$  using the leaf-setting sequence to the MUs required to deliver the beamlet that has the maximum weight. In preliminary tests, the Monte Carlo calculated output factors for fields 1 cm x 1 cm to 40 cm x 40 cm agreed to within 2% with measurements. No fudge factors will be needed in the Monte Carlo based monitor unit calculations. The dose values for different beam energies under the reference conditions will be calculated only once and stored in the system database for calculating the normalization factor  $N$ .

We have been evaluating different leaf sequencing algorithms suitable for both “stop and shoot” and dynamic delivery. A new algorithm also synchronizes the leaf sequences to remove the “tongue and groove” effect. The results showed that the difference in the beam delivery time using a dynamic MLC between “stop and shoot” (including beam-off time for leaf movement) and dynamic delivery was clinically insignificant. We will install a leaf sequence algorithm for our project, which uses the stop and shoot algorithm and also synchronizes the leaf’s movement to remove the “tongue and groove” effect. We will further work on other MLCs when the leaf sequence file format becomes available for operation with electron beams.

#### Install optimization software for clinical treatment planning system

We have developed a software interface called MCCALC to perform treatment planning optimization for clinical applications. The interface provides a graphics user interface (GUI) for the user to input optimization parameters and to acquire patient CT and contour information from a clinical treatment planning system (FOCUS, CMS, Inc). The computation of the beamlet dose distributions will be performed on a networked parallel computer system using the Monte Carlo method that we have established based on previous work<sup>16-23</sup> and the results will be used for the optimization process. The final intensity maps for each treatment field will be sent back to the clinical treatment planning system to beam delivery using custom made electron cutouts or using computer controlled electron multileaf collimator (eMLC). More detailed descriptions of the optimization algorithm and its implementation have been reported in a publication<sup>24</sup> and here we only describe briefly for completion.

A fast iterative optimization algorithm has been developed for electron beam optimization. The algorithm has been tested for both electron treatment planning optimization and photon beam optimization. The beamlet dimensions will be specified through the RTP module, based on the MLC leaf width. For the Varian MLC, the beamlet size will be variable between 10mm x 1mm and 10mm x 10mm. The beamlet profile will be different depending on the location of the beamlet (its slight spatial dependence on the accelerator head scatter, beam flatness and SSD variation). This will be automatically included when using the simulated phase-space data as source input. The MCDOSE code (see below) has been modified for the dose calculations in the patient. The beam incident angles and patient CT data will be obtained from the RTP system. Each beamlet will require 1 – 10 million phase space photons depending on the dimensions of the beamlet. For a typical IMRT case with 9 gantry angles, the beamlet calculation (for 1 – 10 thousand beamlets) will be completed in a few minutes on the PC network after the variance reduction techniques are implemented.

The optimization procedure consists of the following two stages. First, the planner inputs the patient geometry and defines the treatment setup, such as the beam energy, number and orientations of beams, etc. The target volume and the critical structures are defined by the clinician. The planner also determines the size of the beamlets and number of dose or constraint points placed inside the target or critical structures, according to the patient anatomy, the required computation accuracy and the available computation time. Each broad beam is divided into beamlets and the dose and constraint points are uniformly and randomly placed inside the corresponding area. Then a reference monitor unit is assigned to each open rectangular beam and the dose deposition coefficients, which is defined as the dose contribution from a beamlet to a point, are calculated using MCDOSE and a conventional dose calculation model for comparison. Second, using the calculated dose deposition coefficients as input, the optimal intensity profile for each beam is achieved by the optimization process using an objective function. The results of the optimization process are the intensity profiles for the individual gantry angles (photon fields).

The results of the optimization calculation are the beam intensity maps at each gantry angle and then they can be used to generate leaf-setting sequences using a leaf-sequencing algorithm. The final dose calculation will be performed again using MCDOSE. The effect associated with leaf and jaw movement will be accounted for in these calculations. We have been evaluating different leaf sequencing algorithms suitable for both “stop and shoot” and dynamic delivery. A new algorithm also synchronizes the leaf sequences to remove the “tongue and groove” effect. The results showed that the difference in the beam delivery time using a dynamic MLC between “stop and shoot” (including beam-off time for leaf movement) and dynamic delivery was clinically insignificant. We will install a leaf sequence algorithm for our project, which uses the stop and shoot algorithm and also synchronizes the leaf’s movement to remove the “tongue and groove” effect. We will further work on other MLCs when the leaf sequence file format becomes available for operation with electron beams.

### **Task 3. Perform Monte Carlo dose verification:**

#### Verify electron EIMRT and conventional electron treatment using EGS4/MCDOSE

We have developed a Monte Carlo EGS4<sup>16</sup> user code MCDOSE<sup>25</sup> for electron beamlet and treatment plan dose calculations. Good agreement was achieved between the MCDOSE results and measurements<sup>24-28</sup>. Features of MCDOSE include a multiple-source model to reconstruct the beam phase space<sup>26</sup>, inclusion of beam modifiers such as jaws, wedges, blocks, compensators and electron cutouts in the patient simulation, the implementation of several variance reduction techniques, and suitable for both conventional and intensity modulated radiation therapy (IMRT) treatment planning.

Before MCDOSE is used reliably for dose calculation in clinic, it must be properly validated. The clinical validations for beam modifiers and dose calculation are presented. A comparison of the dose distribution with 45-degree wedge in a water phantom made between MCDOSE and EGS4/BEAM/DOSXYZ<sup>17,19,20</sup> demonstrates that MCDOSE can give accurate result with wedge. The dose distributions for a blocked 10x10 15MV photon beam in a water phantom are also compared between them. They agree very well. The comparison of Electron cutout factors between MCDOSE and measurements show a good agreement too. After applying variance reduction techniques in MCDOSE, the agreement of dose distributions in specifically designed inhomogeneous phantoms between MCDOSE and DOSXYZ is within the statistical uncertainty of 0.5%. All of these results demonstrate that MCDOSE is accurate for routine dose calculation in radiotherapy treatment planning. The heterogeneity correction factors calculated by MCDOSE for layered-lung or layered-bone phantoms were consistent with results from measurement to within 1%. Due to the elegant variance reduction techniques, MCDOSE is also faster than EGS4/DOSXYZ<sup>20</sup> dose calculation by a factor of up to 30. A nine field IMRT planing can be done in 1-4 hours on a personal computer, including pre- and post-optimization dose calculation<sup>25</sup>.

Initial schedule included the implementation of the PEREGRINE code system for this study. This was not done because of the delay of the commercial availability of the PEREGRINE system and the complication of the commercialization of the PEREGRINE system. The system has been exclusively licensed to the NOMOS, Corp. and it is no longer available as free software. Instead, we have developed the MCDOSE system, which is based on the EGS4 system, which is free for research and education use. The MCDOSE code has been validated against the DOSXYZ code and proved to be equally accurate but about 20 times faster in computation speed. Therefore we have replaced the PEREGRINE software with MCDOSE for this study.

#### Verify photon IMRT and conventional tangential photons using EGS4/MCDOSE

To verify the IMRT dose distributions calculated by the commercial treatment optimization system (CORVUS, NOMOS Corp.) we have modified the MCDOSE code to compute the dose from the leaf sequence files from the commercial system. First we performed calculations for a homogeneous PMMA phantom. Since CORVUS calculates dose to water while PMMA is not water equivalent it is important to establish a conversion scheme so that the measured ionization in the PMMA phantom can be converted to dose to water in order to verify the dose distributions computed by CORVUS. The Monte Carlo method computes both dose to water and dose to PMMA and the results can be used to study the energy fluence perturbation correction factor for the PMMA phantom (compared to a water phantom). IMRT plans have been computed using a water phantom and the PMMA phantom (all have the same dimensions) by CORVUS and Monte Carlo and the results were compared to measurements. For example, for a 9 field IMRT plan with 15 MV beams, the difference between the calculated and measured dose was 1.4% for the water phantom and 1.6% for the PMMA phantom. In general, the difference between the measured and calculated values was within 4% with an estimated measurement uncertainty of about 3% and Monte Carlo calculation uncertainty of 2%.

We have also computed dose distributions in heterogeneous PMMA phantoms with either lung or bone inserts. The dose distributions calculated by Monte Carlo for a 15 MV beam plan and Table I shows the measured and calculated dose values in the heterogeneous phantom for both 4 and 15 MV beam plans. The two plans were generated using our Monte Carlo based inverse planning system using the same parameters except for the beam energy. The Monte Carlo calculated values differed from the measured values by less than 1% for both 4 and 15 MV beam plans while the CORVUS values differed from the measured values by 1% for the 4 MV plan and 5% for the 15 MV plan. Similar

agreement was obtained for other phantom configurations and beam energies. It is concluded that the Monte Carlo method is more accurate in predicting the dose in heterogeneous phantoms for IMRT dose verification and can be used to validate the dose distributions for IMRT as part of the QA procedure<sup>27</sup>.

TABLE I  
Dose at the reference measurement point.

Beam energy	Monte Carlo	CORVUS	Measurement
4 MV	2.177 Gy	2.201 Gy	2.177 Gy
15 MV	2.146 Gy	2.276 Gy	2.161 Gy

**Task 4. Compare optimized treatment plans for 20 patients:**

The specific steps for this task included:

- Acquire and outline CT sets
- Generate optimized plans using our software and CORVUS for breast treatment
- Compare these plans with conventional photon plans
- Evaluate the results using different objective functions

To carry out this task, we have recruited breast patients with different disease sites and treatment conditions. These included patients with large/small intact breasts, chestwall only treatments, left/right breasts, and a variety of nodal conditions. The patients were CT scanned using our AcQsim system and the target and critical structures were outlined by oncologists. We then use our own software to calculate the dose distributions and then optimize the treatment plans for these patients. The same treatment criteria were used for the photon IMRT using the NOMOS CORVUS treatment optimization software and for the conventional treatments using the CMS FOCUS RTP system. Different objective functions were used to check the results and evaluate the optimization software. It is found that the effect of the objective function on the treatment plans is significant but independent of the treatment techniques. This means that the plan evaluations for different treatment techniques are consistent for any objective functions.

We have compared the following photon treatment techniques:

(a) *Standard tangential photon beams*

Lateral wedge and optimize relative weights

(b) *Tangential fields with intensity-modulation*

Improved dose homogeneity compared to standard tangents

Breathing motion still dictates a posterior margin and a flash region

Tangential IMRT fields can provide for heart and lung sparing by “curving” the posterior edge of the field. However, the high dose region caused by the concavity of the chest wall cannot be avoided.

(c) *More ‘general’ intensity modulation*

Additional en face type fields may improve dose conformity

We use a four field setup described by Gagliardi et al, Prediction of excess risk of long-term cardiac mortality after radiotherapy of stage I breast cancer (Radiotherapy and Oncology 46 (1998, 63-71) Adding more beams may help homogeneity, but this example demonstrates the lack of acceptable beam angles, even with IMRT

In figure 1 we show the isodose distributions for the photon beam treatments. The dose volume histograms (DVH) are shown in figure 2 for the corresponding plans. The best homogeneity is achieved with IMRT tangents. 4 field IMRT reduces peak heart and lung dose, but at the expense of a large low dose region. For the electron plans, we have used two gantry angles: 119 and 164 degrees (Fig. 3). Both angles are only slightly affected by breathing motion, in contrast to the photon plans, which required a 1-1.5 cm margin to account for the severe respiration effect. Four electron energies were delivered down each of the two fields: 6, 9, 12, 16 MeV, resulting in a plan with a total of 8 fields to be delivered. Electron dose calculation and optimization was based on a source model for a Varian Clinac 2100C, including contaminant photons. Figure 4 shows the intensity maps for the 8 fields. Figure 5 compares the isodose distributions for conventional photon beams, IMRT tangents and modulated electron beams. For convenience, we removed the 4-field IMRT since it is considered to be inferior and would not be recommended by the oncologists for treatments. Figure 6 shows the DVH comparisons for the three techniques shown in Fig. 5. The best homogeneity is still achieved with IMRT tangents. MERT gives homogeneity comparable to that of conventional tangents. Heart sparing is greatly improved, while high dose to the lung is also reduced. Healthy tissues, defined as all non-target volumes in the body receive the least dose in the MERT treatment. The residual electron range increases low lung dose, but the high dose regions are significantly reduced in normal tissues (Fig. 7). Compared with tangential photon beams, MERT reduces both peak cardiac and lung doses by over 20 Gy. Almost no volume receives greater than 20 Gy in MERT (Fig. 8).

It is found that conventional tangents can lead to overdosing in the thinnest regions of the breast (superior and inferior borders near the apex). These region specific hotspots are most likely to manifest as clinical side effects, such as edema. Greater flexibility in intensity modulation allows IMRT tangents to correct for this, though hotspots in entrance tissue may occur. MERT has more dose heterogeneity than IMRT tangents, but unlike conventional tangents, dose hotspots are more widely dispersed throughout the target region, not localized in specific geometric features. The 110% hotspots are also absent in the MERT plan. More work has been reported elsewhere on dose verification, beam delivery techniques and optimization methods for modulated electron radiotherapy, which support the results and conclusions of this project<sup>30-34</sup>.

Based on our results and the above discussions we have reached the following conclusions on the plan comparisons:

- MERT provides significant (20 Gy) reduction in peak lung and heart dose compared to either conventional tangents or IMRT tangents. The breathing margin may be significantly reduced because electron beams do not require direct tangential fields. There is an increase in the low dose region (< 5 Gy) in the use of electrons.
- The best target homogeneity is achieved with IMRT photons. MERT offered significant improvement over tangents for the plan shown, with dose heterogeneities dispersed throughout the target region.
- The use of intensity modulated electrons or photons offers significant advantages over conventional treatments. However, continued research is necessary into aspects of beam planning and delivery for both photons and electrons.
- The ideal scenario may involve combined modulated electrons and photons, such as the use of MERT for tumor bed boosts or mediastinal node irradiation and photons for whole breast irradiation.

## Investigate beam characteristics of electron beams modulated by magnetic fields

From our previous studies, it seems clear that an electron MLC will be able to deliver modulated electron fields accurately and therefore He-filled accelerator design may not be necessary (but will further improve the characteristics if implemented). We also set out to study other means to improve the electron dose distributions for modulated electron beam therapy for breast cancer. We have achieved some preliminary results and based on these results we have put out another DOD grant application to investigate the feasibility of using magnetic fields to modulate the beam at depth to reduce the dose to the distal organs such as the lung (see below). We have published a paper on this subject (Lee and Ma 2000, see peer-reviewed papers) and we briefly describe our results in the following paragraphs.

In this study, Monte Carlo simulations were employed to study the characteristics of the electron beams of a clinical linear accelerator in the presence of 1.5 and 3.0 T transverse magnetic fields and to assess the possibility of using magnetic fields in conjunction with modulated electron radiation therapy (MERT). The starting depth of the magnetic field was varied over several centimeters. It was found that peak doses of as much as 2.7 times the surface dose could be achieved with a 1.5 T magnetic field. The magnetic field was shown to reduce the 80% and 20% dose drop-off distance by 50% to 80%. The distance between the 80% dose levels of the pseudo-Bragg peak induced by the magnetic field was found to be extremely narrow, generally less than 1 cm. However, by modulating the energy and intensity of the electron fields while simultaneously moving the magnetic field, a homogeneous dose distribution with low surface dose and a sharp dose fall-off was generated. Heterogeneities are shown to change the effective range of the electron beams, but not eliminate the advantages of a sharp depth-dose drop-off or high peak-to-surface dose ratio. This suggests the applicability of MERT with magnetic fields in heterogeneous media. The results of this study demonstrate the ability to use magnetic fields in MERT to produce highly desirable dose distributions.

Monte Carlo calculations were performed in which electrons were constrained to travel down the same initial axis, i.e. a pencil beam. These results clearly show the curvature of the electron beam dose distribution generally along the expected track. By superposition of many pencil beams along the lateral axis, it is clear that Bragg-peak-like dose peaks can be constructed. Narrow fields, defined by 3 x 3 cm<sup>2</sup> Cerrobend cutouts on 6 x 6 cm<sup>2</sup> applicators, were also simulated. What is apparent is that there is a significant shift of the dose peak off the central axis. These data are presented primarily as a demonstration of the effects of the magnetic field, i.e. the curvature of the beam and the formation of a dose peak. However, it is difficult to make any definitive statements regarding depth-dose distributions because of the lack of equilibrium, as discussed below.

Previous discussions of MERT have considered the use of fields with non-uniform energies and intensity distributions. Here, the technique is extended to the use of multiple magnetic field positions, i.e. a single port may include a 20 MeV field with a magnetic field beginning at 3.0 cm depth for a certain number of monitor units, and then another 20 MeV field with the magnetic field located at a depth of 5.0 cm delivered down the same axis, etc. It is instructive to compare this methodology with the more familiar method used in proton beam therapy. In that system, physical blocks of different thicknesses are used to shift the Bragg peak to different depths, and intensity modulation is provided by allowing the beam to dwell on a given modulator for a variable time period. In this case, the only difference is the means of moving the pseudo-Bragg peak, that is, by moving the magnetic field. Beam intensity is modulated simply by changing the number of monitor units delivered with the magnetic field at a given position. It was believed that, as with the proton beam, useful fields can be constructed while maintaining the desirable depth-dose drop-off and low skin dose. We have used magnetic fields

modulated electron beamlets for optimization and generate treatment plans for a realistic breast treatment plan. Figure 9 show the isodose distributions and the dose volume histogram data for the two plans, one with magnetic fields and the other without. It is very clear that magnetic field modulated electron beams show much better target conformity and normal tissue sparing. Based on these results, we have submitted an “IDEA” proposal to further investigate this novel method, and if funded we hope to improvement breast MERT significantly.

We can demonstrate how this works as follows. A simple one-dimensional optimization routine was utilized to generate two fields. In one case, a target profile was generated where the dose would rise linearly from 50% at the surface (relative to the maximum along the profiled axis) to 100% at 1.0 cm, then drop to zero at 6.2 cm. The optimized solution was solved numerically, and a linear combination of the 1.5 T fields was generated to match the target values in a least-squares sense. The results are shown in Fig.10. The width of the treatment region, 80/80, was found to be 5.0 cm, followed by a 80/20 drop-off of 0.77 cm. Between the depths of 1.6 cm and 5.6 cm, the dose was homogeneous to within 5.4% of the average dose. For more detailed descriptions of the magnetic field modulation please see reference.

### **Key Research Accomplishments**

We have accomplished the following tasks:

*Simulation of realistic electron beams from He-filled clinical accelerators:* We have simulated 6 – 20 MeV electron beams from a Varian Clinac 2100C clinical accelerator. We have replaced the air in the treatment head and in the gap between the treatment head and the patient with Helium to study the effect of air scattering in air and the potential improvement with He-filled accelerators.

*Commission simulated beam data by comparisons with measurements:* We have compared the dose distributions simulated by the Monte Carlo method and those measured using ion chambers and diodes. The simulation parameters were modified to improve the simulated beam data to match the measured data until the two sets of data agreed with each other to within 2% of the maximum dose.

*Study the characteristics of intensity-modulated electron beams:* We have studied the characteristics of small field electron beams collimated by blocks and an electron MLC. We have studied the beam profiles from different MLC leaf shapes.

*Characterize the simulated beams using simplified source models:* We have developed simplified source models for all clinical electron beams simulated with the Monte Carlo method. We have compared the dose distributions calculated using the source model and the original simulated phase space data and achieved good agreement (within 2% of maximum dose).

*Install and test MCDOSE for photon calculations:* We have further developed a Monte Carlo EGS4 user code MCDOSE for photon/electron beamlet and treatment plan dose calculations. Good agreement was achieved between the MCDOSE results and measurements.

*Evaluate dose calculation algorithms for optimization calculation:* We have compared the dose distributions calculated by the conventional dose calculation algorithms (the pencil beam algorithm) and the Monte Carlo method (with MCDOSE). It is concluded that Monte Carlo is needed for the electron beamlet distribution calculation to ensure the accuracy of the optimization process.

*Develop fast iterative optimization algorithms and evaluate different objective function for electron beam optimization:* A fast iterative optimization algorithm has been developed for electron beam optimization. The algorithm has been tested for both electron treatment planning optimization and photon beam optimization. We have installed several objective functions for the optimizer used for electron beam EIMRT. We have selected a dose based objective function with soft constraints using dose volume histogram information.

*Study the characteristics of electron beam modulation using magnetic fields:* We have studied the characteristics of small field electron beams modulated by an electron MLC and 0.5-3.0 T magnetic fields. We have studied beam optimization using these modulated electron beamlets.

*Verify electron beam EIMRT, photon IMRT and conventional treatments using EGS4/MCDOSE:* We have developed an EGS4 user code called MCDOSE to calculate dose distribution for electron treatment planning. We have compared the dose distributions calculated using MCDOSE and those by DOSXYZ and achieved good agreement (within 1% of maximum dose). We have used MCDOSE to calculate dose distribution for photon IMRT, electron EIMRT and conventional beam treatment planning. We have compared the dose distributions calculated using MCDOSE and measurements and achieved good agreement (within 3% of maximum dose).

## **Reportable Outcomes**

*Peer-reviewed papers resulting from or supported in part by this grant:*

- C.M. Ma, E. Mok, A. Kapur, T. Pawlicki, D. Findley, S. Brain, K. Forster and A.L. Boyer, Clinical implementation of a Monte Carlo treatment planning system, *Med. Phys.* 26: 2133-43 (1999)
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- Kapur and C.M. Ma, Stopping-powers for clinical electron beams for a scatter-foil linear accelerator, *Phys. Med. Biol.*, 45:2321-41 (1999)
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- Y. Chen, A.L. Boyer and C.M. Ma, Calculation of x-ray transmission through multileaf collimator, *Med. Phys.* (2000) 27:1717-26
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- C.M. Ma, T. Pawlicki, S.B. Jiang, E. Mok, A. Kapur, L. Xing, L. Ma and A.L. Boyer, Monte Carlo verification of IMRT dose distributions from a commercial treatment planning optimization system, *Phys. Med. Biol.*, (2000) 45:2483-2495
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- J. Deng, T. Pawlicki, Y. Chen, J. Li, S. B. Jiang and C.M. Ma, The MLC tongue-and-groove effect on IMRT dose distributions, *Phys. Med. Biol.* (2001) 46: 1039-1060
- S.B.Jiang, J. Deng, A. L. Boyer and C.M. Ma, An extrafocal source model for photon beam dose calculation, *Med. Phys.* (2001) 28: 55-66
- T. Pawlicki and C.M. Ma, Monte Carlo dose modeling for MLC-based IMRT, *Med. Dosimetry.* (2001) 26: 157-168
- J. Deng, S.B. Jiang, T. Pawlicki, J. Li and C.M. Ma, Derive the electron and photon energy spectra from the electron beam central axis depth dose curves, *Phys. Med. Biol.* (2001) 46: 1429-49

*Meeting abstracts resulting from or supported in part by this grant:*

- C.M. Ma, T. Pawlicki, S.B. Jiang, E.Mok, A. Kapur, L. Xing, L. Ma and A.L. Boyer, Verification of IMRT dose distributions using Monte Carlo simulations, ASTRO Annual Meeting (Phoenix, AZ, 1998)
- S.B.Jiang, A. Kapur and C.M. Ma, Electron beam modelling and commissioning for Monte Carlo treatment planning, AAPM Annual Meeting (Nashville, TN, 1999)
- S.B.Jiang, J. Deng, A. L. Boyer and C.M. Ma, An extrafocal source model for photon beam dose calculation, AAPM Annual Meeting (Nashville, TN, 1999)
- J. Deng, S.B.Jiang, J.S. Li, T. Pawlicki and C.M. Ma, Photon beam characterization and modeling for Monte Carlo treatment planning, AAPM Annual Meeting (Nashville, TN, 1999)
- C.M. Ma, S.B. Jiang, T. Pawlicki, S.B. Jiang, J.S. Li, J. Deng, E.Mok, A. Kapur, G. Luxton and A.L. Boyer, MCDOSE- a Monte Carlo dose calculation tool for radiotherapy treatment planning, AAPM Annual Meeting (Nashville, TN, 1999)
- C.M. Ma, S.B. Jiang, T. Pawlicki, S.B. Jiang, J.S. Li, J. Deng, E.Mok, A. Kapur, M. Lee, G. Luxton and A.L. Boyer, Energy- and intensity-modulated electron beams for treatment breast cancer, ASTRO Annual Meeting (San Antonio, TX, 1999)
- C.M. Ma, Monte Carlo as a QA tool for radiotherapy treatment planning, ESTRO Biannual Physics Meeting Symposium on treatment planning QA, (Göttingen, Germany, 1999)
- C.M. Ma, Monte Carlo treatment planning for electron beam radiotherapy, AAPM Annual Meeting Symposium on Monte Carlo treatment planning, (Nashville, TX, 1999)
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- C.M. Ma, T Pawlicki, MC Lee, SB Jiang, JS Li, J Deng, E Mok and AL Boyer, Modulated electron beams for treatment of breast cancer, Oral presentation at the XIII International Conference on the Use of Computer in Radiation Therapy (ICCR), Heidelberg, Germany, May 22 - 25, 2000.
- C.M. Ma, Monte Carlo methods in electron beams treatment planning, invited talk at the 41st Annual Meeting of the *American Association of Physicists in Medicine*, Nashville, TN, July 24-29, 1999. *Med. Phys.*, 26 (1999 AAPM Annual Meeting Program) 1999.
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- M.C. Lee, S.B. Jiang and C.M. Ma, Characteristics of clinical electron beams in transverse magnetic fields, 42st Annual Meeting of the American Society for Therapeutic Radiology and Oncology, Boston, MA, October 22-26, 2000.
- T. Pawlicki, S.B. Jiang, J. Deng, J. Li, and C.M. Ma, Compensating setup uncertainty and organ motion in IMRT treatment optimization, 42st Annual Meeting of the American Society for Therapeutic Radiology and Oncology, Boston, MA, October 22-26, 2000.
- J. Deng, S.B. Jiang, T. Pawlicki, J. Li, and C.M. Ma, Beam delivery for modulated electron radiation therapy, 42st Annual Meeting of the American Society for Therapeutic Radiology and Oncology, Boston, MA, October 22-26, 2000.
- C.M. Ma, SB Jiang, T Pawlicki, Y Chen, JS Li, J Deng, MC Lee, E Mok and AL Boyer, A QA phantom for IMRT dose verification, Chicago 2000 World Congress on Medical Physics and Biomedical Engineering, Chicago, Illinois, July 23-28, 2000.
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- S. B. Jiang, T. Pawlicki, and C.M. Ma, An aperture-based inverse planning algorithm for modulated electron radiation therapy, Chicago 2000 World Congress on Medical Physics and Biomedical Engineering, Chicago, Illinois, July 23-28, 2000.

*Funding applied for based on work resulting from or supported in part by this grant:*

1. US Army Breast Cancer Research Program: Postdoctoral Training Grant (PI: S.B. Jiang; Mentor: C.-M. Ma): Investigation of an electron MLC for energy and intensity modulated electron beam radiotherapy of breast cancer (awarded in 2000)
2. US Army Breast Cancer Research Program: Postdoctoral Training Grant (PI: T. Pawlicky; Mentor: C.-M. Ma): Effect of organ motion and patient setup uncertainty on the treatment of breast cancer with energy and intensity modulated electron beam radiotherapy (awarded in 2000)
3. US Army Breast Cancer Research Program: Predoctoral Training Grant (PI: Michael C. Lee; Mentor: C.-M. Ma): Beam verification for mudulated electron radiaiton therapy treatment of breast cancer (awarded in 2001)
4. NIH R01 (PI: C.-M. Ma): Modulated electron beams for radiotherapy (submitted in 2001)

## **Conclusions**

We have made significant progress during our three-year investigation. We have successfully performed the tasks scheduled in the "Statement of Work". We have simulated the electron beam characteristics for energy and intensity modulated radiotherapy (EIMRT) by simulating medical accelerators with different MLC leaf designs and filling gases (air or He). We have characterized our

electron beams using a well tested source model. We have implemented new dose calculation software for dose calculations to optimize treatment plans for EIMRT of breast cancer. The new software replaces the scheduled PEREGRINE software and provides not only the capability of performing photon dose calculation but electron dose calculations (the PEREGRINE system is designed only for photon dose calculations). The new software is also significantly (a factor of 20-30) faster than the PEREGRINE system. We have developed optimization algorithms for comparisons of the optimized treatment plans with photon IMRT or electron EIMRT. We have developed optimization algorithms for comparisons of the optimized treatment plans with photon IMRT or electron EIMRT. We have installed and investigated suitable objective functions for electron beam optimization. We have verified electron EIMRT and photon IMRT using the EGS4/MCDOSE system and compared dose distributions for photon IMRT, electron EIMRT and conventional treatment modalities. We also studied the effect of magnetic fields on modulation of electron beam profile and depth dose curves for EIMRT of breast cancer.

“So what?”

Our three-year results have provided evidence to support the hypothesis of this proposal that by using He-filled accelerators equipped with computer-controlled multi-leaf collimators and the Monte Carlo treatment planning technique, energy- and intensity-modulated electron beams may be optimized to significantly improve the dose uniformity in the target volume, to reduce the dose to the critical structures nearby and therefore reduce the late effects associated with breast cancer radiotherapy. Further studies are needed to investigate clinical implementation of the electron beam IMRT for breast and head and neck treatments. As a result, a NIH R01 proposal has been submitted to continue this research.

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## Appendices

### List of Figures quoted in the body of text:

Fig. 1 Isodose distributions for conventional photon beams (left), 4-field photon IMRT (middle) and 2-field IMRT tangents (right).

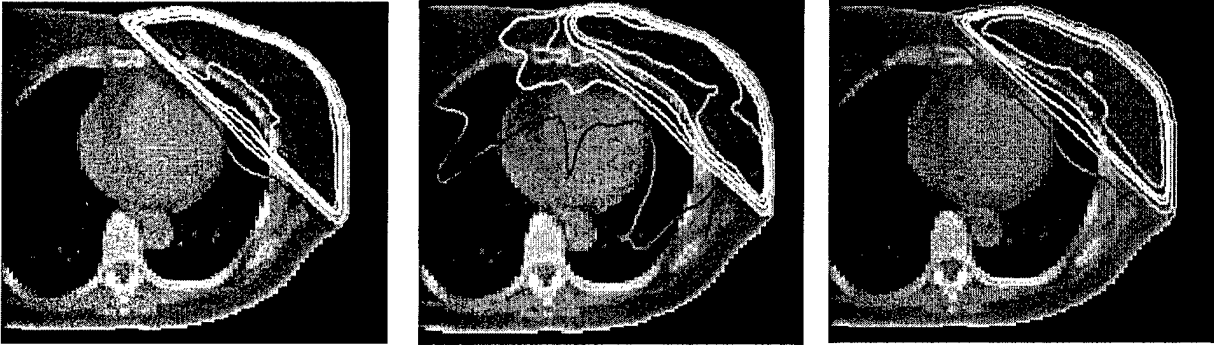


Fig. 2 DVHs for photon treatments.

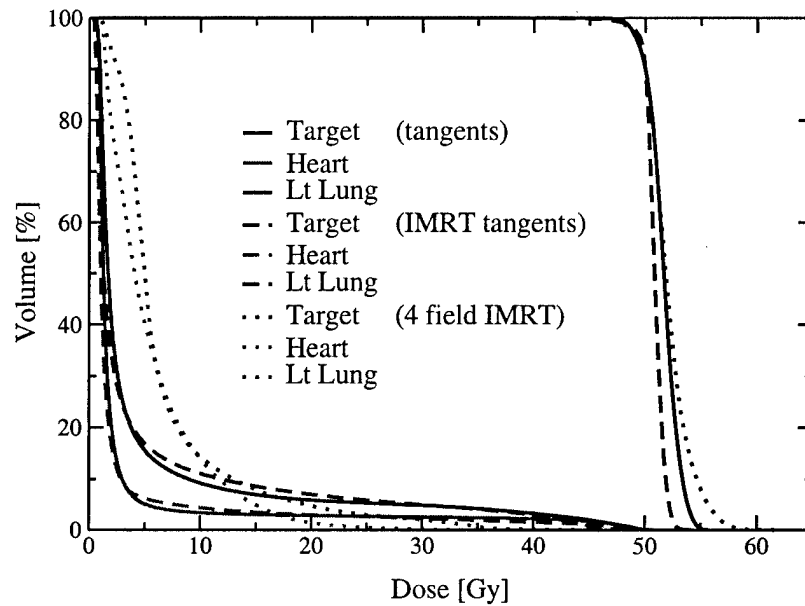


Fig. 3 Electron beams angles for a modulated electron breast plan.

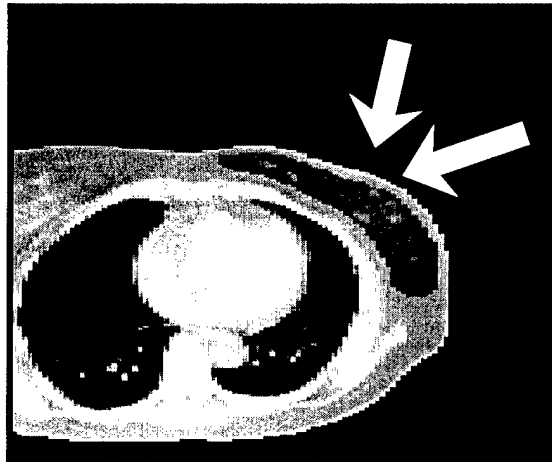


Figure 4 Electron beam intensity maps for the two gantry angles.

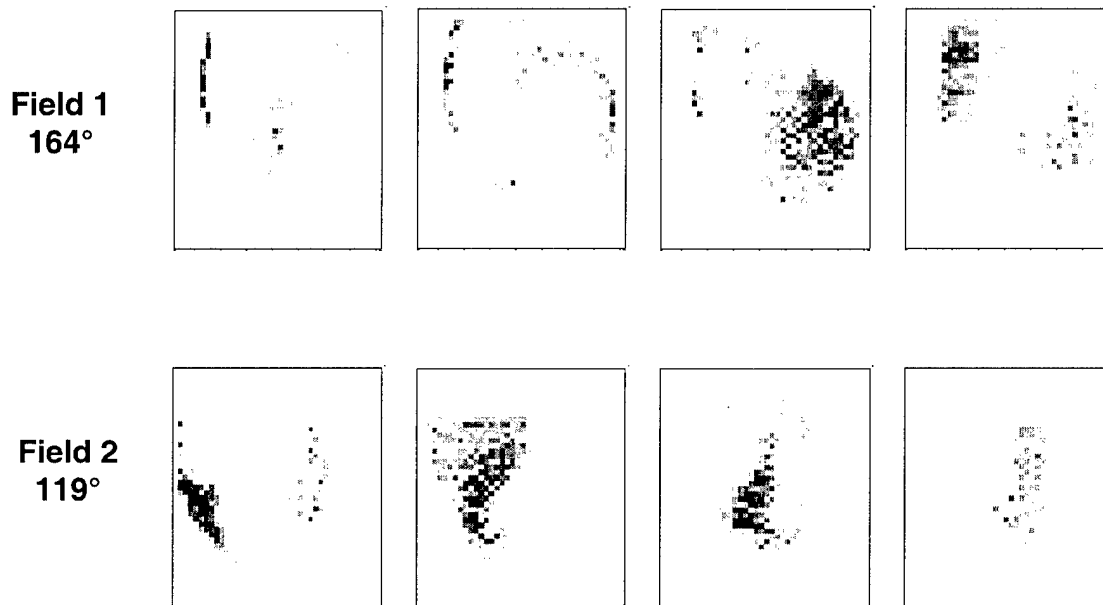


Fig. 5 Isodose distributions for the conventional photon beams (left), IMRT tangents (middle) and modulated electron beams (right).

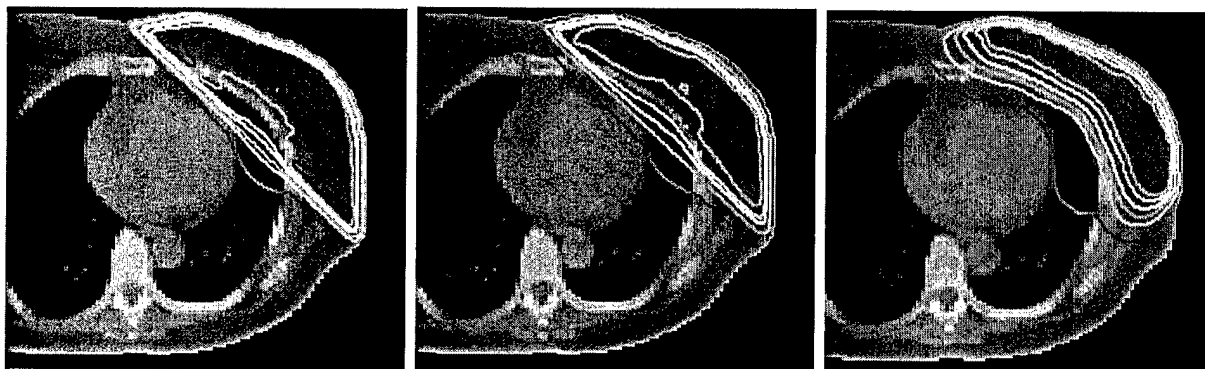


Fig. 6 DVH comparisons for the 3 different treatment techniques in Fig. 5.

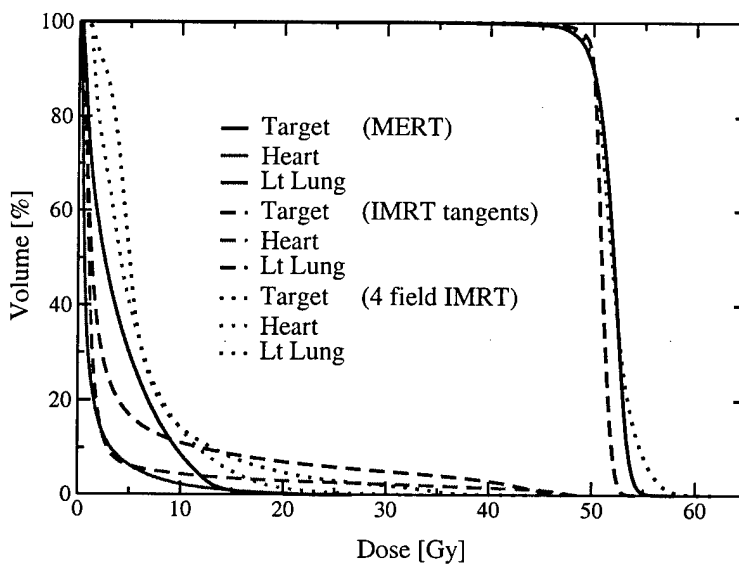


Fig. 7 DVH for the healthy tissues, defined as all non-target tissues in the body included in the calculation phantom.

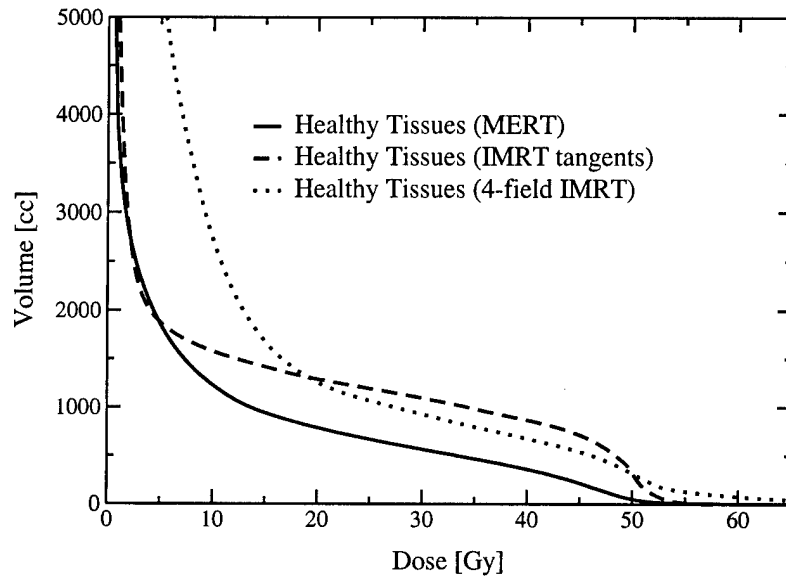


Fig. 8 DVH for the heart tissues.

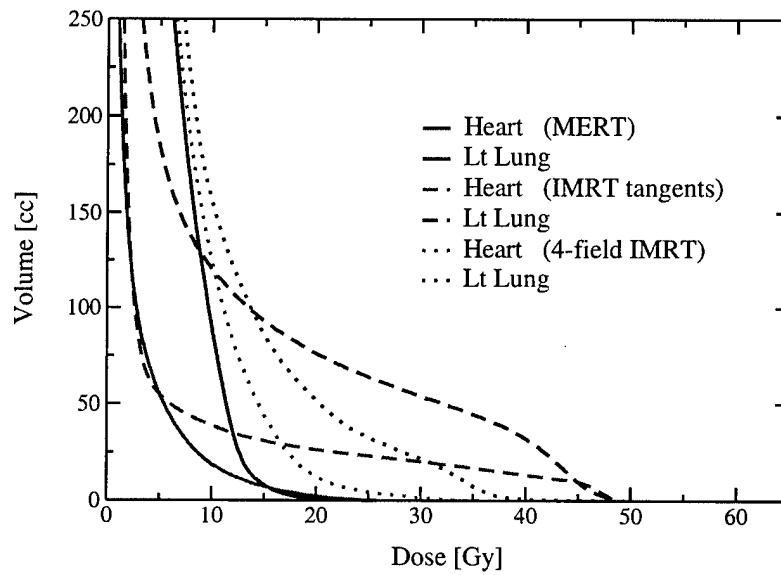
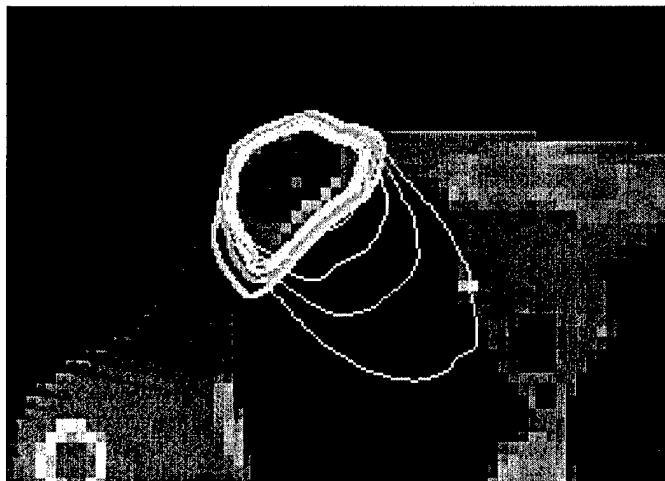
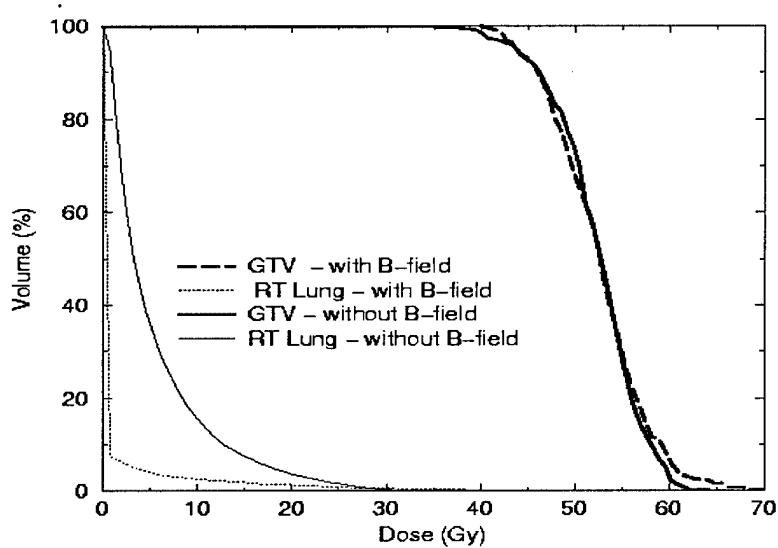


Figure 9 (a) Isodose curves for a breast treatment plan. Thick lines are optimized dose distributions for MERT with 1.5 T magnetic fields and thin lines are MERT without magnetic fields. The target coverage is similar but the dose to the lung behind the target is almost completely removed with the “soft collimation” using a magnetic field. (b) dose volume histograms for the two plans shown in (a).



(a)

#### DVH Data



(b)

Figure 10 Central-axis depth-dose curves demonstrating how linear combinations of 10 x10 cm<sup>2</sup> electron beams in magnetic fields, with different intensities and magnetic field placements, can be used to create homogeneous dose distributions. The dose curves were for the 1.5 T optimization.

