Published in final edited form as:

Health Phys. 2009 December ; 97(6): 581–589. doi:10.1097/01.HP.0000363838.70546.4c.

A STUDY OF THE SHIELDING USED TO REDUCE LEAKAGE AND SCATTERED RADIATION TO THE FETUS IN A PREGNANT PATIENT TREATED WITH A 6-MV EXTERNAL X-RAY BEAM

Bin Han^{*}, Bryan Bednarz[†], and X. George Xu^{*}

^{*}Nuclear Engineering and Engineering Physics, Rensselaer Polytechnic Institute, Troy, NY 12180

[†]Department of Radiation Oncology, Massachusetts General Hospital and Harvard Medical School, Boston, MA

Abstract

A Monte Carlo-based procedure has been developed to assess the shielded fetal doses from 6 MV external photon beam radiation treatments and improve upon existing techniques that are based on AAPM Task Group Report 36 (TG-36). Anatomically realistic models of the pregnant patient representing 3- and 6-mo gestational stages were implemented into the MCNPX code together with a detailed accelerator model that is capable of simulating scattered and leakage radiation from the accelerator head. The phantom was shielded using suggested lead and Cerrobend in different locations and with different thicknesses. Absorbed doses to the fetus both with and without shielding were calculated considering typical mantle, head and neck, and brain treatment plans. The unshielded fetal doses tended to increase with decreasing distance from the field edge to the nearest fetal point and increasing of the field size. The unshielded absorbed doses to the fetus for all treatment plans ranged from a maximum of 4.08 $\mu\text{Gy}/\text{MU}$ (monitor unit) to a minimum 0.09 $\mu\text{Gy}/\text{MU}$. The use of lead or Cerrobend shielding reduced the fetal doses by factors of up to 4. For an optimal shield half-value layer, the dose reduction between lead and Cerrobend was statistically insignificant. The maximum permitted MUs for the mantle treatments with shielding were calculated based on 5 cGy dose limits suggested by TG-36. The study demonstrates an accurate assessing tool that can be used to determine the absorbed dose to the fetus and to design the shielding as part of the treatment planning and risk management.

Keywords

dosimetry; Monte Carlo; pregnancy; shielding

INTRODUCTION

EACH YEAR thousands of women undergo radiation treatments during pregnancy for tumors in a variety of sites including the brain (Magne et al. 2001; Mazonakis et al. 1999), breast (Antypas et al. 1998), nasopharynx (Yan et al. 1984), and knee (Nair et al. 1983), as well as for Hodgkin's disease (Cygler et al. 1997; Woo et al. 1992). Radiation safety of the fetus is of particular concern because of the high levels of radiosensitivity and secondary exposure in these procedures. Treatment planning for pregnant patients must find a delicate balance

between the therapeutic outcome to the mother and the allowable risk to the fetus from radiation exposure.

Several simple techniques have been used to reduce the dose outside the treatment field by adjusting the gantry angle, field size, and patient position (Stovall et al. 1995; Magne et al. 2001). In addition, defining the field edge that is located nearest to the fetus using the lower collimator components, such as the jaws and multileaf collimator (MLC), have been found to reduce the out-of-field dose from the collimator scattering radiation (Stovall et al. 1995; Mutic and Klein 1999). However, the most effective technique to reduce dose to the fetus from radiation treatments is through the use of shielding structures (Stovall et al. 1995; Mazonakis et al. 1999; Antypas et al. 1998; Yan et al. 1984). A factor of two or more in fetal dose reduction can be achieved with shielding structures placed over the fetus (Antolak and Storm 1998; Roy and Sandison 2000; Islam et al. 2001).

To reduce potential risk to the mother and fetus, the medical staff typically develops special treatment plans as discussed by the American Association of Physicists in Medicine (AAPM) Task Group Report 36 (TG-36) (Stovall et al. 1995). The following general procedure is considered: (1) Develop the treatment plan for the known tumor location, (2) based on this treatment plan, estimate unshielded fetal dose using TG-36 data, (3) measure unshielded fetal dose using physical phantoms, and (4) measure shielded fetal dose from the treatment using physical phantoms. By nature, this procedure involves time-consuming and laborious measurements that may underestimate or overestimate the true dose to the fetus (Kry et al. 2006). As the number of pregnant patients continues to increase owing to improvements in cancer detection and the tendency for women to delay their pregnancy until later reproductive ages, it is necessary to develop faster and more accurate tools to improve the treatment planning of pregnant patients.

Recently, Bednarz and Xu (2008) demonstrated the feasibility of combining a detailed medical accelerator model and anatomically realistic patient phantoms to calculate the unshielded fetal dose to pregnant patients undergoing radiation treatments. The new approach described in their paper utilized a set of newly developed anatomical models that are realistic and compatible with the International Commission on Radiological Protection (ICRP) reference values for average pregnant females at the end of 3-, 6-, and 9-mo gestational periods (Xu et al. 2007). Coupled with detailed accelerator models these new computational tools offer an opportunity to quickly and accurately estimate doses to the fetus as part of the treatment planning procedure. In particular, the paper demonstrated the advantages of using Monte Carlo techniques to estimate the unshielded fetal dose from radiation treatments of pregnant patients compared to measurement techniques discussed in the AAPM TG-36 report. However, this previous study did not assess the reduced fetal dose after shielding is used as part of the clinical procedures (Mazonakis et al. 1999; Antypas et al. 1998; Yan et al. 1984).

In this paper, the computational tools developed by Bednarz and Xu (2008) are extended by adding new features to simulate a variety of shielding designs used to reduce the leakage and scatter radiation. In particular, shielding design options recommended by the AAPM TG-36 report are considered in this paper with different treatment plans using Monte Carlo methods. The impact of the shielding material, thickness, shape, and location on the assessed fetal dose is analyzed.

METHODS

A medical accelerator used for external-beam radiation treatment contains internal shielding around the beam-line components such as the target, collimators and jaws. However, a

significant amount of scattered photons and, when operated at a high enough energy, neutrons will escape the collimators and shielding inside the accelerator and deposit doses to various parts of the patient's body. In addition to the leakage radiation from the accelerator head, scattering inside a patient can also lead to irradiation of organs away from the treatment volume. Therefore, to estimate the organ and fetal doses using Monte Carlo methods, detailed models of the patient and the accelerator are needed.

Pregnant patient models

The use of a whole-body computational phantom is required to calculate the secondary radiation doses to various organs in a pregnant patient. This study utilized a set of pregnant patient computational phantoms previously developed by Xu et al. (2007) using a novel boundary representation method. These pregnant patient models are represented in an organ-based surface geometric domain involving a mixture of data structures represented by voxels, meshes, and non-uniform rational B-splines (NURBS) (Xu et al. 2007). Using this novel approach, Xu et al. (2007) assembled individual organs and the fetus in the body of a non-pregnant female to form a pregnant individual. In addition, the organ volumes (and masses) were easily adjusted to match with the recommended anatomical parameters for an average pregnant female at the end of 3-, 6-, and 9-mo gestational stages, respectively (ICRP 2002). To define the patient geometries in a Monte Carlo code for dose calculations, a voxelization tool was developed using VC++ to convert the mesh-based geometries to voxel-based phantoms with the voxel size to be specified by a user. The in-house voxelization software was based on parity-counting and ray-stabbing methods (Nooruddin and Turk 2003) for polygonal surfaces that are closed-meshes. The voxelized organs were assembled into a whole-body model without overlapping. A total of 35 organs, including fetal organs, were adjusted to agree with ICRP reference organ masses and sizes representing a typical pregnant patient. Fig. 1 shows the RPI-P6 model along with the accelerator model plotted in MCNPX.

Accelerator modeling

To calculate out-of-field dose from secondary radiation, a detailed model of a medical accelerator including primary beam-line components as well as surrounded shielding components is needed. A detailed representation of a Varian Clinac 2100C (Varian Medical Systems Inc., Palo Alto, CA) accelerator head was developed for Monte Carlo calculations of radiation dose to pregnant patients (Bednarz and Xu 2008). The primary components such as the target, primary collimator, flattening filter, and jaws were based on blueprints provided by the manufacturer, while additional components of the accelerator head were taken from the combinational geometry input of the same accelerator type published by Kase et al. (1998). These components are important when considering the leakage radiation that emanates from the accelerator. A set of 80 detailed modeled MLC leaves based on blueprints from the manufacturer was also included in the model. All calculations in this work used a 6-MV photon beam to treat the patient. The 6-MV beam was created by an electron beam with a 1.3-mm full width half maximum (FWHM) Gaussian spatial spread and a mean energy of 6.2 MeV with a Gaussian spread of 3% FWHM. The accelerator model was well benchmarked with previously measured in-field and out-of-field data (Bednarz and Xu 2008). Fig. 1 illustrates the detail modeled accelerator plotted in MCNPX.

Monte Carlo calculation procedures

The Monte Carlo code package, MCNPX, a general purpose transport code from Los Alamos National Laboratory, was used to simulate the transport and energy deposition of photons, electrons and neutrons in the patient phantoms (Pelowitz 2005). The absorbed dose to the fetus was calculated using the track-length cell energy deposition tally ($F6:p$). This tally theoretically provides the collision kerma throughout the fetus. The collision kerma is

defined as the energy departed in a volume whereas the dose is the energy deposited in a volume. However, when the rate of energy transfer into and out of a volume is constant, the conditions of electronic equilibrium are met and the kerma theoretically provides the dose to the volume. It has been shown that during radiation treatments at depths below 2 cm, the conditions of electronic equilibrium are satisfied and the dose and kerma are nearly the same (Kry et al. 2006). To test this assumption, we compared the collision kerma to the absorbed dose using the pulse height tally modified to provide energy deposition divided by the fetal mass (*F8: e). For each treatment plan, the collision kerma was well within the statistical uncertainty range of the absorbed dose. The advantage of calculating the collision kerma is that with the same initial simulation number, the relative error is 2 to 3 times lower than the absorbed dose, which is due to the difference in how MCNPX records the photon track length energy deposition tally compared to the modified pulse height tally. The track length energy deposition tally uses a track length estimator to calculate the energy deposition in a cell. The modified pulse height tallies are made at source points and at surface crossings. Consequently, for small tally volumes, nonzero scores occur much less frequently using the modified pulse height tally than the track length tally, since the probability of interaction in the tally volume is low. For all simulations, electrons were tracked with the Integrated Tiger Series energy indexing, which has been shown to provide more accurate results than the default indexing method (Jeraj et al. 1999). The values of the photon and electron cut-off energies were set to 0.01 and 0.1 MeV, respectively.

The F6 tally result provided by MCNPX is normalized per source history. In order to determine the absolute fetal dose from each treatment plan, the dose per source electron in MCNPX was converted to dose per monitor unit (MU) in the following manner: A water tank was modeled at 100 cm source-to-surface distance, and the number of electrons needed to deliver 1 MU (or 1 cGy at $d_{\max}$) in the tank under reference conditions, i.e., a 10 cm $\times$ 10 cm field, was determined. This number was kept constant for all field sizes. Keeping this value constant for all field sizes introduces errors, since the dose at $d_{\max}$ changes with different field sizes. However, as pointed out by Kry et al. (2007) the dependence of the dose at $d_{\max}$ on field size is rather insignificant when calculating dose outside the treatment field.

Treatment and shielding setup

The fetal dose outside of the treatment field was calculated as a function of primary beam energy, prescribed dose (or delivered MU), treatment center location, distance from the field edge to the nearest fetal point, field size, gestational stage (or fetus size), and the thickness of shielding. To evaluate the influence of these parameters on the fetal dose, we adopted typical 6-MV treatment plans involving brain, head and neck, and mantle treatment fields with different field sizes which are commonly used to treat pregnant patients (Magne et al. 2001; Mazonakis et al. 1999; Antypas et al. 1998; Yan et al. 1984; Woo et al. 1992; Cygler et al. 1997). Table 1 provides additional information on the treatment plans including the treatment field, field size, distance from field center to the nearest fetal point (DCF), and distance from the field edge to the nearest fetal point (DEF). *Plan 1* through *plan 4* are 10 cm $\times$ 10 cm treatment plans for the P-3 phantom with decreasing DCF. *Plan 5* through *plan 8* are 10 cm $\times$ 10 cm treatment plans for the P-6 phantom with decreasing DCF. *Plan 9* and *plan 10* are mantle treatment plans with 20 cm $\times$ 20 cm field. *Plan 11* and *plan 12* are head and neck treatment plans for the P-3 and P-6 phantoms, respectively. *Plan 13* and *plan 14* are brain treatment plans for the P-3 and P-6 phantoms, respectively. An example of a typical treatment setup with shielding in MCNPX is provided in Fig. 1. We used jaws as well as MLC to define the field edge, which was found to reduce the dose outside the field (Mutic and Klein 1999).

The TG-36 report provides guidelines to help physicists develop safe treatment plans for pregnant patients. The report recommends that the fetal dose be estimated by using an ionization chamber to take measurements at selected points that will reflect the range of doses throughout the fetus. Once the dose is measured and recorded, shielding should be designed to lower the fetal dose below the recommended dose range of 5–10 cGy. This range of dose limits was determined by TG-36 after a careful evaluation of the epidemiological data of risks to the fetus after radiation exposure. The maximum unshielded fetal dose reported by Kry et al. (2006) was 47 cGy (1.5% of the prescription dose), which was higher than the dose limit. To reduce the potential risk to the fetus, shielding of the head leakage radiation is needed. As suggested by the TG-36 report, we used two types of shielding materials, lead and Cerrobend (a eutectic alloy of 50% bismuth, 26.7% lead, 13.3% tin, and 10% cadmium by weight). The dose to the fetus is typically from three major components: (1) scatter from the useful beam within the patient (labeled as “1” in Fig. 1), (2) scatter off the collimator (labeled as “2” in Fig. 1), and (3) leakage from the accelerator head (labeled as “3” in Fig. 1). With increasing DEF, the dose contribution from head leakage increases while the dose from collimator and patient scatter decrease. The TG-36 report suggests a shielding design, which is shown in Fig. 1, that consists of a bridge over the patient's abdomen that supports four to five half-value layers (HVLs) of metal (approximately 5 to 7 cm of lead or 6 to 8.5 cm of Cerrobend).

In the mantle treatment plans, the superior edge of the lead was aligned with the inferior edge of the field (defined by MLCs and jaws) to decrease head leakage and collimator scattering. For the treatments of the brain or head and neck, the superior edge of the lead was placed 15 cm higher than the superior edge of the fetus. The inferior edge of the shielding plate was aligned with the inferior edge of the fetus and the shield was set wide enough to cover the whole pregnant phantom from left to right. The anterior distance from the shielding plate to abdomen surface will influence the shielding effect because the closer the shielding structure is to the patient, the larger amount of collimator scatter can be shielded. Therefore, this anterior distance was set to a range of 1 to 9 cm in order to study its relationship with the fetal dose. The energy spectra in different out-of-field locations were different, and that will result in the change of shielding effects for different materials with the same HVL thickness. We simulated several treatment plans with the same HVLs of lead and Cerrobend shielding, and the fetal dose variations were within 2%. As a result, we assumed the shielding effects of the two different materials were indistinguishable, thus we present only the fetal dose calculations with lead shielding.

Following the treatment plans provided above, the absorbed doses to the fetus with and without shielding were calculated for the 3- and 6-mo pregnant patient models. The lead shielding plates with thicknesses ranging from 1 to 9 cm were considered. All calculations were performed with at least 1×10^8 initial electron histories in order to achieve reasonable statistics in the tally results. More initial electron histories were needed for the RPI-P3 phantom, which has a smaller fetus than the RPI-P6 phantoms.

RESULTS AND DISCUSSION

In the following discussion we present calculated fetal dose values for several treatment plans of a pregnant patient with and without shielding. The dependence of fetal dose on several different treatment parameters is investigated and discussed. For calculations involving shielding, the anterior distance between the shielding structure and the abdomen of the pregnant patient was set to 5 cm, a place close to the patient's body that does not compromise the ease of patient positioning. It will be shown in a subsequent section that the fetal dose dependence on anterior distance is small. This study assumed that the orientation of the fetus inside the mother is fixed and known. Furthermore, all treatment plans were

assumed to be in the anterior-posterior (AP) gantry angle, which is the most practical and common field orientation for treating pregnant patients using shielding.

Comparison of shielded and unshielded fetal dose for selected plans

The calculated absorbed dose to the fetus with and without 7-cm shielding for two pregnant patient models (3- and 6-mo) using all 10 cm × 10 cm field treatment plans are plotted in Fig. 2 with increasing DEF. The relative statistical errors of the fetal doses for the mantle treatment plans were below 5% while those for the brain as well as head and neck treatment plans were below 10%. For the same number of histories, the relative error for all treatment plans improved with increasing stage of gestation, as more particles interacted inside the increasing fetal volume. The fetal dose decreased approximately exponentially with increasing DEF, which agrees well with previously measured and calculated out-of-field dose distributions (Kry et al. 2006). The leveling off of the fetal dose at longer distances is due to the increasing contribution of leakage radiation from the accelerator head at these distances. The unshielded fetal doses ranged from a maximum of 3.63 $\mu\text{Gy}/\text{MU}$ to a minimum 0.09 $\mu\text{Gy}/\text{MU}$. In other words for typical 7,000 MU treatment plans, the absolute absorbed dose of the fetus ranged from a maximum of 25.4 cGy to a minimum 0.63 cGy. Shielding with a 7-cm lead plate, the fetal doses decreased to a maximum of 2.30 $\mu\text{Gy}/\text{MU}$ and a minimum 0.02 $\mu\text{Gy}/\text{MU}$.

Next, we introduce the dose reduction ratio (DRR) to represent the effect of shielding thickness on the fetal dose. The DRR was calculated as follows:

$$DRR = (D_n - D_s) / D_n, \quad (1)$$

where D_s and D_n represent fetal dose with shielding and without shielding, respectively. Fig. 3 plots the fetal DRRs for all the treatment plans listed in Table 1 as a function of lead shielding thickness. The thickness ranged from 1 to 9 cm. The relative statistical errors of the fetal doses for mantle treatment plans were below 5%, while for the brain as well as head and neck treatment plans the errors were below 10%. The DRRs increased with increasing shielding thickness, since more accelerator head leakage and collimator scattered photons were attenuated by the shield. When shielding thickness increased from 0 to 3 cm, the DRRs increased rapidly. This is due to the fact that the predominate source of fetal dose at this thickness is from head leakage and collimator scatter. At a certain shield thickness, depending on the plan, the dose reduction curve reaches a plateau. This is because at the plateau, head leakage and collimator scatter photons are predominantly being shielded and patient scatter becomes the main source of fetal dose. Between 3-cm to 9-cm thickness, the variation in dose reduction is on average 2% per cm. The height of the dose reduction curve is primarily dependent on the distance between the field edge and nearest fetal point for a given field. For example, for the mantle treatment plans the fetal DRR ranged from 0.3 to 0.72, while for the brain and head and neck treatment plans the fetal DRRs ranged from 0.6 to 0.75. This is due to the increase in DEF. As the DEF increases, the dose from head leakage contributes more to the total fetal dose, and only that part of dose can be shielded. The DRR of the head and neck plans and the brain plans could be lower than several mantle plans because of the location of the shielding plates. For mantle plans, the superior edges of the shielding plates were aligned to fit the radiation field edge so that more head leakage radiation and collimator scattering radiation could be shielded.

It is interesting to note that the TG-36 report suggested shielding thickness (5-cm to 7-cm lead plate as shown in Fig. 3) seems to be too conservative for some of the treatment plans. The DRR presented in this paper from Monte Carlo simulations can be used to make sure the shielding used is not more than necessary. Shielding materials can be expensive, so using more shielding than is necessary is not cost effective. Also, using excessive shielding

increases the danger to the patient and hospital staff since the shields will be heavier and more difficult to maneuver.

The effect of distance from the field edge and treatment center to the nearest fetal point on fetal dose

Fig. 4 plots the fetal doses from treatment plans with different gestational stage and treatment fields with the same treatment center, DEF, and DCF, respectively. As seen in Fig. 4, with the same treatment center (tumor location), the fetal dose increases with gestational stage. This behavior is primarily due to the decrease in the distance between the fetus and the treatment field edge as the fetus increases in size. Therefore, the 6-mo fetus, which was closer to the field edge, received a higher dose than the 3-mo fetus. With a constant treatment center and the same gestational stage, the larger field size contributed a higher dose to the fetus, which was due to the decrease in the distance from the fetus to field edge. Similarly, the fetal dose increases with the field size when keeping the DCF and the fetus size constant because of the decreasing distance from the fetus to the field edge.

In addition, when keeping the DEF constant, the treatment plan with a larger field size delivered a higher dose to the fetus. This can be attributed to the increase in patient and collimator scatter for the larger field size. With the same DEF, the P-3 fetal dose was higher than the P-6 fetal dose which was mainly due to three reasons: (1) During fetal development, the volume of soft tissue in the abdominal region between the treatment field and fetus increased and shielded more of the patient body-scattered radiation, (2) the average distance from the P-6 fetus to the field edge was larger than the P-3 fetus, and (3) the mass of the P-6 fetus was larger than the P-3 fetus.

The effect of the anterior distance between the patient and shielding on fetal dose

In this section the dependency of fetal dose on the anterior distance between the pregnant patient and shielding structure is investigated. This effect is shown in Fig. 5 where the fetal dose is plotted as a function of anterior distance from the shield to the abdominal surface of the patient. The fetal doses were calculated using three selected treatment plans: a mantle treatment described as *plan 3*, a head and neck treatment described as *plan 11*, and a brain treatment described as *plan 13* in Table 1. For all plans the same shielding thickness of 7-cm lead was used. The anterior distance will only influence the shielding effect of the dose from collimator scatter (labeled “2” in Fig. 1). As the anterior distance from the shielding to the phantom body decreases, more collimator scatter radiation can be shielded, thus the fetus receives a lower dose. Fig. 5 also shows how the anterior distance depends on the treatment site. The DEF increases from 9 cm for the mantle field to 56 cm for the brain treatment. With increasing DEF, the fetal dose from collimator-scattered radiation decreases. Therefore, the relationship of fetal dose to the anterior distance is different for different treatment fields. For mantle treatment *plan 3*, a 4-cm change in the anterior distance from the pregnant patient to the shielding plate resulted in the fetal dose changing 3%. For head and neck treatment *plan 11* and brain treatment *plan 13*, a 4-cm change in the anterior distance resulted in the fetal doses increasing 2% and 1%, respectively. Overall, the influence of the anterior distance to the fetal dose is small, so in the remaining calculations we set the distance to 5 cm, which is not too far from the patient body and makes clinical positioning easier.

Investigation of fetal dose as a function of prescribed MU

The amount of prescribed MUs per treatment for the plans used in this study was typically lower than 20,000 MU. For the brain and head and neck treatment plan, the unshielded fetus would receive a dose of less than 5 cGy. However, this study also found that the use of 3-cm lead or 4-cm Cerrobend shielding can most effectively reduce the dose to the fetus by factors

of 4 and 3 for the brain and head and neck treatment plans, respectively. For the mantle treatment plans, on the other hand, the unshielded fetal dose was found to be up to 4.08 $\mu\text{Gy}/\text{MU}$ or 28.6 cGy for an arbitrary but clinically relevant 7,000 MU treatment plan, thus additional shielding was required. The absorbed doses to the fetus of all 10 cm $\times$ 10 cm mantle field treatment plans in Table 1 with shielding of 3- and 7-cm lead plates were calculated. Based on the TG-36 suggested 5-cGy dose limits, the maximum permitted MU for each treatment plan was calculated and plotted in Fig. 6. For the treatment plan of the P-3 phantom with 9-cm DEF, the maximum prescribed MUs are 2,027 MU and 2,173 MU for 3- and 7-cm lead plates, respectively. For the treatment plan of the P-6 phantom with 9-cm DEF, the maximum prescribed MUs are 6,112 MU and 6,358 MU for 3- and 7-cm lead shielding plates, respectively.

CONCLUSION

This study has developed a Monte Carlo-based computational procedure to calculate both unshielded and shielded fetal absorbed doses for an average pregnant patient at the end of 3-mo and 6-mo gestational periods. The approach demonstrates an accurate yet more flexible fetal dose assessment method than that suggested by AAPM TG-36 (Stovall et al. 1995). In addition, newly developed pregnant patient models were used in this study to provide better anatomical details that afforded a more realistic dose assessment. This study suggests that the absorbed dose averaged over the entire fetal body, which can vary in size and shape during the gestational period, offers a more accurate estimate than the point-wise measurements in a water phantom suggested by the AAPM TG-36. Both models of the patient and accelerator described in this paper can be readily adopted by others in routine treatment planning and risk communication. The unique contribution of this study includes a thorough investigation of shielding designs of 14 treatment plans for sites covering the mantle, head and neck, and brain. This study found that, for common head and neck treatment plans and brain treatment plans involving 20,000 MU or less, the unshielded fetus would receive a dose of less than 5 cGy. However, this study also discovered that the use of 3-cm lead or 4-cm Cerrobend shielding can most effectively reduce the fetal dose by a factor of 3 for these treatment plans. For mantle treatment plans, on the other hand, the unshielded fetal doses were found to be up to 4.08 $\mu\text{Gy}/\text{MU}$ or 28.6 cGy for typical 7,000 MU treatment plans and the use of 3- to 7-cm-thick lead or 4- to 8.5-cm-thick Cerrobend was required to reduce fetal doses by a factor of up to 4 depending on the location of the fetus. This study also revealed that, for a treatment plan in which the field edge is 15 cm or closer to the nearest fetal point where the body scattering dominates, the shielding is not as effective and the total MU should be limited in order to reduce the overall exposure to the fetus.

It is clear that the radiation treatment of a pregnant patient is a complex task requiring the assessment of both unshielded and shielded fetal doses. The data and methodology provided in this paper can facilitate the planning process by considering such detailed information as specific gestational stage, shielding, field size, and distance from field edge to the nearest fetal point. It is hoped that the accurate radiation dose estimates and shielding design for pregnant patients and fetuses provide an opportunity to improve the effectiveness of radiation treatments for pregnant patients while reducing the risk to the fetus.

Acknowledgments

The project was supported in part by a grant from the National Cancer Institute (R01CA116743). Bin Han was supported by the Van Auken Research Fellowship from Rensselaer Polytechnic Institute as well as the Robert Gardner Fellowship from the Health Physics Society.

REFERENCES

- Antolak JA, Storm EA. Fetal dose estimates for electron-beam treatment to the chest wall of a pregnant patient. *Med Phys*. 1998; 25:2388–2391. [PubMed: 9874832]
- Antypas C, Sandilos P, Kouvaris J, Balafouta E, Karinou E, Kollaros N, Vlahos L. Fetal dose evaluation during breast cancer radiotherapy. *Int J Radiat Oncol Biol Phys*. 1998; 40:995–999. [PubMed: 9531386]
- Bednarz X, Xu XG. A feasibility study to calculate unshielded fetal doses to pregnant patients in 6-MV photon treatments using Monte Carlo methods and anatomically realistic phantoms. *Med Phys*. 2008; 35:3054–3061. [PubMed: 18697528]
- Cygler J, Ding GX, Kendal W, Cross P. Fetal dose for a patient undergoing mantle field irradiation for Hodgkin's disease. *Med Dosim*. 1997; 22:135–137. [PubMed: 9243468]
- International Commission on Radiological Protection. Basic anatomical and physiological data for use in radiological protection: reference values. Elsevier; New York: 2002. ICRP Publication 89
- Islam MK, Saeedi F, Al-Rajhi N. A simplified shielding approach for limiting fetal dose during radiation therapy of pregnant patients. *Int J Radiat Oncol Biol Phys*. 2001; 49:1469–1473. [PubMed: 11286855]
- Jeraj R, Keall P, Ostwald P. Comparison between MCNP, EGS4, and experiment for clinical electron beams. *Phys Med Biol*. 1999; 44:705–717. [PubMed: 10211804]
- Kase KR, Mao XS, Nelson WR, Liu JC, Kleck JH, Elsalim M. Neutron fluence and energy spectra around the Varian clinac 2100C/2300C medical accelerator. *Health Phys*. 1998; 74:38–47. [PubMed: 9415580]
- Kry SF, Titt U, Pönisch F, Followill D, Vassiliev ON, White RA, Mohan R, Salehpour M. A Monte Carlo model for calculating out-of-field dose from a Varian 6 MV beam. *Med Phys*. 2006; 33:4405–4413. [PubMed: 17153419]
- Kry SF, Starkschall G, Antolak JA, Salehpour M. Evaluation of the accuracy of fetal dose estimates using TG-36 data. *Med Phys*. 2007; 34:1193–1197. [PubMed: 17500450]
- Magne N, Marcie S, Pignol J, Casagrande F, Lagrange J. Radiotherapy for a solitary brain metastasis during pregnancy: a method for reducing fetal dose. *Br J Radiol*. 2001; 74:638–641. [PubMed: 11509400]
- Mazonakis M, Damilakis J, Theoharopoulos N, Varberis H, Gourtsoyiannis N. Brain radiotherapy during pregnancy: an analysis of conceptus dose using anthropomorphic phantoms. *Br J Radiol*. 1999; 72:274–278. [PubMed: 10396218]
- Mutic S, Klein E. A reduction in the AAPM TG-36 reported peripheral dose distributions with tertiary multileaf collimators. *Int J Radiat Oncol Biol Phys*. 1999; 44:947–953. [PubMed: 10386654]
- Nair RP, Nair TKM, El-Akkad S. Evaluation of fetal dose from megavoltage irradiation of the knee and neonate followup. *Med Phys*. 1983; 10:862–865. [PubMed: 6419033]
- Nooruddin F, Turk G. Simplification and repair of polygonal models using volumetric techniques. *IEEE Trans Vis Comput Graph*. 2003; 9:191–205.
- Pelowitz, DB. MCNPX user's manual. version 2.5.0. Los Alamos National Laboratory; Los Alamos, NM: 2005. Report No. LA-UR-02-2607
- Roy C, Sandison GA. Shielding for neutron scattered dose to the fetus in patients treated with 18 MV x-ray beams. *Med Phys*. 2000; 27:1800–1803. [PubMed: 10984226]
- Stovall M, Blackwell CR, Cundiff J, Novack DH, Palta JR, Wagner LK, Webster EW, Shalek RJ. Fetal dose from radiotherapy with photon beams: Report of AAPM Radiation Therapy Committee Task Group No. 36. *Med Phys*. 1995; 22:63–82. [PubMed: 7715571]
- Woo SY, Fuller LM, Cundiff JH, Bondy ML, Hagemeister FB, McLaughlin P, Velasquez WS, Swan F, Rodriguez A, Cabanillas F, Allen PK, Carpentier RJ. Radiotherapy during pregnancy for clinical stages of IA-IIA Hodgkin's disease. *Int J Radiat Oncol Biol Phys*. 1992; 23:407–412. [PubMed: 1587764]
- Xu XG, Chao TC, Bozkurt A. VIP-MAN: an image-based whole-body adult male model constructed from color photographs of the visible human project for multi-particle Monte Carlo calculations. *Health Phys*. 2000; 78:476–486. [PubMed: 10772019]

- Xu XG, Taranenko V, Zhang J, Shi C. A boundary representation method for designing whole-body radiation dosimetry models: pregnant females at the ends of three gestational periods—RPI-P3, -P6, and -P9. *Phys Med Biol.* 2007; 52:7023–7044. [PubMed: 18029991]
- Yan JH, Liao CS, Hu YH. Pregnancy and nasopharyngeal carcinoma: a prognostic evaluation of 27 patients. *Int J Radiat Oncol Biol Phys.* 1984; 10:851–855. [PubMed: 6735768]

Fig. 1. Plot of a typical treatment setup for the pregnant patient (RPI-P6) with shielding and accelerator model in MCNPX. The thickness and location of the shielding plate are varying to estimate their impact to the shielding effect. Three major sources of the fetal dose are labeled in the figure: (1) patient body scatter; (2) collimator scatter; and (3) head leakage.

Fig. 2. Calculated shielded and unshielded fetal absorbed dose per MU of 10 cm $\times$ 10 cm field treatment plans for 3- and 6-mo fetuses. The unshielded fetal doses decreased approximately exponentially with the increasing distance from field edge to the nearest fetal point from a maximum of 3.63 $\mu\text{Gy/MU}$ to a minimum 0.09 $\mu\text{Gy/MU}$. The shielded fetal doses decreased to a maximum of 2.30 $\mu\text{Gy/MU}$ and a minimum 0.09 $\mu\text{Gy/MU}$.

Fig. 3. Plot of dose reduction ratios (DRR) using 1- to 9-cm-thick lead shielding plates. (a) P-3 phantom; (b) P-6 phantom. As can be seen, there is a plateau in the dose reduction curves where the effect of increasing shielding thickness is limited because the fetal dose from patient scattering cannot be shielded.

Fig. 4. Fetal dose comparison plot of treatment plans with different gestation stage, treatment field, treatment center, distance from field edge to the nearest fetal point (DEF), and distance from treatment center to the nearest fetal point (DCF).

Fig. 5.

The plot of fetal dose vs. anterior distance from the shield to the patient abdomen surface. For the mantle treatment *plan 3*, head and neck treatment *plan 11*, and the brain treatment *plan 13*, a 4-cm change in the anterior positions of the shielding plate resulted in the fetal doses changing 3%, 2% and 1%, respectively.

Fig. 6.

Plot of the maximum permitted MUs of 10 cm × 10 cm field treatment plans with shielding of 3- and 7-cm lead plates. As shown above, the maximum permitted MUs increase with the distance from field edge to the nearest fetal point, gestational stage, and the shielding thickness.

Table 1

Parameters for typical treatment plans for pregnant women.

Plan no.	Treatment field	Field size (cm ²)	Phantom	DCF (cm)	DEF (cm)
1	Mantle	10 × 10	P-3	29	24
2	Mantle	10 × 10	P-3	24	19
3	Mantle	10 × 10	P-3	19	14
4	Mantle	10 × 10	P-3	14	9
5	Mantle	10 × 10	P-6	29	24
6	Mantle	10 × 10	P-6	24	19
7	Mantle	10 × 10	P-6	19	14
8	Mantle	10 × 10	P-6	14	9
9	Mantle	20 × 20	P-3	29	19
10	Mantle	20 × 20	P-6	19	9
11	Head & neck	10 × 10	P-3	45	40
12	Head & neck	10 × 10	P-6	35	30
13	Brain	10 × 10	P-3	61	56
14	Brain	10 × 10	P-6	51	46