

# Design of a generally applicable abdominal shield for reducing fetal dose during radiotherapy of common malignancies in pregnant patients

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**Background:** In most cancer cases, the treatment choice for a pregnant patient is radiotherapy. In these patients, the abdomen is usually not exposed; therefore fetus exposure is due to peripheral dose (PD). The purpose of this study was to estimate the fetal dose (the maximum PD in each pregnancy stage) for modalities available and to fabricate and evaluate a generally applicable fetal shield. **Materials and Methods:** PD values were measured for brain, breast and mediastinum irradiation in a whole body anthropomorphic phantom using a NE 2571 ionization chamber. An external shield was then designed to reduce the fetal dose to the standard dose limit, 5 mSv. **Results:** The range of PD values as a function of distance from the field's edge were as follows 1) 9.4-259 cGy for Mantel field; 2) 6.5-95 cGy for chest wall irradiation with 10 MeV electrons, 3) 8.5- 52.5 cGy for tangential field with Co-60 and 4) 4.8-7.8 cGy for brain radiotherapy with 9 MV photon. PD values for the same setups using the fetal shield were as follows: 1) 1.4-22 cGy, 2) 0.5-4 cGy, 3) 1.5-5 cGy and 4) under 1 cGy. **Conclusions:** The measured PD data sets can be used to estimate fetal dose for specific treatment setups and pregnancy stages. The use of external shield designed in this research reduced the fetal dose effectively to under the threshold (a 70-90% reduction), except for the final stages of pregnancy in Hodgkin's patients. **Iran. J. Radiat. Res., 2012; 10(3-4): 151-156**

**Keywords:** Fetal Dose, radiotherapy, pregnancy, fetal shield.

## INTRODUCTION

Radiotherapy plays a major role in the treatment of malignancies in pregnant patients. It is often one of the treatment modalities used as a sole treatment, or in combination with other modalities. The most common tumors of pregnant patients are lymphomas, leukemia, melanomas, and tumors located in the breast, uterine cervix and thyroid <sup>(1)</sup>. In some cases it is not

possible to postpone the radiation treatment until after delivery <sup>(2)</sup>. Biological effects of ionizing radiation on the fetus include malformations, growth impairment, mental retardation, cancer induction, hereditary defects and death; therefore it is important to keep the dose to the fetus as low as possible. The frequency and magnitude of effects differ depending on the absorbed dose, type of radiation and the gestation age at which exposure occurs. As recommended by AAPM TG36, the equivalent absorbed dose received by a fetus should be kept under 5 mSv <sup>(1)</sup>. Radiotherapy in pregnant patients is aimed towards controlling tumor growth while also giving the fetus the best chance for a normal life. To achieve the optimum balance between the risks and benefits of radiotherapy, the fetal dose has to be evaluated before treatment.

The treatment volume in a pregnant patient should never contain the abdomen <sup>(1, 3)</sup>. Therefore, the dose absorbed outside of the radiation treatment field, also called the peripheral dose (PD), is responsible for fetal irradiation. The principal sources of PD are: (1) leakage radiation through the treatment head of the machine; (2) radiation scattered from the collimators and beam modifiers and (3) radiation scattered within the patient from the irradiated volume. For higher energy photon beams (>10 MV), the photo-neutron contribution to the absorbed dose becomes considerable and has to be taken into account <sup>(1)</sup>.

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PD depends on the following factors: (a) Distance from the radiation field edge: PD decreases almost exponentially with increasing distance from the field edge; (b) Depth in tissue or in phantom: PD dependence on depth is small; (c) Field size: PD increases with increasing field size; (d) The use of wedges and other beam modifiers: wedges increase the dose near the beam by a factor of 2 to 4 and the use of lead shielding devices can increase the PD by a factor of 2 to 5; (e) Beam energy: for a given depth and field size the PD for photons from 4 to 25 MV is of the same order of magnitude and is qualitatively similar. In contrast, the PD from Co-60 at distances greater than 10 cm away from the field edge is considerably higher because of a larger amount of head leakage<sup>(1)</sup>. Reduction in PD is achieved using gantry angle, field size and patient position adjustments. However, when treating a pregnant patient, the most effective method to reduce the fetal dose is through the use of shielding<sup>(1, 4)</sup>.

Fetal dose estimates in radiotherapy have been reported by many authors<sup>(1, 2, 4-15)</sup>. However, most of the information available on the measurement of fetal dose in radiotherapy has been reported for individual patients, specific treatment protocols and using phantoms specifically made for that patient. In a busy radiotherapy department, however, individual patient dosimetry is a demanding option. The aim of the present study was to create a dataset for PD as a function of distance from the edge of the field for common treatment protocols and modalities available in our department. The PD dataset was then used to design and fabricate a generally applicable fetal shield.

## MATERIALS AND METHODS

In this research, a generally applicable fetal shield is defined as a shield that is effective for all treatment protocols in order to protect the fetus in all stages of pregnancy. Design of the fetal shield involved the following steps: (1) Simulate and implement

the treatment plan for common malignancies in pregnant women in our center using physical phantoms, (2) measure unshielded fetal dose in the phantom, and (3) implement a fetal shield and measure shielded fetal dose from the same treatment protocols.

### Treatment plans

The malignancies considered were Hodgkin's disease (HD) and Non Hodgkin Lymphoma (NHL), as well as breast and brain tumors for which the following radiotherapy protocols are commonly employed in our center:

Plan A: HD and NHL treatment with mediastinum involvement using a Mantle field with 12 MV photon beams and a total dose of 3600 cGy.

Plan B: Chest wall treatment, using a 10 MeV electron beam and a total dose of 5000 cGy.

Plan C: Breast treatment, using two Co-60 tangent beams and a total dose of 5000 cGy.

Plan D: Brain treatment, using two 9 MV photon lateral beams and a total dose of 6000 cGy.

### Measurement setup

Treatment portals were delivered on a series of in-house acrylic phantoms which included the head and neck, chest and pelvis phantoms (figure 1). Phantoms were fabricated using acrylic sheets of 1 cm thickness. Measurements of peripheral dose were done in the abdomen and pelvis phantoms. The dosimeter was a 0.6 cm<sup>3</sup> NE 2571 ionization chamber ( $N_{D,W} = 4.51$  cGy/nC). For measurements, a 2 cm thick slab of acrylic was used. A prefabricated hole of appropriate dimensions allowed the chamber to be inserted into this slab. The slab was then placed between the phantom slices at the measurement positions, specified by the distance from the portals inferior edge. PD does not depend strongly on depth and a depth of 10 cm has been recommended as a reference depth for fetal dose estimation<sup>(1, 4, 8, 10, 11, 14, 16)</sup>.

Fetal dose was estimated by taking PD measurements at selected points away from the field's inferior edge. Considering the location of fetus in different stages of pregnancy period, these values will reflect the range of dose throughout the fetus. In order to correlate the measurement points to the position of the fetus, a rate of 1 cm per week of pregnancy can be used for increase in the height of the fundus uteri with respect to the symphysis pubis<sup>(16, 17)</sup>.



Figure 1. PD measurement set up for brain lateral fields in radiotherapy of the brain.

### PD determination

Absorbed dose determination was performed according to the recommendations of IAEA's TRS-398 protocol<sup>(18)</sup>. Calculation of absorbed dose requires knowledge of the average energy of the photon spectrum at the point of measurement. The PD was measured at several points in the peripheral region of the fields, where the average energy of the photon spectrum cannot be measured accurately. Furthermore, the methodology for calculating absorbed dose is not clear, in the peripheral region of an electron field, where dose is deposited primarily by bremsstrahlung photons originated in the treatment head and phantom.

Therefore, defining an average energy for the complex photon spectrum in the peripheral region of an electron beam is not straight forward. For the following reasons it was assumed that electron and photon beams with roughly equal nominal beam

energies result in similar photon energy spectra outside of the primary beam: 1- the parameters in the dose calculation protocol do not vary by more than a few percent over a wide range of photon energy, 2- the response of a thimble ion chamber is quite flat over this same energy range, and 3- an accuracy of 10% for a fetal dose estimate is acceptable<sup>(14)</sup>. Therefore, in this work, the photon energy spectrum in the peripheral region of the 9 MV photon beam and 10 MeV electron beam were defined by the central axis ionization ratio for the 9 MV beam. Also, since the measurement of PD involved very small magnitudes of dose, for each measurement, the number of MU was raised to 1000 and the readings were then normalized to the maximum dose along the central axis of the treatment field.

### Shielding design

In this study a thickness of 4 cm of Cerrobend (a Lipowitz's alloy containing 50% bismuth, 26.7% Lead, 13.3% tin, and 10% cadmium with a mass density of 9.6 g/cm<sup>3</sup> and a melting point of 70 °C) material (about 2 HVL)<sup>(19)</sup> for a nominal energy of 12 MV was used to design a portable shield. Shield thickness and its material were selected based on the following data. According to the TG-36 report, regardless of energy, shielding thicknesses of 5 to 7 cm of lead or 6 to 8.5 cm of Cerrobend are adequate to reduce PD effectively<sup>(1)</sup>. However, There are reports indicating that these thickness values seems to be too conservative for some treatment plans and some authors have reported using smaller thicknesses<sup>(1,4,10,11)</sup>. Also, in a Monte Carlo evaluation of effective shielding of PD for different treatment plans (Mantel, head and neck and brain), it was shown that the dose reduction curve shows a plateau at a certain shield thicknesses<sup>(7)</sup>. It was indicated that depending on the plan, there is an optimal thickness for which the head leakage and collimator scatter photons are effectively shielded and patient scatter becomes the main source of fetal dose. It was shown that

when a 6 MV beam, was shielded with 3-9 cm thickness of lead, the average dose reduction was about 2% per cm of shielding, i.e. additional shielding did not change the PD significantly. Therefore, since energy dependency of PD is negligible <sup>(1)</sup>, one can conclude that for each treatment plan, an optimal shield thickness may be less than the thickness values recommended by TG36. It was also concluded that when the optimal number of HVL of lead and Cerrobend shielding was used, the shielding effects of the two materials were identical <sup>(7)</sup>.

The highest nominal energy used in this research was 12 MV. For high energy photon beams produced by clinical accelerators, several beam quality specifiers has been proposed in the literature <sup>(20, 21)</sup>. These specifiers include: TPR<sub>20,10</sub> (tissue phantom ratio at depths of 20 and 10 cm), PDD10 (the percentage depth dose at 10 cm depth) and d80% (the depth at the 80% dose mark). In this research, d80% for the highest used energy photon beam (nominal energy of 12 MV) was 85 mm. This depth corresponded to a 10 MV photon beam <sup>(21)</sup>. Therefore, the highest photon energy used in this study was 10MV, for which photoneutron production was negligible <sup>(1)</sup>.

To support the shield independently from the couch, an iron table was positioned over the treatment couch. This table made treatment possible in different SSD's. The vertical position of this table was made adjustable in 3 steps. To assemble the shield, Cerrobend sheets were placed on the table top, and also attached to the sides and front of the table. It is possible to wheel the iron table over the treatment coach as close as possible to the gantry without interfering with the treatment portals. The phantoms were then put into treatment position underneath the shield. In figures 2 the application of this shield in the treatment position is shown.

#### **Evaluation of abdominal shield**

The effectiveness of the designed shield was evaluated by repeating PD measure-



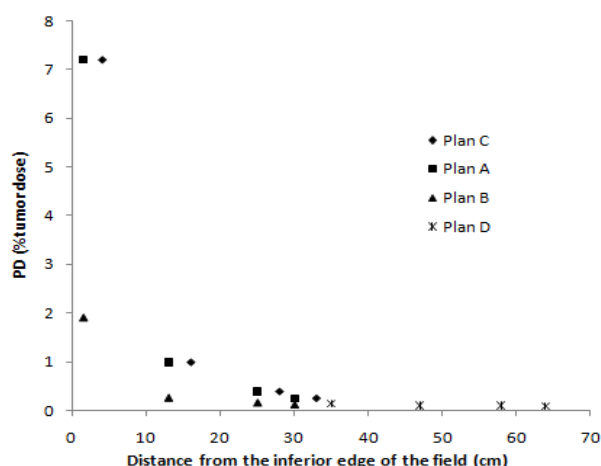
**Figure 2.** Application of fetal shield in treatment of anterior field (lateral view).

ments while using the fabricated shield. For each treatment field, measurements were done in two distances, corresponding to the closest (point 1) and furthest (point 2) fetal positions relative to the beam edge. Peripheral dose at point 1 is the maximum dose that fetus receives at the end of third trimester. However, the dose at the second point reflects the dose that fetus receives in its most sensitive stage. PD for shielded fields at points 1 and 2 are compared to unshielded values for each plan.

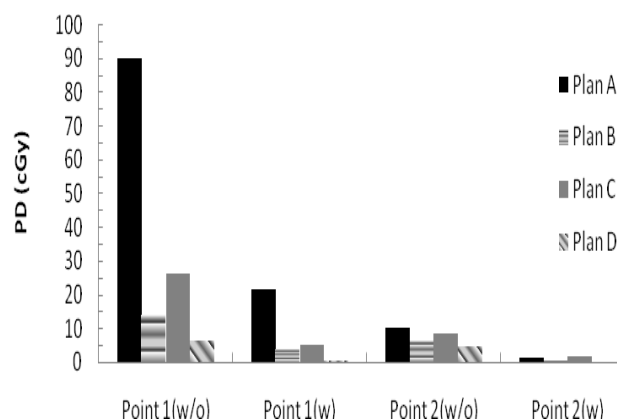
#### **RESULTS**

The results of PD measurements for each treatment protocol at different distances are presented in figure 3. As reported in literature <sup>(1)</sup>, PD values vary exponentially as a function of distance from the field's edge. Maximum PD values correspond to the sub-diaphragm points closest to the inferior field edges, i.e. points at 1.5 cm distance from plans A and B, 3.5 cm from plan C and 35 cm distance from plan D. The ranges of PD values were as follows 1) 9.4-259 cGy for plan A; 2) 6.5-95 cGy for plan B, 3) 8.5- 52.5 cGy for plan C and 4) 4.8-7.8 cGy for plan D.

The effectiveness of the designed shield was evaluated by repeating PD measurements using the fabricated shield. Shielded PD values measured at points 1 and 2, and



**Figure 3.** The results of PD measurements as a function of distance from the inferior edge of the field's edge during radiotherapy of plans A, B, C and D. PD values were normalized to central axis tumor dose. PD values vary exponentially as a function of distance from the field's edge.



**Figure 4.** Comparison PD values with (w) and without (w/o) a fetal shield in two sub-diaphragm positions, corresponding to the closest (point 1) and furthest fetal position (point 2) relative to the inferior beam edge. This shield reduces PD effectively in all plans but this thickness of material did not provide adequate protection of the fetus in final stages of pregnancy in plan A.

**Table 1.** Evaluation of shield effectiveness in fetal dose reduction for Plans A, B, C and D.

Treatment plan	Distance from inferior edge of the field (cm)	Percentage of fetal dose relative to the dose of field center		Fetal dose reduction percentage
		Fetal dose without shield	Fetal dose with shield	
A	6	2.5%	0.6%	75%
	30	0.28%	0.04%	86%
B	10	0.28%	0.08%	71%
	33	0.13%	0.01%	92%
C	10	0.52%	0.1%	81%
	33	0.17%	0.03%	82%
D	40	0.11%	0.01%	91%
	64	0.08%	<0.01%	<0.01%

were compared to the unshielded values for each plan in figure 4. Similar results, but in term of percentage of tumor dose, are illustrated in table 1. These values show a fetal shield efficiency of 70-90 %.

## DISCUSSION

A PD database was generated as a function of distance from the field edge for treatment protocols used to treat HD, breast and brain malignancies in our center. The generated PD dataset can be used to predict the range of fetal doses corresponding to the patient's stage of pregnancy, for each treatment protocol. The range of PD values were as follows 1) 9.4-259 cGy for 12 MV

Mantel fields; 2) 6.5-95 cGy for 10 MeV chest wall irradiation, 3) 8.5- 52.5 cGy for tangential breast field with Co-60 and 4) 4.8 -7.8 cGy for 9 MV brain radiotherapy. This research shows that the fetal dose for the evaluated treatment protocols was more than the standard limit of 5 cGy and it is therefore necessary to use an external shield when treating pregnant patients. Similar dose range values have been reported by others (4, 5, 10, 13, 14, 22, 23). For example, Cygler *et al.* investigated the fetal dose in radiotherapy of a 23 weeks pregnant woman for HD with mantle field and 10 MV photons (5). The dose to the fundus and pubis were 10 and 3 cGy, in a 35 Gy total dose treatment. In another study, the fetal dose

was measured in a brain radiotherapy protocol using opposed 6 MeV lateral 10×15 cm<sup>2</sup> fields, both clinically and in phantom. Fetal dose was estimated to be 0.09% of the tumor dose<sup>(22)</sup>. In another study, fetal dose in chest wall irradiation with electron beam was simulated using an anthropomorphic phantom and the measured dose to the unshielded fetus for this plan was 5.3 cGy<sup>(23)</sup>. Implementation of the shield type described in this work reduced the peripheral dose effectively (70-90% reduction). The shield design in this work has purposely been made flexible so that it can be used with minimal modification in the irradiation of almost all treatment volumes. A thickness of 4 cm of Cerrobend was found to be very effective in reducing the fetal dose in a standard brain and breast and HD radiation therapy course but this thickness of material did not provide adequate protection of the fetus in final stages of pregnancy in Hodgkin's patients. For these patients, extra protection will be required.

## ACKNOWLEDGMENT

*The authors are indebted to Dr. M. Jalilian and Sh. Monadi for their valuable advices during this research.*

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