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FULL PAPER

Effectiveness of abdominal shields in chest radiography: a Monte Carlo evaluation

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Objective: The aims of this study were to assess the effectiveness of abdominal shields in reducing the dose to shielded organs during posteroanterior (PA) chest radiographic examinations, to evaluate the various factors contributing to the dose under the shield and to address the concern that internal scatter from the shield may lead to a dose increase to shielded organs.

Methods: Monte Carlo simulations of radiographic exposures to a soft-tissue “patient” phantom of $80 \times 34 \times 24$ cm were performed at the peak X-ray tube voltage of 100 kV. The abdominal shield was simulated as 0.5-mm lead with/without a 0.2-mm-thick plastic film cover. The shield-to-phantom spacing varied from 0 to 40 cm.

Results: The dose to the phantom at different phantom depths and distances from the X-ray field edge was

recorded. At the level of the uterus and ovaries (approximately 8-cm deep and 15 cm from the field of view), the dose reduction was approximately $0.035 \mu\text{Gy}$ or 4%. A small dose increase (maximum: $0.4 \mu\text{Gy}$) to the phantom surface layer was also observed.

Conclusion: In general, there was a small dose reduction in regions protected by the shield. However, the overall risk benefit to a patient would be negligible.

Advances in knowledge: Abdominal shields used during PA chest radiography provide a small dose reduction to organs at depth by removing the primary X-ray beam photons that are scattered in air, while giving a small increase in skin dose due to X-rays scattered from the abdominal shield internal surface.

INTRODUCTION

Good radiographic practice aims for the patient dose from any radiographic procedure to be “as low as reasonably achievable” (ALARA). Basic aspects of the ALARA principle were first introduced by the International Commission on Radiological Protection in 1977¹ and received further developments in subsequent publications of the International Commission on Radiological Protection and other safety agencies.^{2,3} An important point in lowering patient doses is optimization of equipment and methods, so that the radiation administered for diagnostic purposes is sufficient but not greater than that which enables the procedure to provide the required information.³

Although the radiation exposure from radiographic examinations of the chest is low compared with most radiological examinations, plain film chest radiography was listed in the European Commission Report No 54 as the most frequent study.⁴ It was among the 20 procedures which had the highest contribution to the collective effective dose in the 10 European countries who participated in the DOSE DATAMED project,⁴ highlighting the need for patient dose optimization.

One of the methods used to reduce the dose to organs that are not in the direct path of the primary beam is to shield that part of the patient’s body with radiation-absorbing material. Gonad and abdominal shields have been used on patients, from neonates to adults, during chest X-ray procedures as part of the ALARA principle.^{5,6} A similar practice, often referred to as “out-of-plane” shielding, has also found an application in dental radiography and CT, where there were reports of a noticeable dose reduction to shielded organs.^{7–9} During chest X-ray examinations, the use of abdominal shields complicates and possibly compromises the procedure, and their role in reducing patient dose is controversial. While Njeh et al⁶ found a negligible radiation protection effect from the use of abdominal shields during chest X-ray examinations, Hashimoto et al¹⁰ and Jackson et al¹¹ reported a significant level of dose reduction. Regardless of any potential benefit of the use of shields, they must be carefully placed so as not to interfere with imaging the organs of interest.

The aim of the present study was to assess, using Monte Carlo simulations, the effectiveness of an abdominal shield in reducing the dose to shielded organs during

posteroanterior (PA) chest radiographic examinations. The study aimed to assess the individual components (external and internal scatter) that contribute to the complex pattern of dose under the shield and, in particular, to address the concern that internal scatter from the shield may possibly lead to a dose increase to shielded organs.

METHODS AND MATERIALS

In the present work, simulations of coupled photon-electron transport were performed using the DOSXYZnrc module of the EGSnrc Monte Carlo simulation package.^{12,13} The “patient” phantom was a soft-tissue-equivalent rectangular parallelepiped of $80 \times 34 \times 24$ cm (x, y, z). The following parameters for a PA chest X-ray were simulated: peak tube voltage = 100 kV, source-to-skin distance = 152 cm, source-to-image distance = 180 cm, phantom-to-image distance = 4 cm and X-ray image field size = 35×43 cm (x, y).

Dose measurements were made in voxels of 2×32 cm (x, y). The z -sizes (thicknesses), dependent on the voxel location (depth) from the top surface of the phantom, were (in cm): 3×0.2 , 1×0.4 , 1×1.0 and 11×2.0 (total 24 cm). The abdominal shield was simulated as a 43×43 cm (x, y) flat sheet of 0.5-mm-thick lead with/without a thin (0.2 mm) plastic film cover. It was positioned on the tube-facing side of the phantom (*i.e.* simulating a posterior shield for the abdominal region in a PA chest study) and was tested with a shield-to-phantom spacing of 0, 20 and 40 cm—the latter simulations representing the clinical situations where the abdominal shield cannot be placed close to the patient’s back owing to their body habitus. The top edge of the shield was 0.1 cm from the primary beam edge (on the surface of the phantom).

The number of photon histories in each simulation was set to achieve a low standard deviation (SD) for voxel dose—resulting in SD values of $<0.3\%$. Absolute voxel doses were calculated using a normalization coefficient to achieve a dose–area product in air of $69 \text{ mGy} \times \text{cm}^2$. The dose–area product and other settings stated above are typical of chest PA radiographs taken at the Royal Adelaide Hospital, Australia.

To assess the individual components of dose (from external and internal scatter of X-rays) under the shield, the following experiments were performed. Firstly, the simulations (with/without shield) were performed in a “vacuum” to assess the contribution from internal scatter off the shield alone (*i.e.* excluding scatter from the primary X-ray beam in air). Secondly, simulations were performed with the phantom, without the shield, surrounded by either air or vacuum—to assess the contribution from external scatter in air alone. In all simulations in vacuum, the phantom entrance dose in the X-ray field of view was normalized to that of the phantom in air.

Finally, a separate simulation was performed within the plastic cover to analyze the potential effect on the dose to the surface layer of the phantom from any electrons ejected from the abdominal shield. 10 depth dose (z) measurements were made in $2 \times 32 \times 0.002$ cm (x, y, z) voxels.

RESULTS

The absolute doses (in microgray) to the phantom outside the primary beam, without the abdominal shield, are shown in Figure 1. Figure 2 illustrates the effectiveness of the abdominal shield at varying depths and distances from the edge of the X-ray beam (the distances were measured along the x -axis on the top of the phantom). In general, there was a small reduction in absolute dose in regions protected by the shield that did not exceed $0.09 \mu\text{Gy}$ for voxels deeper than 4 cm. In a voxel at a depth of 8 cm and at 15 cm from X-ray field (corresponding approximately to the level of the uterus and ovaries), the dose reduction was about $0.035 \mu\text{Gy}$ or 4%.

At a fixed distance from the edge of the X-ray field, the dose reduction was generally smaller for deeper voxels. In relative dose terms, the dose reductions were greater for sections of the phantom away from the X-ray field and especially in low-depth regions of the phantom (Figure 3). The abdominal shield was most effective when there was no gap between it and the phantom (Figure 4).

When the contribution from scatter in air on the tube side of the phantom was removed (experiments in vacuum with and without lead shield), there was a small net increase in dose to the shielded regions of the phantom compared with no shield owing to backscatter of internal X-rays by the abdominal shield (Figure 5). This net increase in dose reduced rapidly at depth within the phantom and as the distance from the edge of the primary beam increased. Figure 6 shows the magnitude of the contribution from external scatter in air alone (air vs vacuum) with no lead shield in place—*i.e.* with no internal backscatter from the lead shield. The contribution rapidly diminished at depth within the phantom and as the distance from the edge of the primary beam increased.

In the simulations performed with the lead shield with/without a thin (0.2 mm) plastic cover on the inner side of the shield, there was a noticeable dose increase in the superficial layers of the phantom compared with no shield—owing to photon backscatter, and ejection of electrons, from the lead shield (Figure 7). The dose increase did not extend beyond 4 mm into

Figure 1. Phantom doses at four depth locations—no abdominal shield.

Figure 2. Dose reduction as a function of distance from X-ray field edge—with abdominal shield.

the phantom. Electrons ejected from the shield did not extend beyond 0.05 mm from the lead and were completely removed by the plastic cover (Figure 8). A maximum dose increase to the surface layer of 0.4 μGy (averaged over the 2-mm-thick voxels) was recorded for the phantom with the shield and plastic cover.

DISCUSSION

This study gives a detailed picture of the various factors that contribute to the dose under a tube-facing abdominal lead shield used for protecting organs outside the primary beam in PA chest radiography. Without the abdominal shield, the radiation exposure to regions of the phantom outside the primary beam (from both internal and external scatter) was small and decreased rapidly with depth and distance from the primary beam edge (Figure 1). There was a measurable (absolute) dose reduction in regions within the phantom, outside the primary beam, with the use of the abdominal shield (Figure 2), which was greatest near the edge of the X-ray field and closer to the surface. In relative terms, however, the dose reduction increases as the distance from the primary beam edge increases and the depth in the phantom decreases (Figure 3)—which is in agreement with the results of a previous phantom study that

Figure 3. Relative dose change (shielded vs no shield) as a function of distance from X-ray field edge.

Figure 4. Dose reduction at 4-cm depth in phantom due to abdominal shield placed directly on the phantom, at 20 cm and 40 cm from phantom.

investigated the effectiveness of an abdominal shield used during chest CT.⁹

The dose reduction to the phantom voxels protected with the shield was very small and did not exceed 0.09 μGy for voxels at 4 cm or deeper from the phantom surface. In a voxel at a depth of 8 cm and at 15 cm from X-ray field (corresponding approximately to the level of the uterus and ovaries), the dose reduction was about 0.035 μGy or approximately 4% of the dose value in that voxel when the abdominal shield was not used (Figures 2 and 3). This study, however, did not simulate the spine and in particular pelvic bones, and it is likely that the reduction will be lower than this value—as the bony structures will provide some shielding to internal organs from primary beam photons scattered in air. Jackson et al¹¹ reported a statistically significant (19%) dose reduction to the uterus, but a non-significant effect for ovaries, when a tube-facing abdominal shield was used for a PA chest radiographic study in a tissue-equivalent adult torso phantom. There was, however, considerable variability (relatively high SDs) in their thermoluminescent dosimeter measurements. In the present study, the SD was <0.3%.

Figure 5. Absolute dose increase due solely to internal X-ray scatter from abdominal shield (phantom in vacuum).

Figure 6. Absolute dose increase due solely to X-rays scattered in air (no shield).

Backscatter from an image receptor/bucky and wall was not simulated in this study owing to the complexities in modelling this part of the system—and only a tube-facing lead abdominal shield (not a wrap-around) was evaluated. In the present study, only approximately 3% of the primary beam passed through the phantom. As only a fraction of this radiation would be scattered off the image receptor/bucky, predominately into the primary beam area, any error in the measured dose reduction value to organs at depth from not including this scatter data would be small.

In this study, the use of Monte Carlo simulation enabled the contributions from internal and external scatter to be evaluated separately. Scatter of primary beam X-rays inside the phantom was the biggest contributor to dose in the great majority of phantom regions outside the primary X-ray beam. Scatter of the primary beam photons in air also contributes to the dose, and these photons were shielded by the tube-facing abdominal shield. This external scatter only made a significant contribution to the dose value at a considerable distance from the edge of the X-ray field, where the absolute dose from all scatter sources is very low—for example, up to 15 cm from the field of view its contribution was <9% at a depth of 4 cm (Figure 3). If a shield is used, it should be positioned as close as possible to the patient surface—as the shield is moved away from the surface of the

Figure 7. Effect of abdominal shield (with/without plastic cover) on surface phantom dose.

Figure 8. Dose in plastic cover under the abdominal shield (at 3 cm from primary beam edge), normalized to the first voxel.

phantom, the shielding effect reduces with an increasing amount of radiation (scattered in air) passing under the shield (Figure 4).

The dose change resulting from the use of an abdominal shield is caused by the net effect of two competing processes. Firstly, some X-rays scattered within the phantom strike the inner surface of the abdominal shield, where they are backscattered and absorbed in the phantom—resulting in a dose increase (compared with no shield). This is illustrated in the simulations performed with/without the abdominal shield in a vacuum—*i.e.* without a contribution from photons scattered in air (Figure 5). The dose increase effect reduces at increasing depth and distance from the primary beam edge. The magnitude of the other contributor to dose, from photons scattered in air in the direction of the phantom, was assessed by comparing the doses to the phantom in air and in vacuum, without the shield—*i.e.* with no contribution from internal backscatter off the shield (Figure 6). The dose from scatter in air (that is removed by the abdominal shield) diminished with increasing depth into the phantom and distance from the primary beam edge. The net result to dose from the two competing effects was a small reduction in dose for the shielded region of the phantom—except, however, in the surface layer (<4-mm deep), where there was a small dose increase (maximum 0.4 μGy: in a 2-mm-thick surface layer at 3 cm from the beam edge, with the plastic cover on the shield) (Figure 7). The increased surface dose was due to internally scattered photons (within the phantom) being backscattered from the inner side of the lead shield, but not due to ejected electrons, which did not extend beyond 0.05 mm from the lead and were absorbed in the plastic cover (Figure 8).

The results obtained in this study apply to lead shields; “light-weight” (non-lead) shields were not evaluated, as the required detailed information on their composition was not available for accurate simulation. Although lightweight aprons have a different material composition (atomic structure) compared with lead, and hence a different absorption/scatter profile, they are designed to have a certain “lead equivalence” for attenuation of X-rays at a set peak tube voltage (usually 100 kVp). As the simulations have shown, the dose reduction to organs outside

the primary beam from the use of lead shields is very small, and it is expected that the use of lightweight shields of the same lead equivalence would not significantly alter these results.

This study has shown that the reduction in dose to organs outside the field of view, with the use of an abdominal shield, is very small, as the major contribution to dose comes from photons scattered internally—from within the primary beam region of the patient. Consequently, any risk benefit from the use of shields in PA chest procedures would be negligible. The most important technical method for reducing the radiation exposure (and risk) to the patient from chest radiographic examinations, besides the optimal setting of imaging parameters, is the use of appropriate collimation (size and positioning) of the X-ray beam. However, in relation to other higher radiation studies, such as chest CT, the level of protection from the use of shields will be higher and this warrants further investigation using similar techniques to those outlined in this study.

CONCLUSION

This study has shown that the use of abdominal shields during PA chest radiographic examinations provides very little protection to shielded organs, as the primary source of radiation exposure is from X-rays scattered within the patient. The risk benefit from this low level of protection is negligible.

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