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Scatter dose calculation for anti-scatter linear grids in mammography

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ABSTRACT

Monte Carlo simulations were used to optimize the geometry of a mammography anti-scatter linear grid to achieve minimum scatter-to-primary ratio (SPR) for different X-ray tube voltages. A single optimum design of the grid with 0.9 mm septa height, 12 μm septa thickness and 100 μm interspace thickness was found for breast phantom thicknesses between 30 and 80 mm. The optimal grid has 0.153–0.330 scatter-to-primary ratio, a Bucky factor (BF) less than 2.5 and a contrast improvement factor (CIF) of 1.3.

1. Introduction

Breast Cancer is the most common kind of cancer in women all over the world (American Cancer Society, 2004). Early diagnosis and treatment improve the chance of survival for breast cancer patients. Presently, mammography is the most reliable diagnostic screening method. Annual mammogram is recommended for women over the age of forty (American Cancer Society, 2004). However, the usefulness of mammography examination depends strongly on the diagnostic image quality. The fraction of false negatives in reading the mammograms is estimated to be 15–25% (American Cancer Society, 2004). The major image degrading factor in screen-film mammography is scattered radiation. In mammography, the scatter-to-primary radiation ratio ranges from 0.35 to 1.0 depending on field size and breast thickness (Boone et al., 2000). The requirements for decreasing scattered radiation without increasing dose to the patient, lead to continuous efforts to optimize mammographic techniques. Various methods have been developed to reduce the scattered radiation incident on the film-screen receptor. The most successful device is the anti-scatter grid that was first introduced by Bucky (1915). The use of the grid attenuates the exposure to the film. In order to compensate for the grid attenuation, dose to the breast has to be increased. Because of the potential carcinogenic risk of ionizing radiation, the radiation dose to the breast should be minimized. Therefore it is desired to optimize the anti-scatter linear grid geometry to improve the image contrast without significant increase in radiation exposure to breast. Increasing the X-ray source energy may also reduce the scatter contribution. However, this is not an option because clinically the optimal mammography contrast is obtained at tube voltages 24–32 kVp.

Performance of different types of grids in mammography application has been investigated by several authors (Barnes and Brezovich, 1978; Muntz et al., 1983; Dance and Day, 1983; Boone et al., 2002; Chan and Doi, 1982; Rezentes et al., 1999; Boone, 1999; Skubic and Fatouros, 1989). Investigations have been carried out to study angular and energy dependency of grid transmission (Muntz et al., 1983), the scatter-to-primary ratio dependence on the image receptor and breast area (Dance and Day, 1983). Boone et al. (2002) have studied a cellular grid with 15 μm tungsten septa and air interspace (with a 4:1 grid ratio) and observed better rejection of scatter and an improved image contrast but with higher Bucky factor, which is the ratio of the total radiation recorded by the image receptor without grid to that with the grid for constant incident radiation on the receptor. Further studies have been conducted for ultra-high-strip-density grids [with 7–10 lines/mm strip density and 20–37 μm strip thickness] and 40–90% increase in contrast was reported for Bucky factors of 2–3.5 (Chan and Doi, 1982). The results for ultra-high-strip-density grids are comparable to those from conventional grids. Rezentes et al. (1999) have reported better contrast improvement factor (CIF) for cellular grid at 25 and 30 kVp and superior *BF* for linear grids.

The effect of grid on mean granular dose was also significant. Boone (1999) calculated the mean glandular dose for a variety of

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X-ray tube–anode filter combinations for different breast compositions and thicknesses without considering grid effects. Skubic and Fatouros (1989) made mean glandular dose calculation taking into consideration grid effects and observed an increase in absorbed dose with increasing glandular content and breast thickness for low-energy beams. Due to the grid, the dose increased by a factor of 2–2.6 while the image contrast decreased with the increase of glandular fractions, especially for small breasts and low-energy beams.

In this study, the linear grid geometry for different breast sizes has been optimized by minimizing scatter-to-primary ratio, and the Bucky factor within required safe levels. Optimum grid geometry will minimize scatter and hence improve the image contrast. This will enhance the ability of the radiologist to diagnose the disease at an early stage and may save the lives of many patients.

2. Materials and methods

Monte Carlo Simulation plays an important role for optimization studies in diagnostic radiology, radiation physics and dosimetry (Andreo, 1991; MCNP, 2005). The optimization of grid geometry is done in this work by studying *SPR* and *BF* using a general-purpose Monte Carlo N-Particle transport code MCNP5 running on Pentium-based PCs. In the following, important terms and relations in mammography relevant to this study are described briefly.

2.1. Image contrast and scatter-to-primary ratio

Image contrast is the difference between regions in an image. Assuming a tumor is imbedded in tissue, if there is no scattering then the contrast from primary photons only C_p is defined by

$$C_p = \frac{I_t - P}{P} \tag{1}$$

where I_t is the primary radiation that corresponds to the tumor region and P is that corresponding to the surrounding tissue. If there is scattering, and assuming the amount of scattering due to both the tumor and the surrounding tissue is the same, then the contrast with scatter C_s becomes

$$C_s = \frac{(I_t + S) - (P + S)}{P + S} = C_p \frac{1}{1 + S/P} \tag{2}$$

where S is the scattered radiation and P the primary radiation. S/P is the scatter-to-primary ratio. It is obvious from Eq. (2) that the image contrast is decreased with the increase in scattered radiation. High contrast is very important to differentiate tumor masses from surrounding normal tissue. Scatter also degrades resolution by lowering the high spatial frequency content of the resultant image (Boone et al., 2000). High resolution is very important in mammography in order to detect microcalcifications, which are early indications of breast cancer.

2.2. Anti-scatter linear grid

The linear grid used for the simulation is made of parallel lead septa with thickness d and height h. The interspace between the septa is made of carbon fiber ($(C_6H_{10}O_5)_n$, density 1320 kg/m^3) with thickness D. The grid ratio (R) is defined by septa height to interspaces thickness ratio (h/D). Ideally, the grid should allow most of the primary rays to go through and block all the scattered rays. However, in reality not all primary rays go through the grid and not all scattered rays are stopped. The Bucky factor is the factor by which the absorbed dose in the patient is increased so that the image receptor, usually film, density remains the same as that without the grid. Therefore, *BF* is equal to the ratio of the total radiation recorded by the image receptor without grid to that with the grid for constant incident radiation on the receptor. The contrast improvement factor is the ratio of image contrast with the grid to that without the grid. It can also be expressed in *BF* and the primary transmission T_p.

$$CIF = \frac{C_p(1/1 + (S/P)_{grid})}{C_p(1/1 + (S/P))} = T_p \times BF \tag{3}$$

Another factor used to characterize the grid is the selectivity $\sum$, which is the ratio of primary transmission factor T_p to scatter transmission factor T_s.

3. Simulation setup

The setup used for this simulation (Fig. 1(a)) is based on a typical mammography machine (Boone et al., 2002). The breast equivalent phantom contains 50% adipose tissue and 50% glandular tissue composition (Hammerstein et al., 1979). The phantom includes 5 mm skin equivalent tissue at the entrance

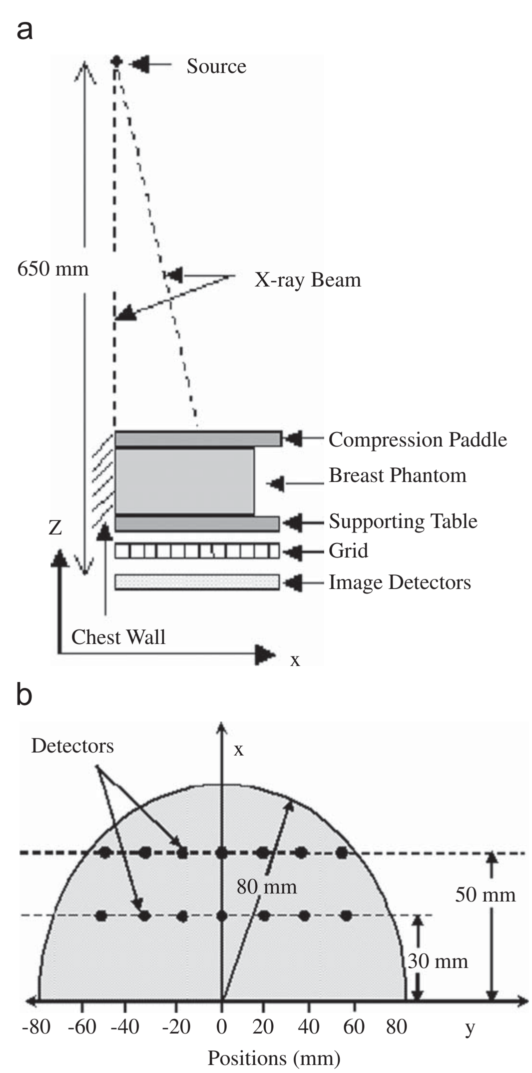

Fig. 1. (a) Simulation setup geometry to mimic the mammography machine, (b) the detector positions relative to breast phantom (top view).