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A method for calculating effective lifetime risk of radiation-induced cancer from screening mammography

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ABSTRACT

Purpose: To propose a method for evaluating the effective lifetime risk of radiation-induced cancer from screening mammography and to present initial data for the UK National Breast Screening Programme.
Material and methods: The imaging was undertaken using a Hologic Selenia full field digital mammographic unit. The proposed method utilises an ATOM phantom containing thermoluminescent dosimeters and a perspex-polyethylene breast phantom to measure organ doses during a standard four view screening mammogram. Effective dose was calculated and effective risk was modelled for a range of client ages. The total lifetime effective risk was then calculated for the UK national screening programme. Calculation of effective risk includes the radiation dose to examined and contralateral breasts in addition to other body organs; this is an advantage over the mean glandular dose.
Results: The contralateral breast, thyroid, thymus, brain, lung, salivary glands, and bone marrow all receive more than 1 μGy radiation dose during screening mammography. A major difference exists for total effective lifetime risk of radiation-induced cancer between clients with average and high breast cancer risk. Differences are attributed to the commencement age of screening and time interval between screens.
Conclusion: This study proposes a method to evaluate effective lifetime risk of radiation-induced cancer from screening mammography in order to compare different mammography screening programmes.
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Introduction

Medical imaging represents the major source of man-made ionizing radiation for people.[1,2] In the United Kingdom (UK), mammography is the sixth largest source of ionizing radiation to the population from diagnostic imaging.[3] However, the benefits of screening mammography are reported to outweigh the risks.[4,5]

Since the glandular tissue is the most radiosensitive component of the breast tissue, the risk of radiation induced cancer from mammography is generally related to mean glandular dose (MGD).[6] MGD is utilised as a standard quantity in breast dosimetry and is recommended by the International Commission on Radiological Protection (ICRP), the National Council on Radiation Protection (NCRP), and the Institute of Physics and Engineering in Medicine (IPEM). MGD can be used as a parameter to evaluate the mammographic system performance, patient risk assessment, and different mammographic imaging techniques.[7] Accordingly, it is of great interest to a large number of researchers.[5,8–12]

Surprisingly few investigators have considered the radiation dose received by body organs and tissues, other than the breast during mammography. They have investigated the dose to other tissues utilising mathematical models to simulate mammography.[13–15] Direct measurement of the radiation dose to the skin overlying the thyroid has been investigated using thermoluminescent dosimeters (TLDs) during screening mammography.[16] In the study by Whelan, McLean, and Poulos[16] they considered that the thyroid dose, which was 0.04 mGy, was insignificant compared to the 4 mGy dose received by the breast. Hatziioannou and colleages[17] also utilised TLDs, accommodated inside an upper body anthropomorphic lucite phantom for measurement

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of dose to the breast, sternum red bone marrow (SRBM), thyroid, liver, lung, stomach, and oesophagus during screening mammography. They found that the breast dose contributes over 98% of the overall effective dose. SRBM and thyroid receive a radiation dose between 0.4–1.27 and 0.05–0.17 mGy/mA, respectively, whereas the other organ doses were negligible. In summary, there is evidence to suggest that the radiation dose received by organs other than the breast requires further consideration as this dose and the risk associated with it is not captured by MGD. For a more thorough and accurate estimate of the risk from radiation during mammography, the dose to other organs should be taken into account.

The internationally accepted method of estimating risk from an X-ray procedure is to use effective dose. Effective dose has enabled doses to be summed from whole and partial body exposure from external radiation of various types to estimate the risk of cancer development. Recently there is a new trend to replace effective dose by effective risk.[18] Calculation of effective dose depends on tissue weighting factors which are regularly updated by ICRP based on the available evidence from epidemiological data.[19] However, effective dose does not take into account an individual's age or reproductive capacity. The effective lifetime risk of developing cancer is less for people who have 20 years to live compared to those who have 60 years to live. Therefore, effective risk includes age, and gender lifetime-attributable risk of cancer incidence per unit equivalent dose.[18]

The lifetime attributable risk of radiation induced breast cancer from mammography has been calculated by Hendrick in 2010[20] and more recently by Yaffe and Mainprize in 2011.[21] However, these authors did not progress their work to include the effective lifetime risk of radiation induced cancer from screening mammography. Based upon the UK National Health Service Breast Screening Programme (NHSBSP), our study proposes a new method for the calculation of effective lifetime risk of radiation induced cancer during screening mammography for females across the screening age range (47–73 years average risk clients and 40–73 years high risk clients).

Method

This was an experimental approach for the calculation of effective lifetime risk of radiation-induced cancer from screening mammography for different client ages, from 40 to 73 years old. It required an accurate measurement of radiation dose received by the examined breast and other body tissues.

Measurement of organ dose

280 Harshaw TLD-100H dosimeters (Thermo Scientific, USA) were placed inside a CIRS adult ATOM dosimetry phantom (CIRS Inc, Norfolk, Virginia, USA) to measure the absorbed radiation dose to tissues and organs during screening mammography (craniocaudal [CC] and mediolateral oblique [MLO] for each breast). TLDs-100H were chosen because they have high sensitivity, wide dose range (1 pGy–10 Gy), and low fading rate.[22] In order to avoid any residual charge, the TLDs were annealed at 240 °C for 10 min before use.

According to the European Commission[23] the total uncertainty in dosimetric results by TLD should be less than 10%. Therefore, the TLDs sensitivity and consistency were established. In terms of sensitivity, all TLDs were exposed three times and according to their average response they were divided into five groups. The sensitivity difference for each group was less than 3%. For TLD consistency estimation, all TLDs were exposed and read three times with time intervals of around five days between the exposures then TLD responses were analysed using SPSS 20.0 (IBM, Armonk, New York, USA) to determine TLD consistency (Intra-class Correlation-Consistency). The calculated consistency was 99%. Consequently, in our work, the total uncertainty of dose results was 4%. The average background signal of three unexposed TLDs was subtracted from the readings of exposed TLDs.[24] As described by Tootell, Szczepura, and Hogg (2013),[25] the TLDs were calibrated against Unfors Multi-O-Meter solid state detector (Billdal, Sweden) on three slabs (1 cm thick each) of Perspex scatterer with beam quality the same as that which was used for dose measurement in the experiment, using the same mammography machine. Usually the calibration process is achieved for a complete batch because the calibration of each individual TLD is time consuming and shows minimal improvement in accuracy (the sensitivity difference of TLDs was less than 3%). For greater precision, the dose-TLD response curve was utilised to obtain the TLD calibration factor.[26]

A breast phantom described by Bouwman et al. (2013)[27] was then used as standard breast. According to the design of this phantom, for simulating the standard breast, which is 53 mm thick, a 32.5 mm thick Poly Methyl Methacrylate (PMMA) and 20.5 mm thick polyethylene (PE) slabs were used; the PE slabs were placed at the top of the phantom. The shape and area of PMMA-PE phantom depends on the mammographic projection of the simulated breast. For the average breast in CC projection, the shape of compressed breast is approximately semicircular with 95 mm radius. However, because the pectoral muscle is included in MLO projection a rectangular phantom with 100 mm × 150 mm was used with the required thickness of PMMA (32.5 mm) and polyethylene (25.5 mm).[28,29]

The ATOM and breast phantoms were positioned on a Hologic Selenia full field digital mammography system (Bedford, USA) to simulate the real woman's position during screening mammography.

For MLO position simulation, firstly, the gantry was tilted 47°. This value of angulation is determined depending on ATOM phantom body contour. Secondly, the MLO phantom was centred on the detector in the chest wall side and the compression paddle was used to fix it in position. Thirdly, the ATOM phantom was arranged in contact with the breast phantom where the midpoint in the side of the breast phantom coincides with the centre point in the breast site on the ATOM phantom. The detector was placed against the ribs with its corner in the axilla.

The CC position was achieved with the X-ray beam perpendicular to the floor. Initially, the CC breast phantom was centred on the detector in the chest wall side and fixed in position by the compression paddle. The midpoint in the side of the breast phantom was arranged to coincide with the centre point in the breast site on the ATOM phantom, which faced the mammography system.[30]

Since automatic exposure control (AEC) is the most commonly used technique in screening mammography, it was adopted to expose the breast phantom. For each position the phantom was exposed three times and then averaged (CC and MLO for each breast) to minimise random error. TLDs were then collected and read in a Harshaw 3500 TLD reader (Thermo Scientific, USA). Using the manufacturer's TLD location map,[31] radiation absorbed dose of each organ was calculated by averaging the radiation dose values inside the organ. In addition to contralateral breast, doses received by 19 other organs were measured. The numbers of TLDs used for each organ are listed in (Table 1). The whole process was repeated three times on different days in order to examine the reproducibility of the data.

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