There has been much recent debate about the benefits of routine screening mammography. However there has been rather less discussion regarding possible radiation-related risks associated with these examinations, specifically the risk of radiation-induced breast cancer, although some risk-benefit analyses have been reported.

Glandular examination doses for screening mammography are small, typically around 3 mGy of 26-30 kVp low-energy x rays. A particular issue here, however, is that these very low-energy x rays are expected to be more hazardous, per unit dose, than higher-energy x or gamma rays (i.e., those on which radiation risk estimates are based, such as from the Japanese A bombs). The underlying biophysical reason for the expected increase in biological effectiveness for these lower-energy x rays is that they set in motion slower secondary electrons, with correspondingly higher LET.

An increase in the relative biological effectiveness (RBE) of low-energy vs. high-energy photons is of relevance in assessing the risk side of the benefit-risk equation for mammography, in that the radiation-related risks are calculated based on studies of populations (A-bomb survivors and women who received multiple fluoroscopies) exposed to higher-energy photons.

Of course the significance of any enhancement in the biological effect of mammographic x rays depends on its magnitude: We have measured in-vitro oncogenic transformation frequencies in C3H10T1/2 cells, induced by monoenergetic x rays in the 15 to 25 keV range, produced at the Brookhaven National Synchrotron Light Source (NSLS).

The studies present unique dosimetric problems: The dominant mechanism for the interaction of X rays with matter at energies below 25 keV is photoelectric absorption. Because the cross section for this effect varies approximately as Z^4, measurements of microdosimetric spectra can be distorted by the presence of even small amounts of metals in the beam (e.g. center wire and helix) that produces a spectrum of primary electrons significantly different from that in tissue. At 25 keV, half the photon cross section for muscle tissue is due to scattering while for iron it is only 4%, the remainder in both cases being due to the photoelectric effect. The photoelectrons have an energy of 25 keV but Compton scattering produces electrons with a distribution of energies having a maximum of 2 keV. The presence of iron increases the average secondary electron energy and therefore reduces the mean energy deposition. Secondary electrons at these low energies have curvilinear tracks and it is estimated that ~20% of all energy deposition events in a walled counter would be due to re-entry wall effects.
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In order to minimize these problems, a wall-less microdosimetry chamber in which only low-Z materials are exposed to the X ray beam or to the electrons generated by the beam has been designed and built. The sensitive volume of the proportional counter is a cylinder 3.2 mm in diameter and 3.2 mm high defined by two pairs of helices made of C552 air-equivalent plastic. The struts that form the helical structure have a cross section of 0.1 mm x 0.1 mm and are a combination of clockwise and counter-clockwise helices of different pitch (14 and 12 turns in 2.54 cm) staggered so that no two intersections are at 180° to each other. The helices are made as one piece from a single mould and are joined at their intersections. Four helices are used to obtain rigidity of the structure and better define the active volume. The helices will be biased with a negative voltage to repel low-energy electrons from outside the sensitive region of the detector.

The anode wire is a smooth single carbon fiber 5 μm in diameter, which should provide adequate gas gain at moderate voltage. Field shaping tubes made of C552 plastic with an outside diameter of 0.9 mm are used to define the ends of the sensitive volume. In each of the end supports the anode is insulated from the field tube by a micro-fused quartz, tube with an outside diameter of 0.15 mm inside a cross-linked styrene copolymer tube (Rexolite 1422), which has a high volume resistivity. A circular frame made of Rexolite supports the counter assembly.

The detector housing and rear cover are made of acrylic. The front window of the housing is polyvinylidene fluoride (PVDF), which has a mass energy absorption coefficient similar to that of muscle tissue and is the substrate on which cells are plated for irradiation. The window is 3.8 mm thick to withstand an atmosphere of pressure with little deflection when the chamber is evacuated. This wall thickness does not pose a problem since the unattenuated synchrotron beam is too intense for single event spectra even when only a narrow X ray energy band is selected. The entire inner surface of the housing, rear cover and front window are covered with a very thin layer of carbon to provide a conductive surface that will be connected to ground.

All other interior parts are made of Rexolite because of its high radiation resistance, rigidity and dimensional stability. The counter support frame is mounted on rails and can be moved so that the frame is in contact with the front window or as far away as 8 mm. In these positions the center of the counter is within 6.4 mm and 14.4 mm of the inner window surface respectively, simulating depths of 2 and 4.5 μm in tissue when the counter is operated at a 1 μm site size.
A completely new tissue-equivalent gas has been designed for use with low energy photons. Gases that have been used up to now (TE propane, TE methane, air) have a ratio of mass-energy absorption coefficient to that of cell nucleus which varies very strongly with energy. We have designed and tested 2 new neon-based TE gases:

7.1% CO₂, 2.6% N₂, 10.9% propane, 0.3% argon, 40.9% methane, balance neon

and

3.4% N₂, 39.6% propane, 0.3% argon, balance neon.

Both these gases have excellent properties down to photon energies of about 5 keV, with ratios of $\mu_m / \rho$ to cell nucleus better than 0.99. In addition both these gases work excellently up to photon energies of about 30 MeV, making them of general utility.

Calibration of the counter is done with a $^{244}$Cm-activated Al Kα source. The electrons generated by the 1.49 keV photons have a range in water of only 100 nm and thus will all be absorbed in a 1 μm site size. The source is attached to a rod at the back of the chamber and can be rotated and moved forward so that it is centered very close to the counter, in order to have maximum flux. When not in use, the source is stored at the bottom rear of the counter, out of the X ray beam.

Transformation data for 15.2 keV monoenergetic x rays are shown in Fig. 2. Using linear-quadratic fits (see Fig. 2) to the low-dose data, we estimate a low-dose RBE (ratio of $\alpha$ terms) of 1.96±0.78 for 15.2-keV x rays vs. 662-keV $^{137}$Cs γ rays. In no case have we estimated a low dose RBE (defined, as above, as the ratio of $\alpha$ terms) of greater than 1.5 relative to 250-kVp x rays, or greater than 2.5 relative to $^{137}$Cs γ rays.

Fig. 2. Measured induced oncogenic transformation frequencies per 10⁴ surviving C₃H₁₀T₁₂% cells, as a function of the dose of 15.2-keV monoenergetic x rays and 662-keV $^{137}$Cs gamma rays. Estimated 68% confidence limits are shown. For clarity, only low-dose data points are shown. Curves represent fits to the full data set using the model $TF_{15keV} = \alpha_{15keV}D + \beta D^2$ and $TF_{662keV} = \alpha_{662keV}D + \beta D^2$. 

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These fairly modest RBE estimates are consistent with the earlier experimental data, as well as our theoretical estimates (Brenner and Amols 1989) of 1.3 (vs. 250-kVp x rays) and 2.0 (vs. gamma rays at Hiroshima and Nagasaki), for the low-dose RBE of 23-kVp filtered x rays. The reason for the comparatively small predicted enhancements in effectiveness at mammographic x-ray energies is that the differences in energy deposition patterns between the higher and the lower-energy photons are relatively subtle.

We stress, however, that even if the risks per unit dose of mammographic x rays are just twice as large as those from the radiation at Hiroshima and Nagasaki, this would be of some significance. For example, Fig. 3 shows the age-dependent risk-benefit ratio, as estimated in NCRP Report 85 (1986) for 5 annual mammograms, each producing a glandular dose of 2 mGy. Here the "benefit" is assumed to be a 10% decrease in mortality, and the risk of radiation-related breast cancer was appropriately derived from studies of the Japanese A-bomb survivors. Now if it is assumed that mammographic x rays are twice as hazardous, per unit dose, than the radiations at Hiroshima and Nagasaki, the benefit-risk ratio would also be decreased, by this same factor of 2.

![Figure 3](image_url)

**Fig. 3.** Dotted curve: Estimated benefit-risk ratio for yearly mammographic screening examinations for 5 years, assuming a total glandular dose of 2 mGy per exam. The benefit is assumed to be a 10% reduction in breast cancer mortality, and the radiation risk was based on a 0.042% per mGy increase in the spontaneous breast-cancer rate. Further details of the calculation are given in NCRP (1986).

Solid Curve: Corresponding benefit-risk ratio in which the estimated radiation risk is doubled. Arrows indicate the change in age at which a given benefit-risk ratio would be achieved, assuming the radiation risk were doubled; these suggest that recommended starting ages for routine mammography might reasonably be increased, if the radiation risk from mammographic x rays was larger than that previously assumed.
Such a reduction in the benefit-risk ratio would be reflected in the age after which annual screening is recommended. For example, the American Cancer Society recommendation to begin annual screening at age 40 corresponds to the age when the estimated benefit-risk ratio reaches an acceptable value (numerically equal to 7 in Fig. 3). If the benefit-risk ratio were halved, then the age at which this same benefit-risk ratio is reached would be increased (see Fig. 3), in this case from age 40 to 47. Similarly, if annual screening were recommended from age 50, doubling the radiation risk, while keeping fixed the benefit-risk ratio for commencement of screening, would imply an increase in the recommended age to begin screening, from age 50 to 60 (Fig. 3).

In summary, there is evidence that low-energy x rays as used in mammography have an increased biological effectiveness relative to higher-energy photons. However the RBE values are not large, probably less than 2. Thus it is unlikely that the radiation risk alone could prove to be a "show stopper" regarding screening mammography because, for older women, the benefit is likely to considerably outweigh the radiation risk. Nevertheless, the RBE for low-energy x rays might reasonably be taken into account when assessing the recommended age to commence such annual screening.

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