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OPTIMIZATION OF SIDE SHIELDING FOR CIRCULAR POSITRON EMISSION TOMOGRAPHS

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OPTIMIZATION OF SIDE SHIELDING FOR CIRCULAR POSITRON EMISSION TOMOGRAPHS

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ABSTRACT

This report presents expressions for the image-forming and background event rates seen by circular positron emission tomographs. These are used to determine the side shielding depth that optimizes the signal-to-noise ratio in the reconstructed image of a 20-cm cylinder of water with uniformly dispersed activity. For 1-cm wide NaI(Tl) detectors, a 50-cm patient port, an activity of 200 μCi per axial centimeter, and a shielding gap of 2 cm, the optimum shielding depth is 20 cm which results in a detector circle diameter of 90 cm. For a 25-cm patient port and other conditions as above, the optimum shielding depth is 14 cm. Optimization calculations for detector materials having different efficiency, energy resolution, and time resolution are also presented.

In positron emission transverse section tomography, the image is derived from the detection of unscattered coincident annihilation pairs. Almost all positron emission tomographs employ shielding on either side of the detectors to block activity external to the transverse section being imaged (1-7). Shielding is also used between layers in multiple section devices (8-12). In spite of this shielding, image contrast is degraded by true coincidences of scattered annihilation pairs and by accidental coincidences of unrelated annihilation photons (Fig. 1). Most positron imaging systems operate with scattered and accidental backgrounds that are each typically 20% of the detected coincidences. Even if these backgrounds can be perfectly estimated and subtracted from the detected coincidences, the statistical fluctuations in the result are greater than if the backgrounds did not exist. In the following sections we examine the trade-off between sensitivity and backgrounds and describe a procedure for determining the optimum shielding depth for circular positron emission tomographs.

EVENT RATES

As has been shown analytically for circular positron emission tomographs (13), the overall rate of unscattered coincident events (C_I) is given by:

$$C_I = B_I \varepsilon^2 \rho G_e^2 / R \qquad (1)$$

where ε is the detection efficiency for annihilation photons, ρ is the activity density in μCi per axial centimeter, G_e is the effective shielding gap in centimeters, R is the detector ring radius in centimeters (Fig. 2), and B_I is a constant that incorporates the average attenuation and numerical factors. For activity distributed in a 20-cm cylinder of water, B_I = 1850. Due to edge penetration, the effective shielding gap G_e is slightly larger than the physical shielding gap G:

$$G_e = G + \delta_G \tag{2}$$

The overall rate of coincident scattered events (C_S) is given by (13):

$$C_S = B_S \varepsilon^2 \rho G_e^3/(RH) \tag{3}$$

where H is the shielding depth in centimeters and B_S is a constant that incorporates pulse height thresholds, the angular distribution of accepted Compton scatters and numerical factors.

The overall rate of accidental events (C_A) is given by (13):

$$C_A = B_A \varepsilon^2 \tau \rho^2 G_e^4/H^2 \tag{4}$$

where τ is the full coincidence time window in nanoseconds, and B_A is a constant that incorporates pulse height thresholds, detector efficiencies for scattered and unscattered photons, and numerical factors.

These rates are reduced by system deadtime, which is a combination of the deadtime of the detectors, timing and pulse height discriminators, coincidence circuits and memory. We assume that for a particular scattering medium (i.e., a 20-cm cylinder of water) the ratio of photon interactions to coincident events is fixed and an effective nonparalyzing system deadtime can be defined that applies to the total coincident event rate only. The fraction of events F that is lost to deadtime is given by:

$$F = tC_T/(1 + tC_T) \tag{5}$$

where C_T is the total coincidence rate ($C_I + C_S + 2C_A$) and t is the deadtime per event. We assume that before deadtime losses the system detects C_I + C_S + C_A in the on-time coincidence window and C_A in an off-time window.

The observed system rates are:

$$D_I = (1-F)C_I$$
$$D_S = (1-F)C_S$$
$$D_A = (1-F)C_A \qquad (6)$$

In the central region of the reconstructed image of a cylinder of activity in water, the intensity of unscattered coincident events per square centimeter (d_I) is given by:

$$d_I = (1-F)b_I\varepsilon^2\rho G_e^2/R. \qquad (7)$$

For a 20-cm cylinder, $b_I = 29.44$. The intensity of scattered coincident events per square centimeter (d_S) is given by:

$$d_S = (1-F)b_S\varepsilon^2\rho G_e^3/(RH). \qquad (8)$$

The intensity of accidental events per square centimeter (d_A) is given by:

$$d_A = (1-F)b_A\varepsilon^2\tau\rho^2 G_e^4/H^2. \qquad (9)$$

Before background subtraction, the total intensity d_T is given by:

$$d_T = d_I + d_S + d_A. \qquad (10)$$

FIGURE OF MERIT

A figure of merit (Q) can be defined as the product of the unscattered coincidence rate (D_I) and the image contrast (d_I/d_T) using the arguments of Beck (14):

$$Q = D_I(d_I/d_T) \qquad (11)$$

Q may also be called an "effective" image event rate since the same signal-to-noise ratio would be obtained in an ideal tomograph with $D_I' = Q$ and $d_S' = d_A' = 0$. Note that d_I, d_S and d_A all undergo the same deadtime effects, attenuation correction, and error propagation in the reconstruction process. The latter results in a significant reduction in signal-to-noise and is discussed in Reference 15.

Equations 1-10 show that for a given imaging situation, reducing the shielding depth H improves the imaging rate D_I but also decreases the image contrast d_I/d_T. Choosing a value of H that maximizes Q (Eq. 11) insures the best tradeoff between sensitivity and image contrast.

RESULTS

<u>NaI(Tl)</u>. The constants ε, B_S, B_A, b_S, b_A, and δ_G were determined for NaI(Tl) by fitting Equations 1-9 to measurements of 20-cm phantoms made by the Donner 280-crystal positron tomograph. The overall rates D_I, D_S, and D_A were measured for a 20-cm diameter cylinder of activity in water (<u>6</u>) and $d_S/(d_S + d_I)$ and $d_A/(d_S + d_I)$ were measured at the center of reconstructed images of a 5-cm diameter cylinder containing only water surrounded by a 20-cm diameter annulus of activity in water (Fig. 2). The pulse height threshold was 100 keV, the activity was varied from 100 to 300 µCi/cm, and the shielding gap was varied from 1 to 3 cm. A good fit was obtained with $\varepsilon = 45\%$, $B_S = 5100$, $B_A = 0.73$, $b_S = 71$, $b_A = 5.3 \times 10^{-3}$, and $\delta_G = 2.1$ mm. The penetration factor δ_G was necessary in each of equations 1, 3, and 4 (D_I, D_S and D_A, respectively) for an adequate fit to the data.

Table 1 lists D_I, d_S/d_I, d_A/d_I, d_I/d_T, and Q as a function of shielding depth H, for a 50-cm patient port, $\rho = 200$ µCi/cm, $G_e = 2$-cm effective shielding gap, a 20-cm diameter water cylinder, and a 1-µsec deadtime. Figure 3 presents curves of Q as a function of H for the same conditions and 25-cm and 50-cm patient ports.

Other detector materials. The relative rates D_I, D_S, and D_A for bismuth germanate (BGO) detector crystals have been measured in this system with a 300-keV pulse height threshold (6), and lead to the constants $\varepsilon = 67\%$, $B_S = 5100$, $B_A = 0.36$, $b_S = 71$, $b_A = 2.6 \times 10^{-3}$. Note that the photopeak selection reduces B_A and b_A but not B_S and b_S since the scatter background consists primarily of photons above 415 keV that have scattered through $<40^{o}$ (5).

By using the efficiency, time resolution and energy resolution for other detector materials, it is possible to optimize the shielding depth for each material. Table 2 lists the results for six classes of detector materials at four activity levels. See Reference 16 for a more extensive tabulation. The detection efficiency for Ge(Li) and plastic was determined from Monte Carlo calculations that traced the interactions of a beam of 511-keV photons through a group of 1-cm wide detectors (17). The detection efficiency was defined as the fraction of incident photons that produced a signal above threshold in only one detector.

DISCUSSION

The highest optimum value of Q is achieved with BGO at each of the four activity levels. The second best materials fall in the general class of CsF, pure NaI (cooled) and liquid xenon which have the same detection efficiency as NaI(Tl) but significantly better time resolution.

Ge(Li) is unique in having sufficient energy resolution to reject almost all tissue-scattered photons. This eliminates the coincident scattered background and greatly reduces the accidental background. The low full-energy efficiency for 511-keV photons, however, results in low values of Q.

Plastic scintillators are unique in having such excellent timing resolution that the coincidence time window of 2 to 3 nsec is determined by the size of the patient port and the speed of light. If time resolutions of the order of $\pm$100 psec could be realized for tomographic systems, then the timing information and the use of shorter detectors could localize the annihilation point along the line of flight to $\pm$1.5 cm. This possibility was suggested by Anger in 1966 (18) and is used in a 3-d imaging system built by Nickels and co-workers (19). The effective sensitivity (Q) could then be improved by about a factor of 10 through reduced error propagation in the image reconstruction process (15) in spite of the shorter, less efficient detectors. As may be seen in Table 2, a tenfold increase in Q would make plastic competitive with BGO.

For the case of wire chambers with lead converters, rather optimistic values of efficiency and time resolution have been used but the resulting Quality Factors are still relatively low. The situation may improve, as several groups are investigating schemes to improve their properties (20-22).

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Table 1.
SHIELDING OPTIMIZATION FOR NaI(Tl)[a]

H Shielding Depth (cm)	D_T Observed Total[b] (sec^{-1})	F Deadtime Loss (%)	D_I Image Event Rate (sec^{-1})	d_S/d_I Scatter/ Image Ratio	d_A/d_I Accidental/ Image Ratio	d_I/d_T Image Contrast	Q Quality Factor (sec^{-1})
5	118,582	11.9	8,805	0.97	2.59	0.22	1,932
10	40,000	4.0	8,220	0.48	0.76	0.45	3,672
15	22,350	2.2	7,325	0.32	0.38	0.59	4,295
20[c]	15,352	1.5	6,558	0.24	0.24	0.67	4,418
25	11,718	1.2	5,924	0.19	0.17	0.73	4,337
30	9,513	1.0	5,397	0.16	0.13	0.77	4,175
40	6,971	0.7	4,579	0.12	0.09	0.83	3,789
50	5,541	0.6	3,974	0.10	0.06	0.86	3,422
60	4,617	0.5	3,510	0.08	0.05	0.88	3,102

[a] port = 50 cm, effective shielding gap G_e = 2 cm, pulse height threshold = 100 keV, detector efficiency ε = 45%, ρ = 200 μCi/cm, deadtime t = 1 μsec, coincidence time window τ = 15 nsec.

[b] Total rate in on-time and off-time coincidence windows combined.

[c] H = 19.8 cm for maximum Q.

Table 2.
OPTIMUM VALUES OF H AND Q FOR VARIOUS DETECTOR MATERIALS[a]

	NaI(Tl)	BGO	CsF NaI (cooled) Liquid Xe	Ge(Li)	Wire Chambers	Plastic
pulse height threshold	100 keV	300 keV	100 keV	505 keV	200 keV	100 keV
511 keV efficiency: ε	45%[b]	67%[b]	45%[b]	15%[b]	20%	20%[b]
Coincidence time window: τ	15 ns	30 ns	3.3 ns	15 ns	50 ns	3.3 ns
B_S	5,100	5,100	5,100	0	5,100	5,100
B_A	0.73	0.36	0.73	0.15	0.50	0.73
b_S	71	71	71	0	71	71
b_A	5.3×10^{-3}	2.6×10^{-3}	5.3×10^{-3}	1.0×10^{-3}	3.5×10^{-3}	5.3×10^{-3}
ρ = 100 µCi/cm						
Shielding depth: H (cm)	16.3	16.5	12.7	6.9	19.8	12.5
Image contrast: d_I/d_T	0.68	0.69	0.70	0.88	0.66	0.69
Quality factor: Q (sec^{-1})	2,460	5,417	2,750	459	435	546
ρ = 200 µCi/cm						
Shielding depth: H (cm)	19.8	20.3	14.1	8.9	24.7	13.8
Image contrast: d_I/d_T	0.67	0.68	0.70	0.85	0.63	0.69
Quality factor: Q (sec^{-1})	4,419	9,648	5,263	835	750	1,051
ρ = 500 µCi/cm						
Shielding depth: H (cm)	27.6	29.4	17.6	12.5	34.6	16.7
Image contrast: d_I/d_T	0.65	0.67	0.69	0.80	0.58	0.68
Quality factor: Q (sec^{-1})	8,846	18,858	11,799	1,775	1,421	2,399
ρ = 1000 µCi/cm						
Shielding depth: H (cm)	37.9	42.1	22.4	16.4	46.4	20.4
Image contrast: d_I/d_T	0.63	0.66	0.69	0.76	0.53	0.67
Quality factor: Q (sec^{-1})	13,827	28,561	20,473	3,031	2,149	4,284

[a] Port = 50 cm, effective shielding gap G_e = 2 cm, deadtime t = 1 µsec per coincident event

[b] Detector size 1 cm wide x 5 cm deep

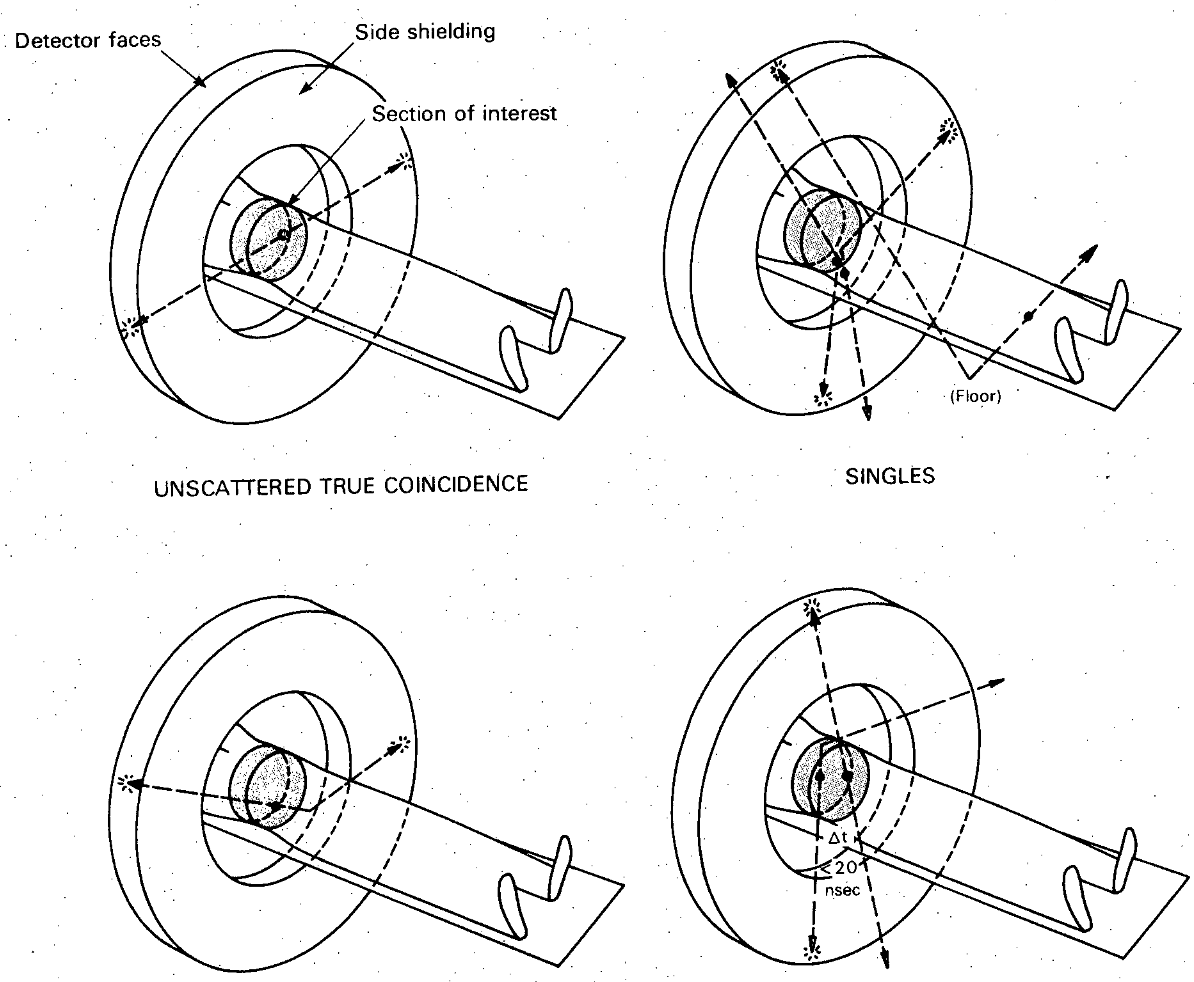
FIGURE CAPTIONS

FIG. 1. Coincident event types detected by positron tomographs. The image is formed from unscattered coincident annihilation pairs. Coincident scattered pairs and accidental coincidences of unrelated photons result in a broad background.

FIG. 2. Detector, shielding, and phantom geometry for optimization calculations. Scatter background can be measured in the inner cylinder by imaging with activity in the outer cylindrical annulus only.

FIG. 3. Effective event rate Q as a function of shielding depth for NaI(Tl) detectors with 45% detection efficiency, a 2-cm shielding gap, 200 μCi/cm, 1-μsec deadtime, and a coincidence resolving time of 15 nsec. Optimum values are indicated with arrows for 25- and 50-cm patient ports.

FATES OF ANNIHILATION PHOTONS



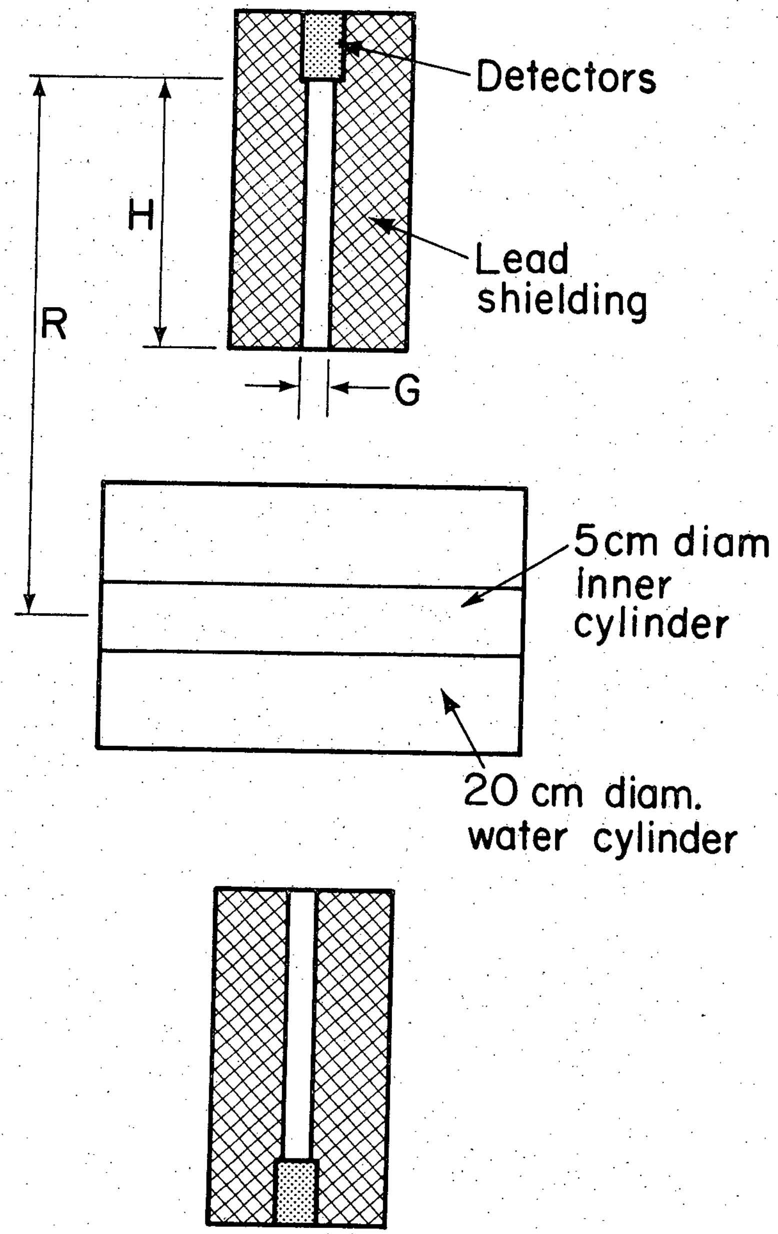
UNSCATTERED TRUE COINCIDENCE

SINGLES

SCATTERED TRUE COINCIDENCE

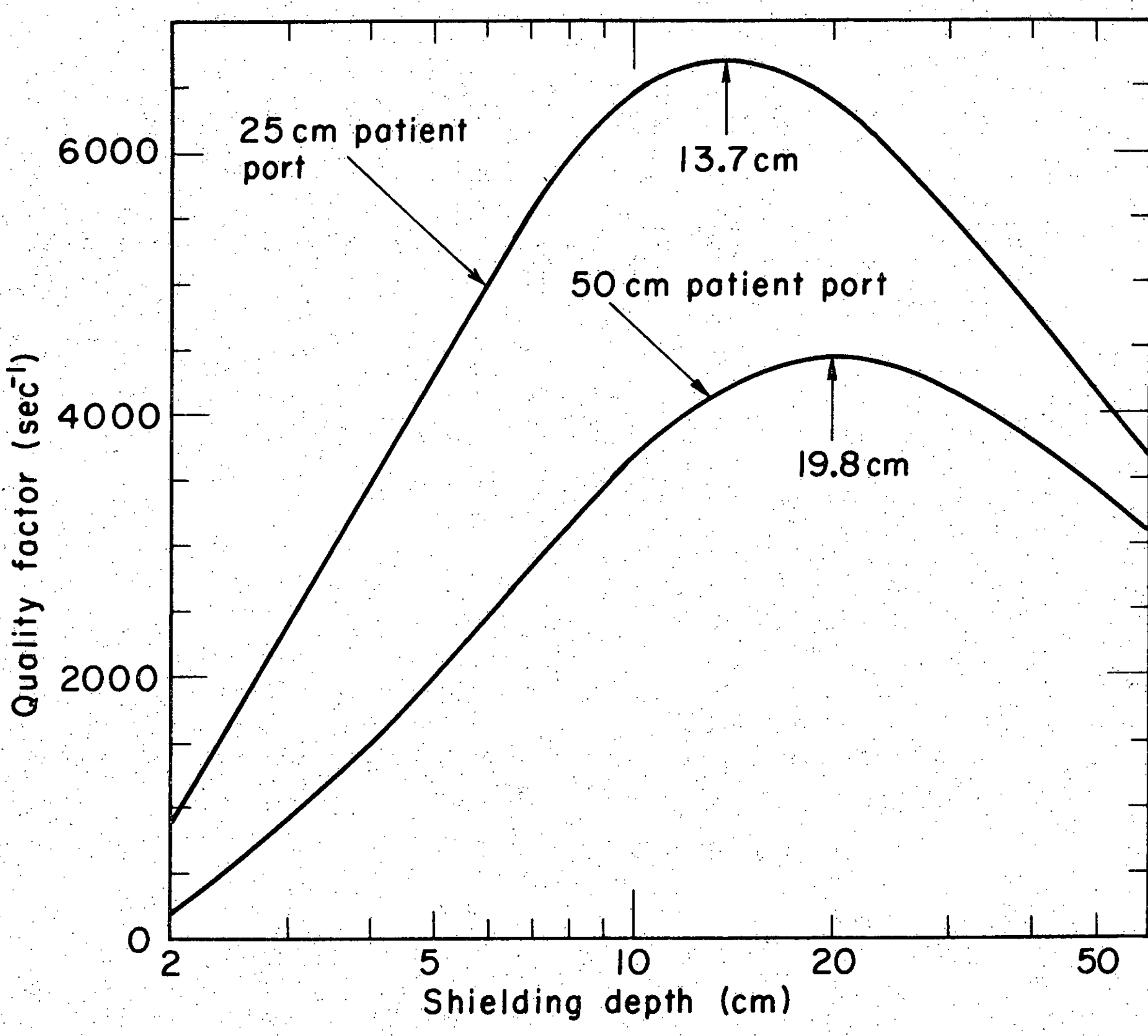
ACCIDENTAL COINCIDENCE

XBL797-3627



Detectors
H
R
Lead
shielding
G
5cm diam
inner
cylinder
20 cm diam.
water cylinder

XBL797-3625


25 cm patient port
13.7 cm
50 cm patient port
19.8 cm
Quality factor (sec^{-1})
6000
4000
2000
0
2
5
10
20
50
Shielding depth (cm)

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