QUANTITATIVE X-RAY FLUORESCENCE MEASUREMENT OF THE STABLE IODINE IN THE THYROID GLAND

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In-vivo detection of the thyroidal iodine content (TIC) by X-ray fluorescence (XRF) was first suggested (1) in 1968. This method has not yet been applied clinically, mainly because of its limited accuracy, about 50% in existing experimental systems (2,3), while it is expected (4) that various thyroid conditions can be differentiated only when the estimation error does not exceed 20-30%.

The low accuracy of the existing experimental systems stems mainly from the geometrical differences which naturally exist between the calibration phantom and the various patients. This results in an uncertainty in the actual distance of the thyroid from the XRF unit and also in the thickness of the tissue covering the thyroid, thus introducing errors in the estimation of absorption and dispersion of the radiation involved.

In order to obtain an improved estimation of the effective depth of the thyroid we examined two methods.

a) The energy difference between the iodine K_α and K_β lines (28.5 and 32.3 keV, respectively) results in a difference of 7% in their absorption per cm tissue (5). Consequently it is possible to obtain the effective depth of the thyroid from the intensities ratio K_α/K_β.

The effective depth T is given by Eq. (1)

$$T = [\ln I(K_\alpha) - \ln I(K_\beta) - \ln(A)]/[\mu(K_\beta) - \mu(K_\alpha)] \quad (1)$$

where $I(K_\alpha)$ and $I(K_\beta)$ are the intensities of the K_α and K_β lines, $\mu(K_\alpha)$ and $\mu(K_\beta)$ are the respective attenuation coefficients and A is the unattenuated K_α/K_β ratio (6).

In phantom measurements that we conducted, simulating normal human conditions, the signal-to-background ratio was 1:5 at K_α and 1:27 at K_β. This relatively high background is due to the strong scattering of the Am radiation by the neck. Thus, if an error of less than 20% is called for, it can be shown by conventional error propagation calculations that the number of counts in the K_α region must exceed 10^6. This would result in a dose of about 2 rads to the patient (as measured by a LiF crystal dosimeter attached to the front of the phantom).

b) The XRF intensity is naturally distance-dependent. Hence, the effective depth of the thyroid can be obtained by comparing the XRF intensity of measurements at two distances.

We developed a computer simulation of our system in which the thyroid lobes had either a cylindrical or an ellipsoidal shape, and their dimensions and separation were the variable parameters. Volume varied from 15 to 40 cm^3. (The normal volume is 20 cm^3 (4)). In this simulation we assume a $1/R^2$ radiation dispersion from each point of the Am source to each point of the thyroid, and from each thyroid point to each point of the Si(Li) detector.

The distance dependence is expressed in Eq, (2)

$$\frac{I_1}{I_2} = \left(\frac{D_2 + T + C}{D_1 + T + C}\right)^p \qquad (2)$$

where I_1 and I_2 are the XRF intensities at the distances D_1 and D_2, respectively, the constant C has the value 1.2, p is taken as 3.54 and T is the effective depth of the thyroid.

Equation (2) is valid under the following conditions:

i) $6cm < D_1 < D_2 < 10$ cm where D_1 and D_2 are the neck-to-XRF unit distances.

ii) $D_2 - D_1 < 2$ cm.

iii) the diameter of the annular source (Fig. 1) is 4-6 cm.

Under these conditions we obtained a ∿20% error (after depth correction) in TIC measurements on a phantom (Fig. 1), with ∿ 10^4 counts in the K_α region. The dose to the patient is ∿ 20 mrad (measured by a TLD dosimeter).

The system is presently being applied to clinical measurements and the results will be reported elsewhere.

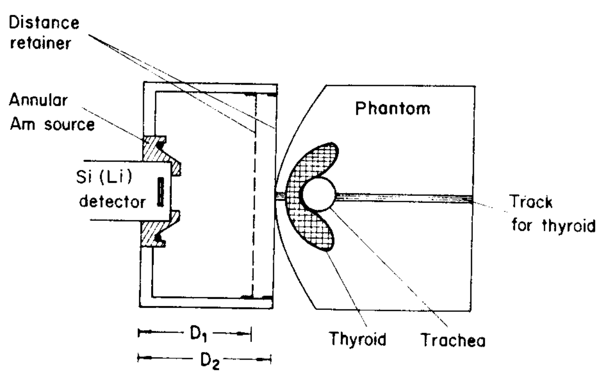

Fig. 1. XRF system for TIC measurements: The XRF unit consists of a 5 cm diam, 40 mCi ^{241}Am source and a 200 mm^2 area, 5 mm depth Si(Li) detector. The phantom consists of a polyethylene exterior and is water filled. The "thyroid" slides on a track for depth variation.

References:

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