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OBJECTIVE SYMMETRY DETECTOR METHOD FOR GAMMAENCEPHALOGRAPHY

I. Physical characteristics

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Various methods have been used to measure and evaluate the distribution of isotope tracers in the skull. Digital recording of the count rate, obtained from measurements with a single detector and a low resolution collimator placed at different positions around the head, was the first method used in gammaencephalography (MOORE 1948). High diagnostic accuracy was reported when the symmetry of isotope distribution in the skull was analysed (PLANIOL 1959, 1966, MUNDINGER & ASAI 1967).

However, detector positioning over symmetrically located regions of the skull was found to be intricate, implying a source of error difficult to evaluate. The interpretation of the results was time consuming. Nowadays, therefore, symmetry methods are only in use in a few hospitals in the world (PLANIOL 1966, BURROWS 1972).

Conventional image producing methods have improved during recent years, having benefited from technical progress including computer processing, high resolution collimators and short-lived isotopes, all in order to obtain more detailed information on the isotope distribution. This has promoted the development of more

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complicated recording systems. For economic reasons, however, these are often designed to suit a wide range of requirements. It is therefore not always possible to take full advantage of the improvements in special clinical applications, since the choice of measuring and evaluation parameters may be restricted by the equipment. A system specially designed for a specific purpose may thus be less expensive while still offering more relevant parameters.

A further development of the gammaencephalographic methods, originally described by PLANIOL and MUNDINGER, based on the symmetrical isotope distribution in the skull therefore seemed desirable. The diagnostic accuracy (obtained by MUNDINGER and PLANIOL) is difficult to evaluate because of the unknown selection of patients and a partly subjective interpretation of results which involved added clinical information. The different sources of error, influencing the reproducibility, were not reported in detail but the results must have been influenced by large uncertainties from the counting statistics and from the positioning of the detectors (PLANIOL 1966, MUNDINGER & ASAI).

The measuring performance and the evaluation of the results of the method now described can be standardized in order to obtain an objective classification of the clinical results with high accuracy. The significance of a deviation from the normal mean should be used as the basis for diagnosis, thus avoiding subjective interpretation of results (LARSSON & LIND, LIND et coll.)

The equipment has been designed to be simple to handle and inexpensive, and the standardization of the measuring performance should make the method well suited to routine clinical work, including screening of large materials, even outside large medical centres (GOTTSCHALK 1971). The purpose of this report is to describe the physical characteristics of the equipment and the method.

Principles of the method. The method is principally based on a precise determination of the relative isotope distribution in different predetermined parts, 'subregions', of the skull, following intravenous administration of $^{99}Tc^{m}$-pertechnetate.

Two equally collimated NaI-detectors, directed towards each other on a common axis and arranged symmetrically on the right and left sides of the skull, measure the γ-radiation emitted from the isotope in a subregion. A complete examination of the whole skull necessitates measurements in turn over each of several different subregions, having the same size (lateral projection), determined by the collimator design. Their locations, in relation to the length and height of the skull, are made reproducible both within and between individuals by means of the detector movement. By this means, the results obtained can be compared both between the right and left parts of each subregion, and also between different subregions of the same individual. The relative evaluation parameters obtained permits objective comparisons between individuals.

In order to illustrate the characteristics of the different evaluation parameters and the influence on the results of different physical factors, theoretical and experi-

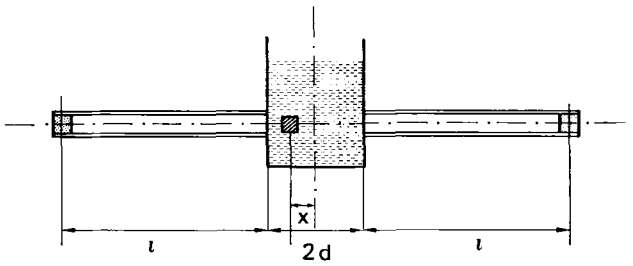

Fig. 1. Phantom system and detector arrangement.

mental modelling has been performed. The skull was represented by a lucite box, containing a homogeneous solution of $^{99}Tc^{m}$-pertechnetate and water (Fig. 1).

Theory. Let the two detectors be located over the right (R) and left (L) side of a subregion of the skull phantom. If the two detection-units are assumed to be identical, the results obtained will be the same for both and hence, only one of the detectors has to be considered for the following calculation.

The count rate from a thin sheet of homogeneously distributed activity is approximately independent of the distance in air from the sheet to the collimator aperture, as long as the sheet extends outside the 'field of view' of the detector (MAYNEORD 1950, BROWNELL 1959, BECK 1964). The degree of approximation depends mainly on the relationship between the length and the aperture of the collimator.

If the thickness of the walls (5 mm) of the phantom is neglected, the phantom may be described as consisting of a stack of thin sheets, each of them with a thickness Δx. Hence, assume a sheet located at a distance $(d+x)$ from the collimator aperture. From geometrical considerations, each point of the detector area can be shown to view approximately an area A_w of the sheet given by:

$$A_w = \frac{A(l+d+x)^2}{l^2} \tag{1}$$

where A is the area of the collimator aperture and l, the length of the straight collimator (Fig. 1). ΔQ (μCi) is the activity of the amount of isotope in the volume $\Delta V = A_w \Delta x$, which is recorded by the detector and can be expressed:

$$\Delta Q = C\Delta V = CA_w\Delta x \tag{2}$$

where C (μCi/cm^3), the activity per cm^3, represents the concentration of isotope in the phantom. If scattering and penetration effects are neglected, the fraction G(x), of photons (generated within ΔV) which can reach the detector, can be calculated according to the inverse square law:

$$G(x) = \frac{A}{4\pi(l+d+x)^2} \tag{3}$$

The symbol A in eq. (3) represents the area of the detector, which in the case of a straight collimator, equals the area of the collimator aperture.

When the attentuation of photons, the limited detector efficiency ε_d and the data-transfer efficiency ε_{dt}, are taken into account, the number of registered events/s ΔR, from the activity ΔQ, can be expressed, assuming k_1 γ-photons to be emitted per decay:

$$\Delta R = 3.7 \cdot 10^4 k_1 \varepsilon_d \varepsilon_{dt} \Delta Q G(x) \exp(-\mu(d+x)) \tag{4}$$

By setting the results of eqs. (1), (2) and (3) into eq. (4) the following result is obtained:

$$\Delta R = \frac{3.7 \cdot 10^4 k_1 \varepsilon_d \varepsilon_{dt} A^2 C}{4\pi l^2} \exp(-\mu(d+x)) \Delta x \tag{5}$$

By setting the thickness of the whole skull phantom = 2d and the location variable x = 0 in the median plane, the total number of registered events/s R, from all the activity within one subregion is obtained by integration of eq. (5) from x = −d to x = +d:

$$R = \frac{k_0 A^2 C}{l^2} \int_{-d}^{+d} \exp(-\mu(d+x)) dx = \frac{k_0 A^2 C}{\mu l^2} (1 - e^{-2\mu d}) \tag{6}$$

where

$$k_0 = \frac{3.7 \cdot 10^4 k_1 \varepsilon_d \varepsilon_{dt}}{4\pi}$$

When a measurement is assumed to be performed during a time T, much shorter than the half-life of the isotope used, the mean number of registered events N from the detector, may be approximated by:

$$N \approx RT = \frac{k_0 A^2 CT}{\mu l^2} (1 - e^{-2\mu d}) \tag{7}$$

As $\varepsilon_d \varepsilon_{dt}$ is assumed to be about the same for both the detectors, the number of counts N_L obtained from the left detector will be approximately equal to N_R, the number of photons registered by the right detector.

A small volume V_t, representing a brain lesion with increased isotope concentration C_t, may be considered to be located within one subregion; (d + x) cm from one of the collimator apertures. The total number of counts from this subregion, registered in the right (R) and left (L) detectors respectively, could approximately be expressed:

$$N_{tR} \approx N_R + \frac{k_0 T(C_t - C) V_t A \exp(-\mu(d+x))}{(l+d+x)^2} \tag{8}$$

$$N_{tL} \approx N_L + \frac{k_0 T(C_t - C) V_t A \exp(-\mu(d-x))}{(l+d-x)^2} \tag{9}$$

assuming that the activity $Q_t = V_t(C_t - C)$ (μCi) of the local accumulation can be approximated by a point-source of the same activity. N_{tR} and N_{tL} may be related to N_R and N_L, the values obtained without this brain lesion phantom V_t in the subregion and increase ratios, I_R and I_L, and a difference ratio D may be defined:

$$I_R = \frac{N_{tR} - N_R}{N_R} \tag{10a}$$

$$I_L = \frac{N_{tL} - N_L}{N_L} \tag{10b}$$

$$D = \frac{N_{tR} - N_{tL}}{(N_R + N_L)/2} \tag{10c}$$

The ratios I_R, I_L and D are to be the evaluated results for each subregion of the standardized pattern of subregions of the skull.

Ideally, if no local increase of isotope in a subregion is present, the ratios I_R, I_L and D obtained from this subregion should be zero. If however a local accumulation of isotope, with the activity Q_t, is located within the subregion, these ratios should deviate from zero. The sensitivity S′ to such a local accumulation of isotope located within a subregion and at a position (d + x) from one of the collimator apertures may be defined as a function of the location x for the parameter D and a parameter defined as the maximum of I_R and I_L; (max (I_R, I_L)):

$$S'_D(x) = \frac{D \cdot 100}{Q_t} = \frac{D \cdot 100}{V_t(C_t - C)} \text{ percental units}/\mu\text{Ci} \tag{11a}$$

$$S'_I(x) = \frac{\max(I_R, I_L) \cdot 100}{Q_t} = \frac{\max(I_R, I_L) \cdot 100}{V_t(C_t - C)} \text{ percental units}/\mu\text{Ci} \tag{11b}$$

if the local increase of activity Q_t is expressed as the product of the volume V_t of the accumulation and the increase of isotope concentration $(C_t - C)$ in that volume.

However, concentration ratios are of greater interest in clinical application of the method than absolute determination of accumulations of isotope.

Therefore, the concentration differences $(C_t - C)$ is related to the homogeneous concentration C in the skull phantom, thus giving:

$$\frac{C_t - C}{C} = \frac{C_t}{C} - 1 = c_0 - 1$$

where c_0 is the concentration ratio between the local accumulation and the surrounding homogeneous concentration.

Corresponding to the definitions of $S'_I(x)$ and $S'_D(x)$, sensitivity functions $S_I(x)$ and $S_D(x)$ can be defined, using the concentration ratio C_0 instead of the concentration difference $(C_t - C)$

$$S_D(x) = \frac{D}{V_t(c_0 - 1)} \cdot 100 \text{ percental units/cm}^3 \text{ of lesion volume} \qquad (12a)$$

$$S_I(x) = \frac{\max(I_R, I_L)}{V_t(c_0 - 1)} \cdot 100 \text{ percental units/cm}^3 \text{ of lesion volume} \qquad (12b)$$

Normally, $N_R \approx N_L$ (LIND et coll.) and hence, by setting $N_R = N_L$ it is obtained from eqs (10 a), (10 b) and (10 c):

$$\max(I_R, I_L) > I_R - I_L = \frac{N_{tR} - N_R}{N_R} - \frac{N_{tL} - N_L}{N_L} = \frac{N_{tR} - N_{tL}}{(N_R + N_L)/2} = D$$

As $\max(I_R, I_L) > D$, it is evident from eqs (12 a) and (12 b) that $S_I(x) > S_D(x)$ which indicates that the parameter D is less sensitive than the maximum of I_R and I_L.

However, for the choice of clinical evaluation parameters, the figure of greatest interest is that for the detectability which is determined by the sensitivity in relation to the normal range (including the measuring error of the parameter). Therefore, as the normal range of a parameter is not known, it is not possible to exclude one parameter as being less valuable than another because of its lower sensitivity (LARSSON & LIND).

As no biologic effects can influence the results in phantom measurements, it is possible to estimate the errors occurring from counting statistics and instability of counting efficiency.

Material and Methods

Detectors and electronic equipment. Two cylindrical NaI detectors (crystal size, diameter 5.1 cm, length 5.1 cm), with one common high-potential supply, were each connected to a pulse-height discriminator and a scaler. The timer was common to both spectrometers, and the measured values were punched on paper tape, thus permitting the use of an automatic measuring routine (Fig. 2).

The collimators. The two detectors were equally collimated. The collimators were made of lead, cast in between an iron tube and a smaller square aluminium profile (wall thickness 2 mm). The collimators were inserted into larger lead shields, facing each other and arranged in such a way that the collimated detectors could be moved along their common axis to obtain a suitable distance between them (Fig. 2). The collimator length was 30 cm and the aperture 3 cm × 3 cm. The two detectors could also be moved congruently in an x − y plane, orthogonal to the central axis, in order to cover the skull in clinical examinations, with a fixed number of subregions.

The isotope. $^{99}Tc^m$ in the form of technetium pertechnetate was used in all determinations and all results were corrected for the activity decay.

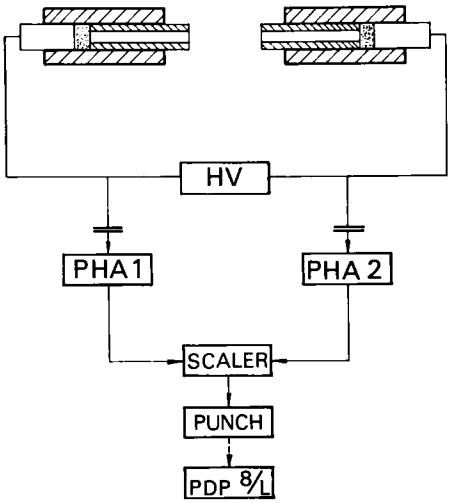

Fig. 2. Block diagram of the detection system. PHA = Pulse height analyser. HV = High voltage supply. PDP 8/L = Computer.

The phantoms. A lucite vessel, length 20 cm, height 20 cm, width 15 cm and wall thickness 0.5 cm, was used as a skull phantom. Three cylindrical lucite containers with thin walls and volumes of 6.3, 21 and 33 cm^3 were used as brain lesion phantoms. The diameter of each cylinder was approximately equal to its length. The lesion phantoms were movable to different distances from the median plane of the skull phantom and always directed with their central axes along the central axis of the collimators (Fig. 1).

Pulse height discrimination. Two different pulse-height discrimination levels are frequently used in photon counting methods. The upper discrimination level can be set high enough to accept most of the pulses from primary photons in the upper part of the total-absorption peak, if the background radiation level is low compared to the level to be measured. The lower discrimination level is commonly set somewhere below or within the lower part of the total-absorption peak, in order to optimize the detectability by compromizing between two counteracting effects; rejection of scattered radiation and good counting statistics. The problem of finding a suitable lower discrimination level for the measuring performance previously described might be important and was therefore analysed.

For this purpose, assume the counting statistics to be the only or the most dominant source of variation, and that a large number of measurements is performed over a subregion of the skull phantom; all of them with measuring times of the same length and with the same homogeneous concentration of isotope in the phantom. From the population of values obtained, $\{N_s\}$, the mean value $\overline{N}_s$, and the standard deviation SD may be calculated. If however, $N_s \geqslant 1$, the measuring values obtained approach a normal distribution with an expected mean, η_s, and standard

deviation, $\sigma_s = V\eta_s$. These values can be expected to be close to the calculated values $\overline{N}_s \approx \eta_s$ and $SD \approx \sigma_s$.

Now, if a small volume of increased isotope content, represented by a lesion or a tumor phantom, is located within the subregion, the corresponding values obtained $\{N_t\}$, may be expected to differ from $\{N_s\}$. The difference z, between a value N_t and the mean value $\overline{N}_s$, can be calculated:

$$z = N_t - \overline{N}_s$$

and for the mean value $\overline{N}_t$ of $\{N_t\}$:

$$\bar{z} = \overline{N}_t - \overline{N}_s;\ \bar{z} > 0 \tag{13}$$

where the value of $\bar{z}$ may be expected to be mostly due to the activity of the lesion phantom.

From the point of view of detectability, the most interesting problem arises when $\bar{z}$ is fairly small compared to N_s, and therefore, for the further calculation, $\bar{z} \ll N_s$ is assumed.

A measure of the probability for a value N_t not to belong to the population $\{N_s\}$, may be expressed by the number of standard deviations, q:

$$N_t - \overline{N}_s = z = qSD$$

For the most probable value $\overline{N}_t$ of $\{N_t\}$; is obtained:

$$\overline{N}_t - \overline{N}_s = \bar{z} = \bar{q}SD \tag{14}$$

which is approximately equal to:

$$\eta_t - \eta_s = \bar{q} \cdot \sigma_s \tag{15}$$

where η_t is the expectation value of the population $\{N_t\}$. The significance of the most probable deviation $\bar{z}$ can hence be expressed in number of standard deviations, $\bar{q}$:

$$\bar{q} = \frac{\eta_t - \eta_s}{\sigma_s} \approx \frac{\bar{z}}{\sigma_s} \tag{16}$$

The size of the ratio $\bar{z}/\sigma_s$ was assumed to be dependent on the energy acceptance interval ΔE. Such a dependence might be of practical importance since the significance of a deviation should be as large as possible in order to increase the possibility of detecting an accumulation of isotope within the subregion.

σ_s could be estimated from the results of a measurement over a subregion of the skull phantom, filled with a homogeneously distributed isotope.

As the difference $\bar{z}$ was assumed to be mostly due to the lesion phantom, the value of $\bar{z}$ could be set approximately equal to $\Delta\overline{N}_t$; the value from a small lesion phantom containing isotope, and located within a subregion of the skull phantom

filled with pure water. This performance simplifies the examination and the following eq. is obtained:

$$\bar{q} = \frac{\bar{z}}{\sigma_s} \approx \frac{\Delta\bar{N}_t}{\sqrt{\bar{N}_s}} \tag{17}$$

Assume $\Delta\bar{N}_{t0}$ and $\bar{N}_{s0}$ to be the results obtained when the pulse-height discriminators are set to correspond to a reference energy acceptance interval ΔE_0. For any other energy acceptance interval ΔE, the corresponding results may be expressed as functions of ΔE:

$$\Delta\bar{N}_t(\Delta E) = \Delta\bar{N}_{t0} \cdot f'_t(\Delta E) \tag{18}$$

and

$$\bar{N}_s(\Delta E) = \bar{N}_{s0} \cdot f'_s(\Delta E) \tag{19}$$

where $\Delta E = E_u - E_l$ is the interval between the upper and the lower discrimination levels. Correspondingly, if E_u is set fixed, $\Delta\bar{N}_t(\Delta E)$ and $\bar{N}_s(\Delta E)$ can be expressed as functions of the lower discrimination level E_l:

$$\Delta\bar{N}_t(E_l) = \Delta\bar{N}_{t0} \cdot f_t(E_l) \tag{20 a}$$

$$\bar{N}_s(E_l) = \bar{N}_{s0} \cdot f_s(E_l) \tag{20 b}$$

From eqs (17), (20 a) and (20 b), $\bar{q}$ as a function of E_l is expressed:

$$\bar{q}(E_l) = \frac{\Delta\bar{N}_t(E_l)}{\sqrt{\bar{N}_s(E_l)}} = \frac{\Delta\bar{N}_{t0}}{\sqrt{\bar{N}_{s0}}} \cdot \frac{f_t(E_l)}{\sqrt{f_s(E_l)}} \tag{21}$$

Spectral distributions of events from the isotope in a subregion of the skull phantom and of events from the isotope in a lesion phantom, located at three different depths within the water-filled skull phantom, were obtained by the use of a multichannel analyzer. From the values thus obtained, the functions $f_t(E_l)$ and $f_s(E_l)$ were calculated giving $\bar{q}$ as a function of E_l with a reference interval $\Delta E_0 = 120$–170 keV (Fig. 3).

The stability of the counting efficiency. The counting efficiency of methods for detection of discrete ionizing particles may be defined as the ratio between the registered number of events/s and the number of ionizing particles/s incident on the detector. The counting efficiency can be expanded into the product of the detector efficiency ε_d and the data transfer efficiency ε_{dt}, the definitions of which have been given by MACINTYRE et coll. (1969) in a report to the International Commission on Radiation and Units (ICRU).

Fluctuations of the counting efficiency, caused for instance by aging or termal effects or by instability of the power supply to the detectors and the discriminators, may be expected to influence on the detectability properties of the method.

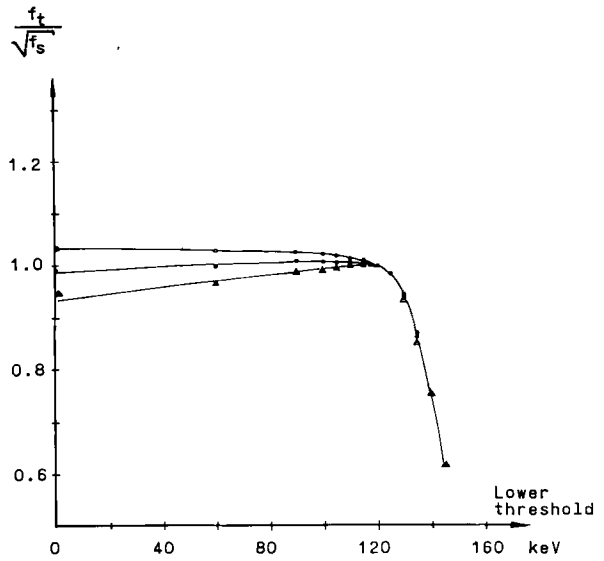

Fig. 3. The function $f_t/\sqrt{f_s}$, obtained from measurements of a tumour phantom located at three different depths plotted against the lower threshold of pulse height discrimination E_1. Estimated errors are approximately within the size of the symbols. ▲ depth = 2.5 cm, ● depth = 6.5 cm, ○ depth = 10.5 cm.

To investigate whether such effects were of importance, the count rate stability was tested on a phantom system. The skull phantom was filled with a homogeneous $^{99}Tc^m$ solution with a concentration corresponding to an initial count rate of about 80 000 cpm from each detector. In view of the results presented in Fig. 3, the pulse height discriminators were set corresponding to the energy interval between 120 and 170 keV. The count rate was registered automatically for one minute periods for several hours. The registrations were also repeated at different times of three days. The values obtained were arranged in 18 groups of 24 consecutive registrations, each group corresponding to a patient examination (LIND et coll.). The mean value and standard deviation of each were calculated and related to the theoretical calculation of the count rate statistics.

Collimation of the detectors. The detector response was determined by measurement of the count rate from a $^{99}Tc^m$ point source at different locations in the water of the skull phantom. The point source was made using an electrolytic procedure in which $^{99}Tc^m$ pertechnetate ions were attached to the tip of a thin needle. The tip was then given a thin cover to prevent loss. The detector response is illustrated by means of isoresponse curves (Fig. 4), which were constructed from the measuring values obtained.

The sensitivity of the method. The sensitivity or the number of units of the evaluation parameter per unit of local accumulations of isotope in a subregion, may be derived experimentally from measurements on a phantom system.

The skull phantom and the three different lesion phantoms were all filled with the

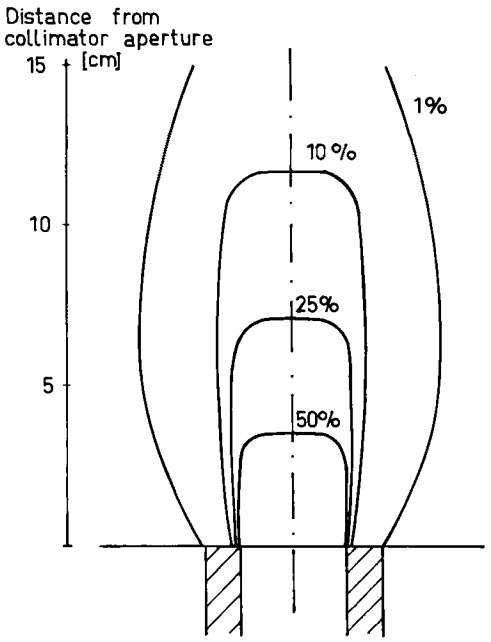

Fig. 4. Isoresponse curves of the detectors.

same solution $^{99}Tc^{m}$ pertechnetate in water. The count rate from one subregion of the skull phantom only was registered by one of the detectors. Thereafter the skull phantom was rinsed, refilled with pure water, and the count rates from the three different lesion phantoms at five different depths within the subregion were measured. From these results I_R, I_L and D were calculated for the concentration ratios $c_0 = 2:1$, 4:1, 8:1 and 12:1, assuming the two detectors to be equal in counting efficiency and assuming I_R, I_L and D to be proportional to $(c_0 - 1)$.

Results obtained with two detectors, corrected for imbalance, were in good accordance with the results obtained with the method described above.

Results

Pulse height discrimination. The upper limit of pulse height discrimination was set to correspond to $E_u = 170$ keV, and the reference energy acceptance interval (ΔE_0) was between 120 and 170 keV. The relation $f_t(E_l)/\sqrt{f_s(E_l)}$ was plotted in Fig. 3 against E_l for three different depths, 2.5, 6.5 and 10.5 cm within the subregion of the skull phantom. This relation, proportional to the significance $\bar{q}$, was almost independent of the location x of the lesion phantom in the subregion (Fig. 3). Moreover the effect of reducing E_l below approximately 120 keV was only small.

Since E_l should be chosen to discriminate against scattered radiation, it should be set as high as possible without decreasing the detectability. Therefore, in view

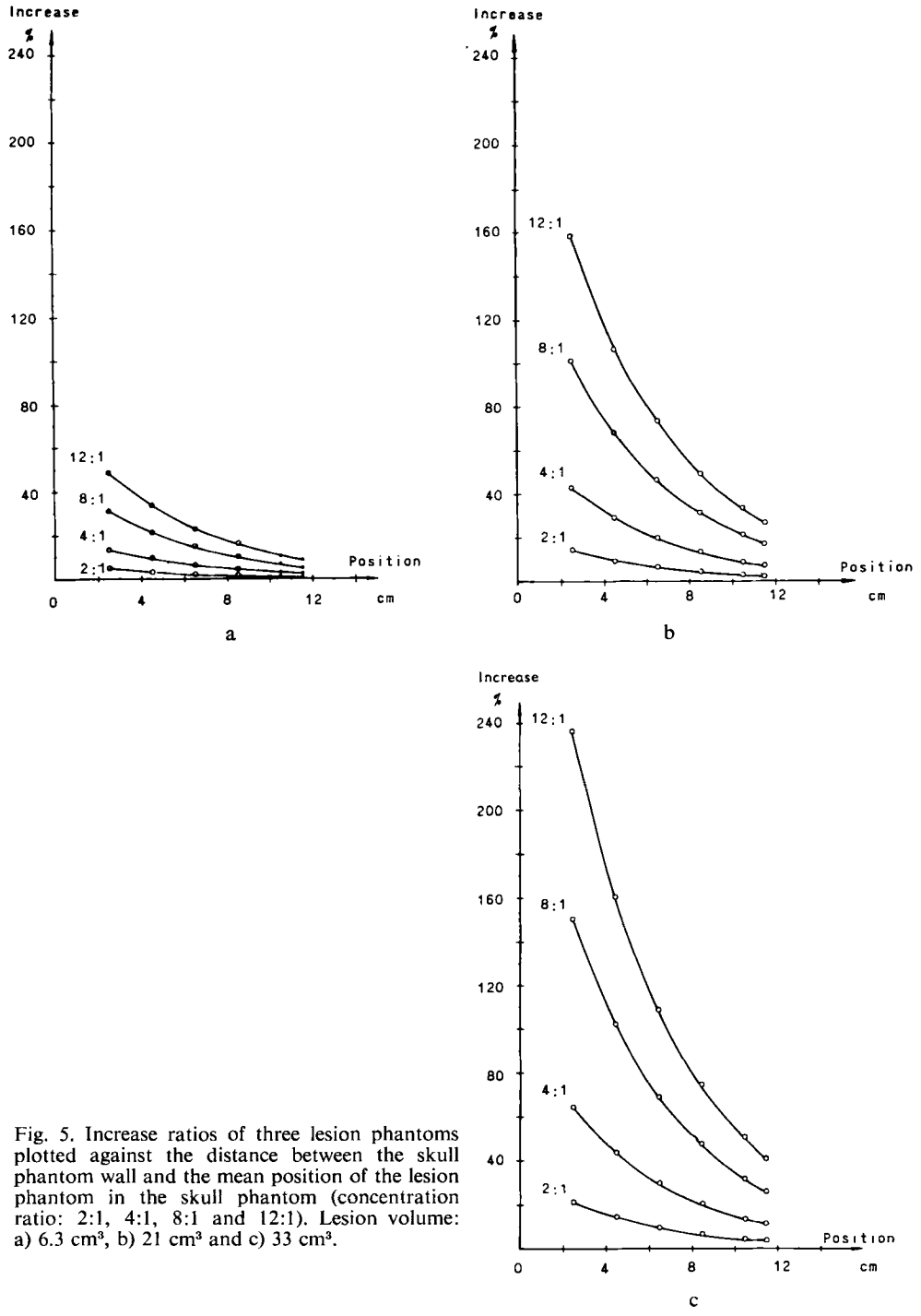

Fig. 5. Increase ratios of three lesion phantoms plotted against the distance between the skull phantom wall and the mean position of the lesion phantom in the skull phantom (concentration ratio: 2:1, 4:1, 8:1 and 12:1). Lesion volume: a) 6.3 cm^3, b) 21 cm^3 and c) 33 cm^3.

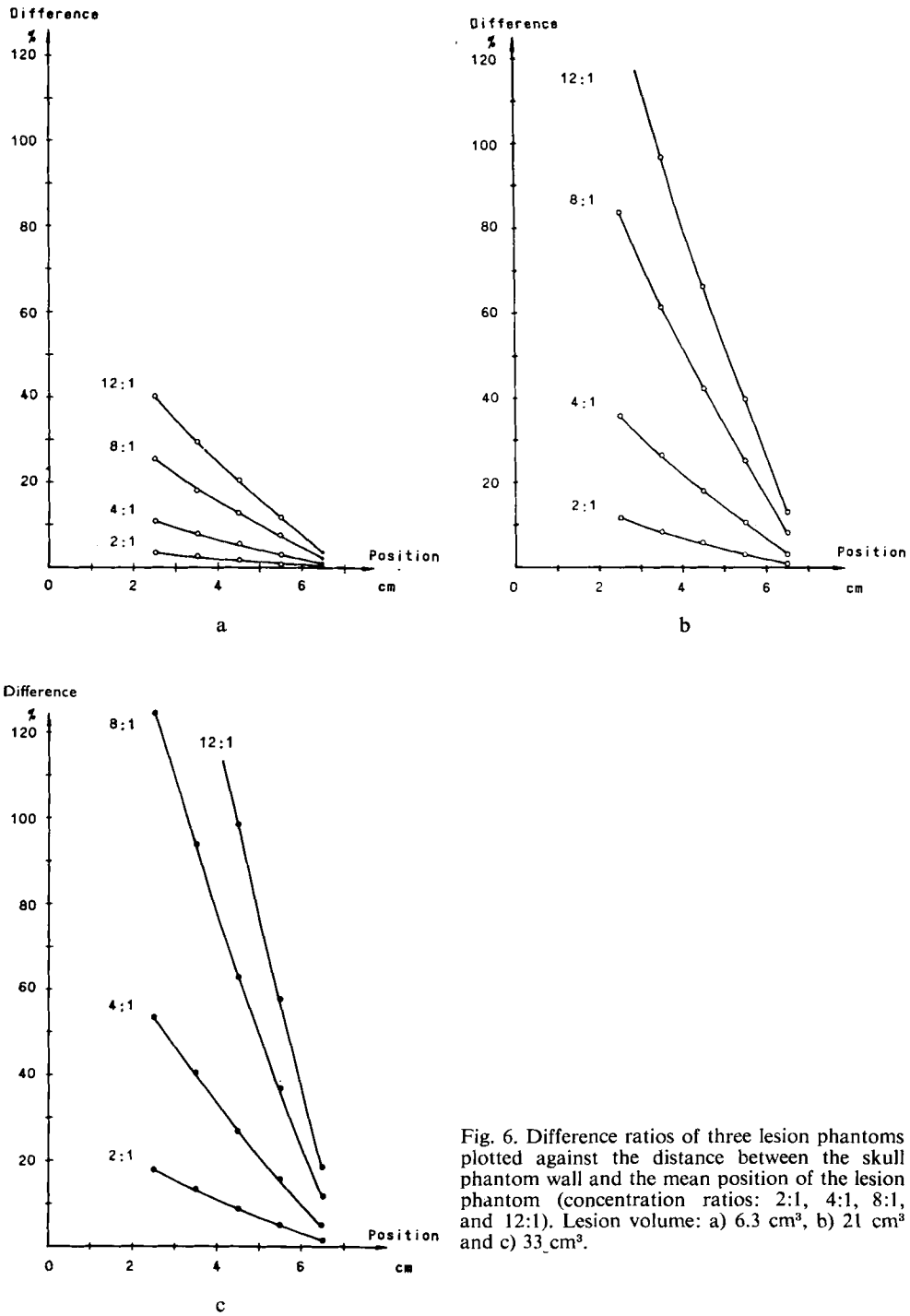

Fig. 6. Difference ratios of three lesion phantoms plotted against the distance between the skull phantom wall and the mean position of the lesion phantom (concentration ratios: 2:1, 4:1, 8:1, and 12:1). Lesion volume: a) 6.3 cm^3, b) 21 cm^3 and c) 33 cm^3.

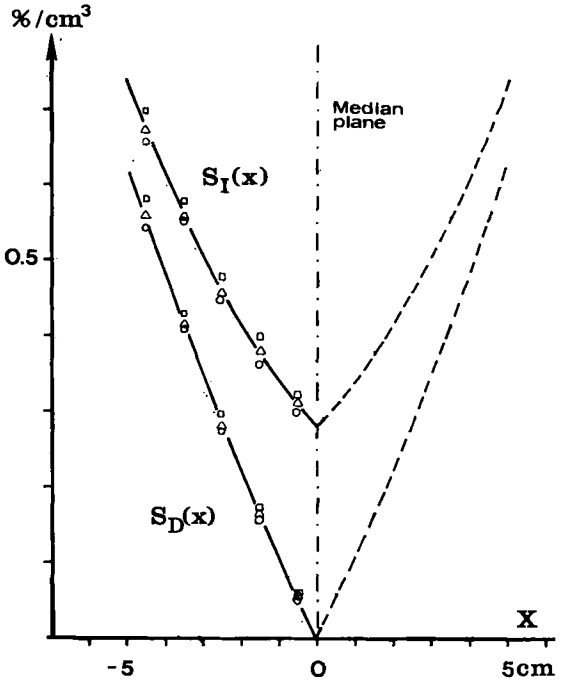

Fig. 7. The sensitivity S_I of the evaluation parameter $\max(I_R, I_L)$ and the sensitivity S_D of the evaluation parameter D as derived from the phantom experiments with different lesion volumes, V_t, are plotted against the location x. □ $V_t = 6.3$ cm³, △ $V_t = 21$ cm³, ○ $V_t = 33$ cm³.

of the results presented in Fig. 3 the lower level was set to correspond to 120 keV. In this interval, 120 to 170 keV, the background radiation was found to be around 100 to 150 cpm.

Stability of counting efficiency. The error due to instability effects in the counting efficiency for the two detectors with pulse height discrimination corresponding to 120 to 170 keV was negligible compared to the error due to counting statistics at the count rate level of interest (10 000 cpm).

The sensitivity of the method. The relation between the ratios I, D and the location of a lesion is plotted in Figs 5 and 6 for the three different volumes V_t. The concentration ratio c_0 has been used as a parameter and as an example a tumour phantom of 21.3 cm³ with a concentration ratio of 4:1 situated 2 cm from the median plane of the skull phantom increased the count rate almost 27 per cent with a difference of about 14 per cent.

The sensitivity as a function of the location x was calculated from the results obtained for the two evaluation parameters $\max(I_R, I_L)$ and D, using eqs (12 a) and (12 b). This is illustrated in Fig. 7 where the values of S_I and S_D are plotted against the location x of the lesion phantoms used. The deviation of these results caused by the

use of different lesion volumes, V_t, may be supposed to be due to a systematic error in positioning of the phantoms and also due to the approximation of assuming the volume source V_t to be equal to a point source.

Discussion

Radiation hazards limit the amount of isotope administered to a single patient or population, and hence the amount of available information is limited because the distribution measurements obtained are disturbed by statistical noise. Most methods for external measurement of the isotope distribution of the skull are based on image interpretation with high geometrical resolution (BURROWS 1972), and the presence and location of a possible lesion is decided subjectively, using an unknown number of parameters. The choice of a small number of standardized subregions decreases the geometrical resolution, but has the advantage of giving a statistical error which may be estimated and a limited number of parameters which may be objectively compared with the corresponding values of a normal group.

The chosen number of such subregions is important for the detectability properties of the method because it decides, within the given examination time, the relation between spatial resolution and precision from counting statistics.

In the method described, the size and number of the subregions was determined by the collimation properties and the standardized movement of the detectors. The area of the collimator apertues, 9 cm^2, was chosen to correspond approximately to the area of increased isotope concentration caused by a small brain lesion (BURROWS). The concentration ratios used were chosen to correspond to concentration ratios in brain tumour cases (SCHWARTZ & TATOR 1972). The length of the collimators was determined by compromising between counting statistical considerations, total investigation time, geometrical resolution at different depths and discrimination against high accumulation of pertechnetate in the skull base.

Energy discrimination. The choice of energy acceptance interval may be made with respect to different parameters, i.e. the discrimination levels can be set to compensate for drift effects in the photomultiplier and pulse electronics (Ross et coll. 1968), or to obtain a high detector figure of merit, Q, in the discrimination between two different count rates (BECK & HARPER 1968). However, highly stabilized solid state devices were used for the pulse electronic equipment and the high voltage supply, and as the instability of the equipment used was shown to be unimportant, there was no need to use the settings of the pulse height discriminators to compensate for drift effects.

To accord with the principles laid down for result evaluation by the method presented, the significance $\bar{q}$ was assumed to be a more suitable parameter than the figure of merit for investigating the effect of energy discrimination.

$\bar{q}$ was shown from eq. (21) to consist of a fixed ratio $\Delta\overline{N}_{t0}/\sqrt{\overline{N}_s}$ and a function

$f_t(E_1)/\sqrt{f_s(E_1)}$ and it was therefore only necessary to examine the latter in order to estimate the influence on $\bar{q}$ of E_1.

The measurements were performed with an idealized measuring situation, in which the lesion phantom was located within the subregion and the skull phantom contained a homogeneous distribution of the isotope. In clinical measurements the results will be affected by regions of accumulated isotope close to the measured subregion. The choice of $E_1 = 120$ keV can nevertheless be expected to be relevant for most of the subregions in a skull examination (Fig. 3).

The results obtained are in agreement with the results obtained by BECK & HARPER, who used the figure of merit concept for an estimation of a suitable setting of the lower discrimination level.

The stability of the counting efficiency. To compensate against aging and long-term drift effects in clinical examinations an accurate check of the counting efficiency of the two detectors is made before each measuring procedure, and from these results, all values are corrected for imbalance. The value for a subregion is also always related to a reference value, obtained from the actual measurement, thus giving a relative distribution pattern independent of long-term effects (LIND et coll.).

The drift effects of interest are thus the short-term effects which could affect the values within the examination time of 30 minutes. However, no such effects of any significance could be observed.

The sensitivity of the method. The sensitivity of measuring devices used in nuclear medicine, is usually defined as the ratio of the number of registered events/s to the total number of γ-photons/s emitted from a point-, line-, plane- or volume-source. As the number of emitted γ-photons is proportional to the activity, the sensitivity will be expressed in absolute values as the count rate per unit of activity.

However, this general definition is not well suited to demonstrate the properties and the differences of the relative evaluation parameters used in the clinical application of the present method. Besides, reports on the accumulation of isotope in brain lesions are concerned with the lesion/brain ratio rather than with any absolute fraction of the total activity administered (BURROWS). The sensitivity was therefore derived from the relation between the relative evaluation parameters I_R, I_L and D, and the relative accumulation of isotope, $V_t(c_0 - 1)$.

The sensitivity could be calculated using eqs (7) to (11 b), but since scattering and penetration effects are not taken into account in these equations the results might not be sufficiently accurate and the relations between I, D and x, V_t and c_0, representing the sensitivity functions $S_I(x)$ and $S_D(x)$, have therefore been determined experimentally (Fig. 7). If a phantom lesion is located between two or four fixed subregions the sensitivity is obviously less than the sensitivity for the same lesion located well within one of the subregions. In patient measurements this effect is counteracted by overlapping subregions.

The sensitivity of an evaluation parameter is possible to estimate from phantom experiments, but a rational choice of evaluation parameters for clinical use must be based on the detectability, which is estimated from the sensitivity of the parameter related to its normal range (LARSSON & LIND, LIND et coll.).

SUMMARY

A symmetry detector method for gammaencephalography is described. It is based on two detectors, collimated by square tubes with large apertures. Increase ratios and side difference ratios should be used as evaluation parameters in clinical applications and are therefore analysed theoretically and experimentally from measurements on a phantom system. The effect of energy discrimination is analyzed and found to be small over a wide range of the lower discrimination level setting. The chosen performance of the measuring routine used made errors caused by the electronic devices negligible in relation to the loss of precision occurring from counting statistics.

ZUSAMMENFASSUNG

Es wird eine Methode mit symmetrischen Detektoren zur Gammaenzephalographie beschrieben. Diese macht von zwei Detektoren Gebrauch, die durch quadratische Kollimatoren mit grossen Apperturen kollimiert sind. Ansteigende Verhältnis-Werte und Seitenunterschied Verhältniss-Werte sollten als Beurteilungsparameter bei der klinischen Anwendung verwendet werden; diese wurden deshalb theoretisch und experimentell von Messungen an einem Phantomsystem analysiert. Der Effekt der Energiediskrimination wurde analysiert; er erwies sich über einen weiten Bereich niedriger Diskrimitator Einstellungsniveaus als gering. Bei der gewählten Durchführung der verwendeten Messroutine können Fehler, hervorgerufen durch die elektronische Anordnungen, im Verhältnis zum Verlust der Genauigkeit, die mit der Messstatistik zusammenhängt, vernachlässigt werden.

RÉSUMÉ

Description d'une méthode à détecteur de symétrie pour la gamma-encéphalographie. Elle est basée sur deux détecteur collimatés par des tubes carrés à larges ouvertures. On devrait utiliser les rapports d'accroissement et les rapports de différence entre les côtés comme paramètres d'évaluation dans les applications cliniques; ces rapports sont donc analysés théoriquement et exérimentalement à partir de mesures faites sur un système de fantôme. Les auteurs examinent l'effet de la discrimination d'énergie et constatent que cet effet est petit sur un grand domaine de réglage du niveau inférieur de discrimination. Le fonctionnement choisi pour les mesures de routine utilisées rend négligeable les erreurs dues aux dispositifs électroniques par rapport à la diminution de la précision provenant des statistiques de comptage.

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