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An Amplifier Providing Fixed Background Elimination and Contrast Enhancement in Scanning Systems

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Methods employed to facilitate visualization of slight differences in density in radioisotope scans fall into two general categories: background attenuation or elimination, and contrast enhancement. Attenuation of extraneous background is routinely accomplished by means of a spectrometer. Indifferent body background can be eliminated completely by use of a ratemeter signal which in the case of photoscanning may establish a threshold below which the photon flux is insufficient to blacken the film, or with a dot scan (1) establish a threshold count-rate below which all counts are rejected. Contrast enhancement may be achieved in a photoscan system by ratemeter control of light intensity in proportion to count-rate. Two reports have appeared describing contrast enhancement in mechanical dot scans. One system depends on a device which controls scan speed in inverse relation to count-rate (2). The other resorts to rescanning the original record with a light-sensitive densitometer which drives a voltage-controlled oscillator (3).

A circuit designed to eliminate background and at the same time enhance contrast by means of an internally generated signal is the subject of the present communication. Rather than utilizing count-rate cut-off, the circuit achieves fixed background elimination at any desired level over the full ratemeter range with the result that the output signal is a function of the net counting rate. It provides contrast enhancement by producing a signal which is generated by a thyratron oscillator at rates in continuously variable proportion to the aforementioned net counting rate.

While the amplifier may be used in both dot and photoscanning systems, it has been tested only with a mechanical system. In either application, as is the case with any recording device which rejects a portion of the gross count, the

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desirability of simultaneously making a companion scan which preserves all raw information is not questioned.

CIRCUIT DESIGN

The amplifier has been designed for use with the ratemeter commonly included in scanner instrumentation. It incorporates a difference amplifier, a grid-controlled thyratron pulse-forming stage, and, for control of the external printing circuit, a silicon controlled rectifier switch (Fig. 1). The potential at plate #2 of the difference amplifier, V-1, is proportional to the difference between the negative signals applied to grids 1 and 2. Negative bias to grid #1 is chosen equivalent to the desired threshold signal level by adjustment of R-1, while that applied to grid #2 is directly proportional to the gross counting rate. Plate #2, reflecting a potential proportional to the net counting rate, is coupled to the grid of the thyratron, V-2, which is normally biased at cut-off for the condition when no net signal appears at the output of V-1. An increase in counting rate above the selected threshold value thus allows the thyratron to oscillate at a rate determined by C-1, R-2 and the thyratron grid potential. The fixed plate resistor is included to prevent the flow of excess plate current when R-2 is decreased. A wide selection of pulse rates is available for a given input when the magnitude of C-1 is variable.

Positive trigger pulses from the cathode of V-2 appear at the input of an unistable switching pair including SCR's 1 and 2 (G.E. Application Notes No. 200.19). SCR-2 is normally conducting but as SCR-1 is triggered into a conducting state, SCR-2 momentarily ceases to conduct as its anode potential is decreased. The magnitude of R-3 determines the length of the conducting cycle of SCR-1, and is selected to provide a sufficiently long "on" phase to activate the

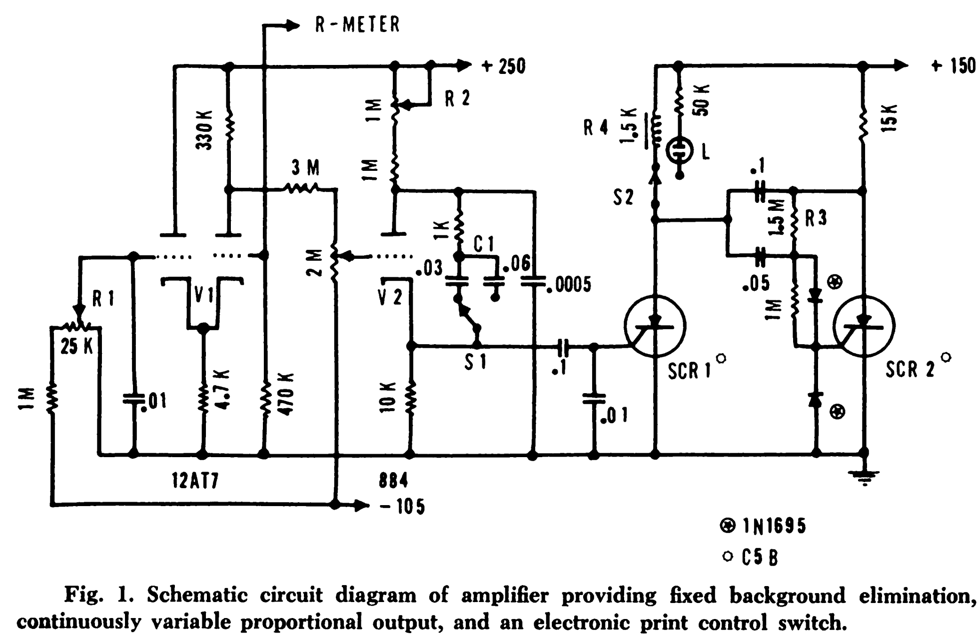

Fig. 1. Schematic circuit diagram of amplifier providing fixed background elimination, continuously variable proportional output, and an electronic print control switch.

printing solenoid, R-4. As the shunting capacitors recharge, SCR-2 is driven back into its stable conducting state while SCR-1 ceases to conduct until the next input pulse is received. A capacitor is included between the input gate and cathode to eliminate the possibility of self triggering introduced by the inductive load of the solenoid. This circuit accomplishes switching without generation of spurious noise commonly generated at conventional relay contacts. The maximum repetition rate is limited to that of the solenoid.

Included in parallel with the solenoid is a neon indicator lamp, L, which may be switched into the circuit in lieu of the printer and used as a visual monitor in preliminary adjustment of the amplifier. Selection of the optimum threshold and printing rate as a function of background and maximum counting rate is simplified through use of this visual indicator.

In a conventional circuit without background elimination or contrast control the relative density of the printed record may be designated as the ratio $\frac{C_I}{C_M}$, where C_I is instantaneously observed as the gross counting rate and C_M is the maximum expected gross counting rate at which, with optimum dot factor attenuation, the printer operates at its maximum practical repetition rate. The dashed curve (A) shown in Figure 2A represents a response curve plotted as a fraction of the maximum. Obviously, introduction of count-rate cut-off would alter only the minimum recorded printing rate in that no response below the selected cut-off level, C_T, would be recorded. As the counting rate exceeded that of C_T, printing rates would be directly proportional to the gross counting rates, *i.e.* they would fall along this response curve. Because the printing rate would be proportional to the total gross counting rate, relative contrast would be unchanged by introduction of the cut-off circuit.

By comparison, the present amplifier achieves contrast enhancement both through background elimination and through the availability of variable printrate factors. It produces an output signal which is a function of the net observed counting rate, $(C_I - C_T)$. The relative density of the printed record is a function of the ratio, $\frac{(C_I - C_T)}{(C_M - C_T)}$, over the limited range C_T through C_M. If maximum recording rates in both systems are identical, much more pronounced contrast can be achieved in the latter, especially over regions where the observed counting rate approaches the value C_T. The advantages of these features are shown in Figure 2. The shaded area (Fig. 2A) represents the region of a typical family of response curves attainable with the amplifier. It is shown based on 0.5 on the upper scale of the abscissa ($\frac{C_T}{C_M} = 0.5$, *i.e.* threshold, or background elimination, is set at 50% of maximum counting rate). The lower scale on the abscissa represents the adjusted, compressed scale for this setting with zero net count now corresponding to 0.5 on the upper scale. The solid curve (B) drawn from the abscissal base to the point of maximum (1,1) within this region denotes an optimum adjustment of printing rate wherein the maximum count-rate is matched by the maximum printing rate. The steeper slope of this curve as compared to that of curve A represents contrast enhancement. The steepest curve forming the left border of the region is an estimated maximum limit of useable contrast enhancement; however, in this instance, the maximum printing rate is attained when the net count-rate is only one-third of the maximum.

The shape of the region occupied by the aforementioned family of response curves remains constant for any value of $\frac{C_T}{C_M}$. Thus, by allowing the point of origin of the depicted region to assume each abscissal value from 0 to 1.0, the total available response of the amplifier would be graphically illustrated. From within a given family the choice of a particular response curve could be made by adjustment of R-2 and C-1. Each curve exhibits a rapidly rising initial phase corresponding to the net input signal range of 0 to approximately -0.2 volts. From -0.2 to -2 volts (full scale ratemeter response) the oscillator output pulse rate is a linear function of $(C_I - C_T)$.

Figure 2B, illustrates how relative contrast enhancement is augmented as the level of background elimination is increased and as the instantaneous net

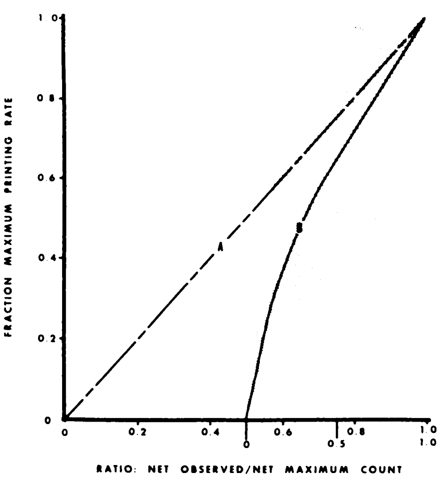

Fig. 2A

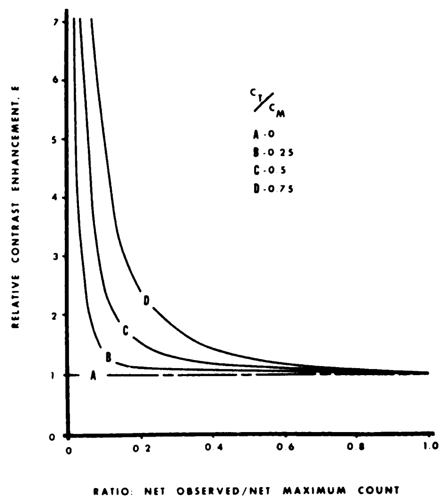

Fig. 2B

Fig. 2. Background elimination and contrast enhancement. A. Examples of range of contrast enhancement (measured as fraction of maximum printing rate) with variation in threshold of elimination. The limits (1,1) represent maximum observed count rate and maximum recordable printing rate. Abscissal scales are $\frac{C_I - C_T}{C_M - C_T}$. (See text for definition of symbols.) The upper scale on the abscissa represents count-rate when no background elimination exists ($C_T = 0$). In the lower scale the zero point has been adjusted for a threshold of elimination, $\frac{C_T}{C_M}$, of 0.5. B. The abscissa is the same as in A. The ordinate represents values of the quantity $\frac{C_I\,(C_M - C_T)}{C_M(C_I - C_T)}$, which is an expression of relative contrast enhancement, E, compared to that of a recording system with no modification save dot factor attenuation. These curves, for several values of $\frac{C_T}{C_M}$, represent relative values of points on response curves of the type shown in A (plotted between any given threshold and the point 1,1) compared to corresponding points on curve (A), for which $C_T = 0$.

counting rate approaches the threshold value. The latter effect may be seen, for example, in curve (C) (elimination below 50% of maximum, $\frac{C_T}{C_M} = 0.5$) where relative enhancement, E, as defined in the legend, is 2.5 when the count-rate is 10% above the threshold point, but only about 1.2 at 50% above the threshold. As the observed net count-rate approaches maximum, the amount of enhancement approaches zero and relative enhancement approaches unity.

In our application two independent amplifier channels and printers are often employed to simultaneously record two scans. Values of $\frac{C_T}{C_M}$ are generally chosen equal to 0.1 and 0.5 respectively. Experience has shown that at the lower value of $\frac{C_T}{C_M}$, the resulting scan closely approximates the actual projected organ size. At the higher value, contrast is greatly increased to more positively delineate intra-organ defects. As an aid to comparison of the two scans, the maximum printing rates are adjusted to be identical in each channel.

An alternate dual recording mode, examples of which are included below, is that in which one amplifier-printer channel is paired with a second printer which records scaler output attenuated only by a conventional dot factor.

EXAMPLES OF PERFORMANCE

Typical contrast enhancement is demonstrated in the scans of Figure 3. They were recorded simultaneously with a probe containing a three-inch diameter, one-inch thick crystal with a 19-hole focusing collimator. The acceptance window of a pulse height analyzer was set at 50 keV. The phantom consisted of 80 μC of ^{131}I in water solution contained in a bath slanted in such a way that the

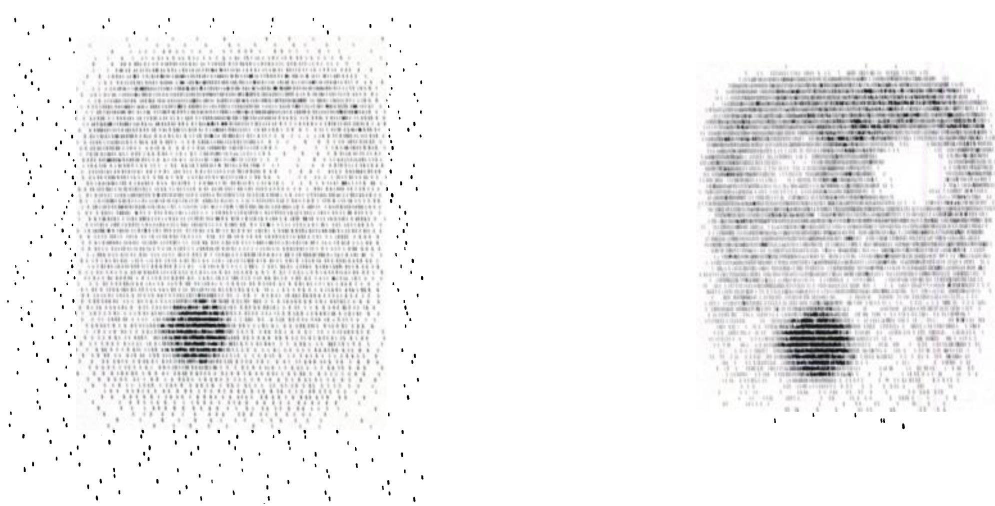

Fig. 3. Scans of phantom source containing iodine-131. Scan speed eight-inches per minute. Scan A *Left:* was recorded with a conventional scaled output, dot factor 1:16. Scan B *Right:* with the amplifier described (see text). C_M = 4500 c/m; C_T = 1000 c/m; time constant = 0.5 seconds.

depth of solution varied from 1 to 3 cm, and in which three 50 ml beakers were placed. The first, representing a "hot" source, contained 10 μC of ^{131}I in water solution; the second, the "cold" source, water alone, 3 cm in depth; and the third, an area of intermediate activity, water alone, 1½ cm in depth, immersed in the bath solution to the 1½ cm depth. The pattern of Figure 3A was produced by the conventional scaling circuit without background elimination, dot factor 1:16. The pattern of Figure 3B, with the circuit described herein, was adjusted to eliminate room background and to produce a maximum printing rate approximating that recorded over the "hot" region of Figure 3A.

Figure 4 is a routine cardiac scan with a conventional scaling circuit, A, and with the new circuit, B. In the latter instance contrast enhancement and background elimination are seen to be accomplished without narrowing the dimensions of the organ.

DISCUSSION

Because of the relative paucity of available photons in radioisotope scanning as compared to x-ray photography, a rendition of a wide scale of shades of gray is not possible. The eye simply must discern whether intensity differences are systematic or random, *i.e.* whether given areas are in fact relatively "hot" or "cold". Background elimination and contrast enhancement both assist the eye in detecting such areas. In photoscanning such modifications ordinarily are achieved by means of ratemeter circuits which control threshold point and light intensity as a function of count-rate. Multicolor rendition, which in a sense is comparable to contrast enhancement, may also be mediated by a ratemeter circuit. It is apparent that the device herein described is comparable to other methods of contrast control in that it depends on a ratemeter, which instead of controlling light intensity or color, controls printing rate directly through its biasing of a

Fig. 4. Cardiac scans (^{131}I Human Serum Albumin). Scan speed 14-inches per minute. Scan A *Left:* was recorded with a conventional scaled output, dot factor 1:16; B *Right:* with the amplifier described. $C_M = 3400$ c/m; $C_T = 1200$ c/m; time constant 0.5 seconds.

grid-controlled thyratron. The rate of oscillation of the thyratron circuit (*i.e.* printing rate) at a given grid potential (*i.e.* net counting rate) is continuously variable over wide limits. Background elimination is achieved by subtracting a chosen threshold-equivalent signal from the ratemeter output.

Because a ratemeter is a conventional component of systems which permit contrast enhancement and background elimination, peripheral scalloping may be annoying unless the time constant of the ratemeter is kept relatively short (approximately 0.1 seconds or less) when the rate of detector traverse is rapid in relation to concentration of activity in the organ. By minimizing scalloping by using a short time constant, the ratemeter fluctuates rapidly and widely. This, however, does not distort the scan more seriously with the present mechanical system than in photorecording or in a multicolor system. A brief high excursion of the ratemeter in the mechanical system triggers a short burst of dots, whereas in photorecording it produces an exceptionally black spot on the film. In multicolor recording the consequence is an impure mixture of colors. In all cases the distortion consists of increased mottling. When isotope concentration in an organ is high, and rate of detector traverse is relatively slow in relation to that concentration, a longer time constant eliminates splotchiness in the record without producing scalloped edges.

It should be noted that in photoscanning, contrast cannot be built up beyond a certain limit because the saturation point of the film restricts the range of linear response of density to increasing light intensity. With the present system, for practical purposes, no upper limit of rate of pulse generation exists. If a photo system were employed which recorded a dispersed pattern of multiple small, totally black dots the degree of blackening to the eye would depend on the number of dots per unit area rather than the opacity of individual dots. The degree of apparent blackening, then, would be linear over broad limits.

SUMMARY

The design is presented of an amplifier capable of producing radioisotope scans of increased contrast through use of fixed background elimination and an internally generated continuously variable read out signal which is a function of the net observed counting rate. Scans are shown for comparison of the response of the amplifier with that of a conventional system employing dot factor attenuation.

REFERENCES

1. MacIntyre, W. J., and Houser, T. S.: A method for the visualization of the configuration and structure of the liver. Part B. A counting rate cut-off circuit for increased contrast in automatic scanning. *Am. J. Roentgenol., Radium Ther. and Nuclear Med.* **47**:471, March, 1957.

2. Curtis, L., and Cassen, B.: Speeding up and improving contrast of thyroid scintigrams. *Nucleonics*. **10**:58, Sept. 1952.

3. Harris, C. C., Bell, P. R., Francis, J. E., Jr., Jordan, J. C., and Satterfield, M. M.: Data recording for radioisotope scanning *IN* Kniseley, R. M., Andrews, G. A., and Harris, C. C. (eds): Progress in Medical Radioisotope Scanning, Chpt. 3, pg. 81, proceedings of a symposium at the Oak Ridge Institute of Nuclear Studies, Oct. 22-26, 1962; USAEC, T.I.D. 7673.