AN IMPROVED SCINTILLATION DOSIMETER

by

E. W. Mazerall

A thesis presented to the Faculty of Graduate Studies and Research, University of Manitoba, in partial fulfilment of the requirements for the degree of Master of Science.

Department of Physics

University of Manitoba

Winnipeg

April, 1963

ABSTRACT

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Prior to 1950, the determination of radiation dosage from radium insertions within body cavities had been made from theoretical considerations, using isodose curves with the assistance of condenser ionization chambers. Ionization chamber probes of small dimensions were made possible when subminiature electrometer tubes became available.

The development of the scintillation counter suggested the use of this technique for dosage measurement. In 1954, the writer was asked to undertake the development of a probe type scintillation dosimeter, following the successful design of such an instrument by Belcher in England the previous year.

This paper describes the mechanical and electronic details of an improved probe type scintillation dosimeter developed at the Manitoba Cancer Foundation in 1954 - 1955, and which has been used in its two clinics in regular daily service since that time.

The theory of the production and measurement of therapeutic radiation is discussed briefly. A short clinical evaluation is given.

ACKNOWLEDGEMENTS

The writer is pleased to acknowledge his indebtedness to Dr. J.M. Gillies of the Manitoba Cancer Foundation for the radiographs showing a typical radium insertion; to Dr. J.E. Bennett of the Manitoba Cancer Foundation and Dr. W.J. Friesen for permission to photograph the instrument in use; and to Dr. Jean MacFarlane for her assistance and encouragement.

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INTRODUCTION

Within the past two and a half decades clinical radiologists have become increasingly concerned with the accurate measurement of radiation dosages. Standardized units and conditions of measurement have been set up, so that, while the assessment of the patient's condition and the prescription of the radiation dosage is still largely an art, and undoubtedly will remain so for some time, the carrying out of the prescription is rapidly reaching the status of an exact science. It is not surprising then, particularly in the larger treatment centres, to find staff physicists working in close cooperation with the radiotherapists.

With this increased demand for more accurate measurement of dosage, there has been a concomitant demand, not only for more accurate equipment for this purpose, but for a continued refinement of this equipment, leading to greater rapidity and simplicity of measurement. The advent of nuclear energy has supplied a great impetus to the development of equipment and techniques for the precise measurement of radiation, and these developments have not been slow in finding their way into the field of therapeutic radiology.

Instruments for the measurement of teletherapy dosage have been in use for some years, but prior to 1950, the determination of radium dosage from insertions within body cavities had been made from theoretical considerations, using isodose curves with the assistance of condenser ionization chambers (1). Attempts had been made to meas-

ure ionization currents from small chambers inserted into the body, using the electrometer techniques of the time (2,3), but accurate measurements were to wait on the development of sub-miniature electrometer tubes which could be incorporated into the probe itself.

Such a device was described in 1950 by Turner and Newberry (4). In 1954, Fedoruk, Johns and Watson (5) described an improved ionization dosimeter using the electrometer circuit due to Moody (6).

The development of the scintillation counter suggested the use of this technique for dosage measurement, and in 1953, Belcher (7) described a probe type scintillation dosimeter. While this was the first such instrument to be used for body-cavity dosimetry, probe type instruments using crystal phosphors of small dimensions has been used by Ittner and Ter-Pogossian (8,9) for the determination of the energy dependence of crystal media, and by Cole, Moore and Shalek for the production of isodose curves of radium sources (10).

In 1954 the writer was asked to build a dosimeter similar to that described by Belcher. The first such instrument followed Belcher's fairly closely in design in order to get it into operation as soon as possible. The circuit differed from Belcher's in that the amplifier was simplified by the use of a cross-coupled pair described by Van Scoyoc and Warnke (11). Zero drift due to heater voltage variation was minimized by the use of a voltage regulating filament transformer.

The probe unit was essentially the same as Belcher's, and suffered from the same fault: it did not have provision for cutting off the light from the phosphor in order to back off the signal caused by the irradiation of the photomultiplier itself. As soon as this

dosimeter was completed, work was begun to develop the model which is the basis of this thesis.

The considerations to which particular attention was given in this development were as follows:
In order to reduce weight to a minimum, the high voltage transformer and shunt regulator were to be replaced by a pulse type supply with corona regulator; the voltage regulating filament transformer to be replaced by a compensating circuit; and the overall power requirements to be reduced by use of a very simple but stable circuit. The instrument as a whole was to be simplified both as to layout and design to promote ease of servicing and duplication; and a shutter was to be built into the probe unit to overcome the fault mentioned above.

The completed instrument was satisfactory in all these respects, and as soon as it was placed in service, the earlier provisional model was rebuilt to the new design.

NATURE AND PRODUCTION OF THERAPEUTIC RADIATION

Radiotherapy is concerned for the most part with electromagnetic radiation in the wavelength range from 0.2 to 0.01 A. Depending on its source, such radiation is known as X- or gamma radiation.

Stated in the simplest possible terms, both gamma and X-radiation are atomic phenomena, the former a result of nuclear reactions, or interactions between nuclear particles, the latter a result of interactions within the extranuclear electron cloud.

The nuclei of all elements of atomic weight greater than lead are inherently unstable, and disintegrate spontaneously with the emission of an alpha particle (helium nucleus), or a beta particle (electron) or a gamma ray, or a combination of any of them. It is considered that the heavy elements, uranium, actino-uranium, and thorium are the progenitors of a series of unstable elements giving rise ultimately to stable isotopes of lead.

In addition to the series of radioactive heavy elements, there exist in nature isotopes of many of the lighter elements which are unstable, and hence radioactive. For the most part these isotopes are so rare as to be negligible, and it was not until the artificial induction of radioactivity in the atomic pile that quantities of these lighter isotopes became commercially available.

While all three types of radioactive emission: alpha, beta, and gamma are utilized therapeutically, we shall confine ourselves, for the purposes of this paper, to gamma and X-radiation.

The fundamental unit of radiation energy is the erg, but the practical unit in general use is the electron volt (ev), or its multiples by 1000, the Kev, Mev, and lately, the Bev.

$$1 \text{ electron volt} = 1.60 \times 10^{-12} \text{ ergs} \qquad (1$$

and is the amount of energy gained when an electron falls through a potential difference of 1 v.

From Planck's equation

$$E = h\nu$$

where E is in ergs, h is Planck's constant, 6.61×10^{-27} erg-seconds, and ν is the frequency in cycles per second, we get by simple manipulation

$$E = \frac{12,400}{\lambda} \qquad (2$$

where E is in electron volts and λ is in angstroms.

X-radiation is characterized by the fact that it consists of line spectra superimposed on a diffuse background of "white" radiation, whereas gamma radiation is usually confined to one or more discrete wavelengths.

The spectral lines of X-radiation are analogous to the lines of optical spectra in that the radiation corresponding to a particular line is produced when an electron fills a void in one of the electron shells. The shells involved in X-radiation, however, are the innermost, or K,L, and M shells, whose binding energies are very high. For tungsten, the binding energies of the K, L, and M shells are 70, 11, and 2.5 Kev, respectively.

The production of white, or continuous radiation does not

receive a very satisfactory treatment by most writers; it is explained by some simply as the result of deceleration of high energy electrons by "collisions" with nuclei. Compton and Allison (12) state that the continuous spectrum cannot be accounted for adequately by classical electron theory and electrodynamics. They discuss various quantum approaches: that of Kramers and Wentzel using the correspondence principle, and that of Sommerfeld using de Broglie wave mechanics.

The minimum wavelength produced, given by equation (2), depends only on the peak accelerating potential, and would occur only when the electron lost all of its energy in one step. The maximum intensity occurs at a wavelength about 1.5 time the minimum wavelength.

INTERACTION OF ELECTROMAGNETIC RADIATION WITH MATTER

Electromagnetic radiation may react with matter in several ways: it may be deflected or scattered by the electrons of an atom; it may give up part or all of its energy to an electron; or it may react with the Coulomb field of the nucleus to produce an electron-positron pair.

Reaction between electromagnetic radiation and the nucleus does not occur at energies with which we are concerned here, except for the (γ, n) reaction with Be^9 (1.63 Mev), and H^2 (2.23 Mev), and these need be mentioned only in passing (13).

Absorption

Consider a collimated source of radiation, S, an absorber of thickness x, and a collimated detector, D. The intensity of radiation measured by D is

$$E = E_o e^{-\mu x} \qquad (4$$

whence the energy absorbed is

$$E_a = E_o (1 - e^{-\mu x}) \qquad (5$$

μ is called the linear absorption coefficient, and has the dimensions cm^{-1}.

It is evident that our detector will give us a measure of all the radiation that has been removed from the beam, whether this has been truly absorbed, or simply deflected out of its range.

It is sometimes convenient to use other absorption coefficients, related to the linear coefficient as follows:

Mass ..... μ/ρ cm^2 gm^{-1}

Atomic ..... $\mu/\rho \cdot A/N$ cm^2 $atom^{-1}$

Electronic ..... $\mu/\rho \cdot A/N \cdot 1/Z$ cm^2 $electron^{-1}$

where ρ is the density, Z the atomic number, A the atomic weight, and N, Avogadro's number (6.02×10^{23}).

The electronic coefficient is sometimes written ${}_e\mu$.

It will be noted that these coefficients have the dimensions of an area, and for this reason are called cross sections.

Within the energy range with which we are concerned, four processes of interaction between radiation and matter are recognized: (a) Classical, or Rayleigh, scattering, (b) Photoelectric effect, (c) Compton process, and (d) Pair production. These are listed in order of increasing energy of the radiation, although classical scattering can be shown to be a special case of the Compton process. The absorption coefficient, μ, may be separated into component coefficients representing the above processes:

$$\mu = \tau + \sigma + \kappa \qquad (6$$

where τ is the coefficient for the photoelectric effect, σ the Compton process, and κ, pair production.

The corresponding mass coefficients are

and the electronic coefficients are ${}_e\tau$, ${}_e\sigma$, and ${}_e\kappa$.

(a) Classical or Rayleigh Scattering. When the energy of the incident radiation is less than the binding energy of the atomic electrons, no absorption occurs, but the energy is re-radiated with a definite phase

relationship and angle to the incident radiation. This phenomenon has been called unmodified or coherent scattering, since the scattering action of different atomic electrons combines coherently. The amount of radiation scattered per electron is given by J.J. Thompson (14).

$$I = I_o / R^2 \cdot r_o^2 /2 \cdot (1+ \cos^2 \phi) \qquad (7$$

where I is the intensity scattered per free electron at an angle ϕ from the direction of the incident beam; R is the distance from the scattering electron to the point of observation of I; and r_o is the classical electron radius $e_o^4 / m_o^2 c^4$.

The total scattering in all directions given by the above formula leads to a scattering coefficient per electron

$$_e\sigma_s = 8/3 \cdot \pi r_o^2 = 6.65 \times 10^{-25} \text{ cm}^2 \qquad (8$$

i.e., the scattering cross section of a free electron is a universal constant, independent of the energy of the incident radiation.

(b) Photoelectric Effect. When the energy of the incident photon is greater than the binding energy of the electron, the latter may be completely ejected from the atom.

The kinetic energy of the electron is given by

$$E_{kin} = h\nu - E_B \qquad (9$$

where $h\nu$ is the photon energy and E_B is the binding energy of the electron. For low energy photons, the interaction takes place with the outer electrons; the inner, more tightly bound electrons becoming involved as the energy increases, until $h\nu$ reaches the energy level of the K shell, when all the electrons can take part in the reaction. The probability that the electrons of any shell shall be ef-

fected is greatest when the photon energy just exceeds the binding energy of that shell. Photoelectric absorption increases with atomic number, and decreases as the energy of the incident radiation increases. For photon energies less than 100 kev, the photoelectric absorption coefficient is found experimentally to be (13)

$$t = \text{const} \cdot Z^4 E^{-3} \qquad (10$$

At higher energies the Z exponent increases to 5, while the E exponent decreases, reaching -1 for energies very much greater than $m_o c^2$.

When an atom has been ionized by a high energy photon, it regains its ground state with the emission of fluorescent radiation. If the ejected electron were from the K shell of a material such as lead, the fluorescent radiation could have a maximum energy of 88 kev, which is sufficiently high to escape from the material. We should, under these circumstances, distinguish between total absorption, and true absorption. However, for less dense material, such as tissue, with which we are mainly concerned here, the maximum binding energy is about 500 ev, and the fluorescent radiation of this energy is readily absorbed within a very short distance, usually within the cell, one of whose atoms gave rise to the radiation.

(c) Compton Scattering(15, 16). In the case of the collision of a photon with a bound electron, where hν is only slightly greater than E_B, as in the photoelectric process, energy and momentum are conserved with the aid of the recoiling nucleus. When the collision occurs with a free electron (and a bound electron may be considered free when $h\nu \gg E_B$) a scattering process occurs in which energy and momen-

tum are conserved between the incident photon, the recoil electron, and the scattered photon.

$$h\nu' = \frac{h\nu}{1 + \frac{h\nu}{m_o c^2}(1 - \cos\phi)} \qquad (11$$

where $m_o c^2$ is the rest energy of the electron, 511 kev, and ϕ, the angle of scattering of the photon.

This equation may be written in terms of wavelength:

$$\Delta\lambda = \lambda' - \lambda = \frac{h}{m_o c}(1 - \cos\phi) \qquad (12$$

where $\frac{h}{m_o c} = 2.43 \times 10^{-10}$ cm is called the Compton wavelength of the electron.

The energy of the recoil electron is given by

$$E_{kin} = \frac{h\nu}{1 + \frac{m_o c^2}{h\nu(1 - \cos\phi)}} \qquad (13$$

E is a maximum when $\phi = 180^o$

$$E_{max} = \frac{h\nu}{1 + \frac{m_o c^2}{2h\nu}} \qquad (14$$

The total Compton absorption coefficient is made up of two parts, the scatter coefficient, and the true coefficient, i.e., the part representing the energy given to the recoil electron.

$$_e\sigma_t = {}_e\sigma_a + {}_e\sigma_s \qquad (15$$

These have been calculated from quantum mechanical considerations by Klein and Nishina (15). It can be shown that at low energies

$$_e\sigma_t = {}_e\sigma_s = 6.65 \times 10^{-25} \qquad (16$$

the classical value derived by Thomson.

$_e\sigma_t$ falls from the classical value as the energy is increased, reaching a value of about 2×10^{-25} cm^2 at 1 mev. $_e\sigma_s$ falls below $_e\sigma_t$ at the higher energies. $_e\sigma_a$ rises from a very low value to a maximum at about 0.5 mev of 1×10^{-25} cm^2, and then falls off.

(d) Pair Production. The rest energy of the electron is $m_o c^2 = 0.511$ mev. When photon energies exceed twice this value, the photon may react with the Coulomb field of the nucleus to produce an electron-positron pair. Any excess energy appears as kinetic energy of the positron and electron. The kinetic energy will not necessarily be shared equally between the two because of the indeterminate momentum acquired by the nucleus in the process.

For a positron nearly at rest, the possibility of collision with an electron, resulting in the annihilation of the two is high. In this case, two quanta, of 0.511 mev are produced, conservation of momentum requiring that they be emitted in opposite directions.

MEASUREMENT OF RADIATION

The necessity for devising some means for the measurement of radiation became evident almost as soon as X-rays came into medical use, which was very soon after their discovery. The earliest methods made use of fluorescence and the darkening of photographic film, although attempts were made to use chemical means, as well as to measure the heat produced in a metallic absorber. All of these methods are in use today, and the first method will be dealt with in some detail later. In the early state of the art, however, they proved unreliable and were largely superceded by ionization techniques. As early as 1896, J.J. Thomson had shown that X-rays could discharge electrified bodies. He explained this on the basis of ionization of the surrounding air, and suggested that it might serve as a means of measurement of the intensity of X-rays.

While some method of measuring total energy flux might at first glance seem the more direct attack, it is evident that only that part of the flux which is actually absorbed is biologically effective. For this reason, a measuring device which has the same absorption characteristic as tissue, in the range of energies under consideration, will give an accurate means of determining the intensity of absorbed radiation. The time integral of the absorbed energy is called the "dose rate".

The effective atomic number of air is very nearly the same as that for tissue and water, and hence the ionization of air may be utilized to give a measure of the dose rate in tissue over a consid-

erable range of radiation energies. This will be true only for electromagnetic radiation however. For neutron and high energy proton radiation, the material of the measuring instrument must approximate that of tissue more closely, especially with regard to its hydrogen content.

The Roentgen. The international unit of X-ray dose was first defined at the Stockholm Congress of Radiology of 1928 thus:

"The roentgen is the quantity of X-radiation which, when the secondary electrons are fully utilised and the wall effect of the chamber is avoided, produces in 1 cm^3 of atmospheric air at 0^o C and 76 cm of mercury pressure such a degree of conductivity that 1 esu of charge is measured at saturation current".

This definition was modified at the Chicago Congress of Radiology (1937) to read:

"The roentgen shall be the quantity of x- or γ-radiation such that the associated corpuscular emission per 0.001293 gm of air produces, in air, ions carrying 1 esu of quantity of electricity of either sign".

The number of ion pairs produced in air by the absorption of one roentgen is $1/3$, where e is the electronic charge in esu. If W is the energy lost per ion pair, the absorbed energy per roentgen is W/e. W has been determined experimentally as 32.5 ev, or 32.5e/300 ergs. Hence the energy absorbed per gm of air per roentgen is

$$\frac{32.5}{300 \times 0.001293} = 83.8 \text{ ergs} \qquad (17$$

This quantity has been given the name "gram-roentgen".

The Ionization Chamber (17). The electroscope was the first device used to measure radiation by the ionization produced, and in the quartz fibre form is in use today. The most generally used apparatus to measure x-ray dosage in terms of the roentgen as defined, is some form of the parallel plate ionization chamber.

The construction of the standard parallel plate ionization chamber is shown diagrammatically in Fig. 1. T represents the target of an x-ray tube; D, a lead collimating diaphragm with aperture of area A; F the high voltage electrode; E, the collecting electrode, surrounded by the guard electrode, G. The potential, E, must be sufficiently high to prevent recombination of the ions. The separation, d, is great enough to permit all secondary electrons to complete their paths in air. Ions leaving the collecting volume, such as b, are replaced by other ions, a, produced outside. Under conditions of equilibrium, the total number of ions collected will be equal to the total number produced in the sensitive volume (the cross-hatched area in the diagram). Because of the inverse square-law effect, the volume of length L at P, at distance l_2, is equivalent to the volume of length L, and cross section A at the point P_1 at the distance l_1.

If the voltage change on the collecting electrode is V, and the capacity of the system C, the dose at P_1 is

$$\frac{CV}{300} \times \frac{l}{AL} \times \left(\frac{760}{p} \times \frac{273 + t}{273} \right) \text{ roentgens} \qquad (10$$

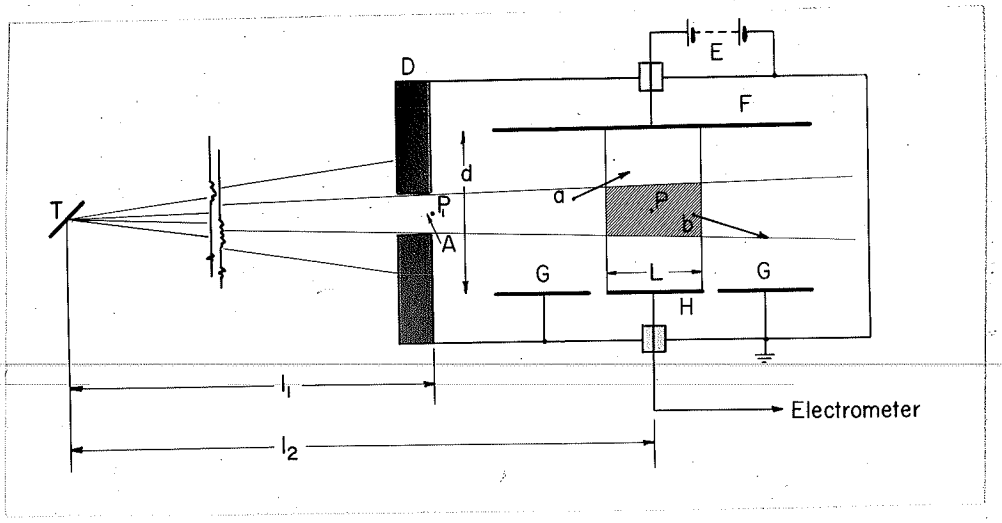


Fig. 1. Parallel plate ionization chamber in schematic form.

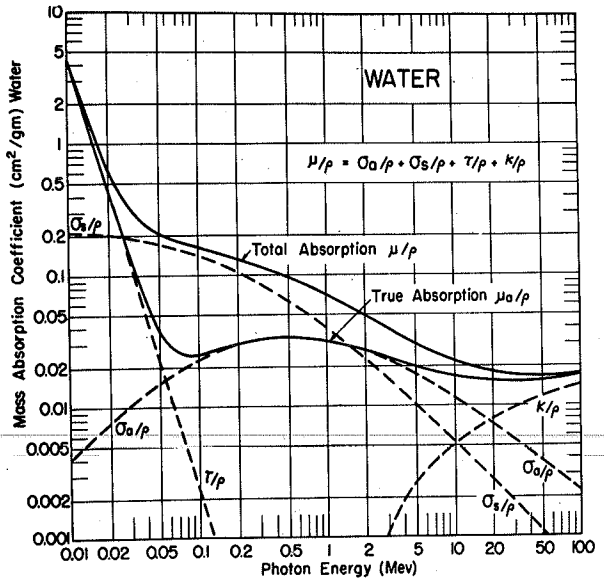


Fig. 2. Mass absorption coefficients for water as a function of photon energy (20).

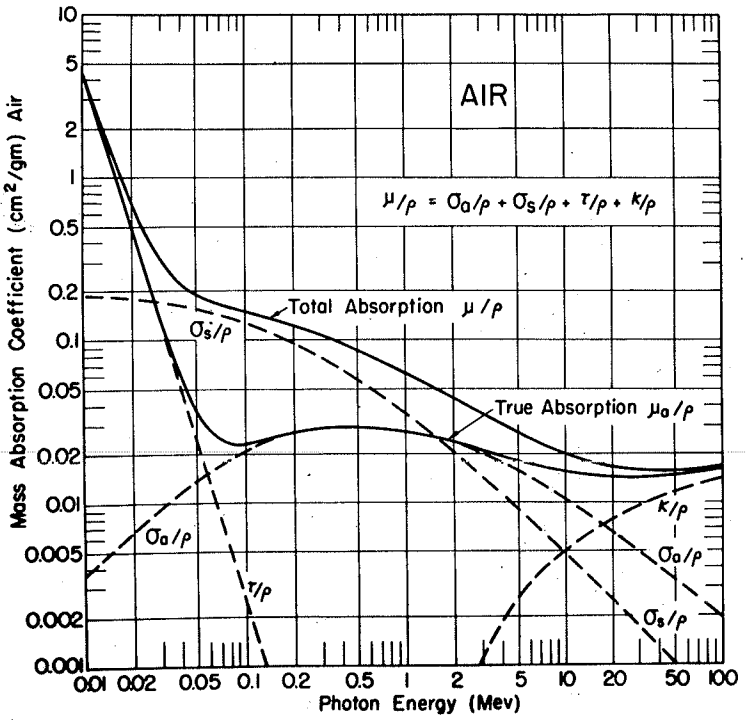


Fig. 3. Mass absorption coefficients for air as a function of photon energy (23).

which includes corrections for air pressure and temperature.

The standard air chamber has air walls, and is used in air. Hence, it is in complete equilibrium. Quite different conditions hold in the case of practical small chambers for the measurement of dose rate in water or tissue. If the dimensions are small, say 1cc or less, and if the wall material is approximately the same composition as air, and of sufficient thickness that secondary electrons produced in the external medium are completely stopped in the wall material, the chamber may be calibrated against a standard chamber and used for the measurement of dose rate. In practice such a chamber is usually constructed of bakelite, or some equivalent material, made conductive by admixture with carbon, or coated on the inside with carbon. The collecting electrode is usually made of thin walled aluminum tubing, or sometimes simply of aluminum wire.

The theory of the small chamber was first treated by Bragg (1912) (18), and later by Gray (1936) (19). It is not necessary for our purposes to consider it at any further length here.

The Scintillation Counter

It had been known for many years that certain materials exhibited the phenomenon of luminescence when exposed to ultra violet and shorter electromagnetic radiation, or to a beam of high energy particles such as electrons, or alpha particles. Many early experimenters, including Rutherford, counted atomic particles by means of a fluorescent screen, viewed under a microscope. The tedium of such a procedure, as well as the impossibility of counting particles when the rate increased

beyond a few per second caused the technique to fall into disuse. With the invention of the multiplier photo-tube, a tool was made available which not only possessed the requisite sensitivity, but was capable of resolving light pulses which occurred at time intervals of less than a microsecond. The first scintillation counter utilizing a scintillating crystal and a photomultiplier was introduced by Curran and Baker in 1944, followed, independently, in 1948 by Kallman. Since that time great advances have been made in scintillation counter techniques, greatly improved photomultipliers have been developed, and many materials, both organic and inorganic, have been examined in an effort to find better scintillation media.

In general, one would expect most substances, particularly those that are transparent or translucent, to luminesce. The absorption of radiation by any of the processes discussed above, whether by photo-electric, Compton, or pair production results ultimately in a spectrum of radiation extending into the infra red. The question is not so much why luminescence exists, but rather, why is it so rare. Most writers suggest that a quenching process occurs, due to the proximity of the atoms to one another in the crystal lattice. Luminescence, when it does occur, does so because of the presence of isolated luminescent centers. Bowen (20) has discussed the luminescence of organic materials irradiated with ultraviolet light, with a brief discussion on the effect of higher energy radiation.

Of greater immediate interest to the user of scintillation equipment are 1), the linearity of the relationship between radiation flux and light output; and 2), the energy dependence of the scintilla-

tion medium; and 3) the variation in intensity of light output with quantum energy of the incident radiation.

Scintillation detectors make use of two general techniques: 1), that in which each quantum of incident radiation produces a scintillation, resulting in a pulse of current at the anode of the photomultiplier, which is then amplified as a voltage pulse and counted on a scaler; and 2), that in which the pulse current of the photomultiplier is integrated and applied to a direct current measuring system. The former has been in most general use, especially at comparatively low flux levels, since it permits the measurement of both the intensity and the energy spectrum of the radiation (by pulse height analysis). At high flux levels however, such as those used in radiation therapy, many scintillations occur simultaneously, producing an almost continuous output of light from the scintillator. For such applications, the second method is used.

In general, scintillation detectors have not been used to any great extent at high flux levels, and no studies seem to have been made on the effect of high intensity radiation on scintillators. Bowen suggests that chemical decomposition may ultimately reduce the efficiency of a scintillator. He also mentions that a rise in temperature facilitates quenching. The present writer has found an apparent saturation effect using 250 kv X-rays at a dose rate of 80 R/min.; although this effect was not evident with cobalt gamma rays at 30 R/min. The effect was not investigated fully because the instrument was not intended for use at such high levels.

Air Equivalence of Scintillation Media

The unit of X-ray dosage, the roentgen, is defined in terms of an air ionization chamber. Any other material to be used as a radiation detector must be air equivalent if its response is to be expressed in roentgens. That is, its energy absorption must be proportional to the energy absorption of air over the energy range with which it is to be used. Ter-Pogossian and Ittner (8), Ittner, Ter-Pogossian and Aly (9), and Brucker have studied various materials in the energy range 30 kev to 250 kev; Carr and Hine (21), to 3 mev; and Belcher (7) to 30 mev.

From Eq. 5, the ratio of the energy absorbed in a medium to that absorbed in air may be stated as

$$R = \frac{1 - e^{\mu x}}{(\mu x)}_{air}$$

since $\mu \ll 1$ for air.

This ratio is plotted for anthracene (21) as a function of x-ray energy, and crystal thickness in Fig. 4. These curves are typical for organic phosphors. Comparison of these curves with Fig. 5, shows that for radium γ-rays anthracene and similar phosphors are essentially air equivalent.

Data for several organic phosphors are given in Table I (22). Anthracene is the material of choice on several points; it has the highest light yield, and the wavelength of the emitted light is fairly close to the maximum of the response curve of the photo-multiplier.

TABLE I

ORGANIC SCINTILLATORS

Scintillator	Density, gm/cm³	Effective atomic number, Z_r	Wavelength of emission, Å	Refractive index	Light yield	Decay time, 10^{-9} sec	Remarks
Anthracene	1.25	5.8	4450		1.00	25	Large crystals not quite clear
Quaterphenyl		5.8	4380		0.85	8	Pure crystals difficult to synthesize
Stilbene	1.16	5.7	4100		0.73	7	Good crystals readily obtainable
Diphenyloxazole		6.1			0.78		
Diphenylbutadiene		5.7	4610		0.67	8	
Diphenylanthracene		5.8	4800		0.65		
Terphenyl (para)	1.23	5.8	4150		0.55	12	Good crystals readily obtainable
Diphenylacetylene		5.8	3900		0.26–0.92	7	
Phenanthrene	1.03	5.8	4350	1.66	0.46	10	Clear crystals difficult to obtain
Naphthalene	1.15	5.8	3450	1.58	0.15	75	Good crystals easy to obtain
Chloroanthracene		9.8			0.03		

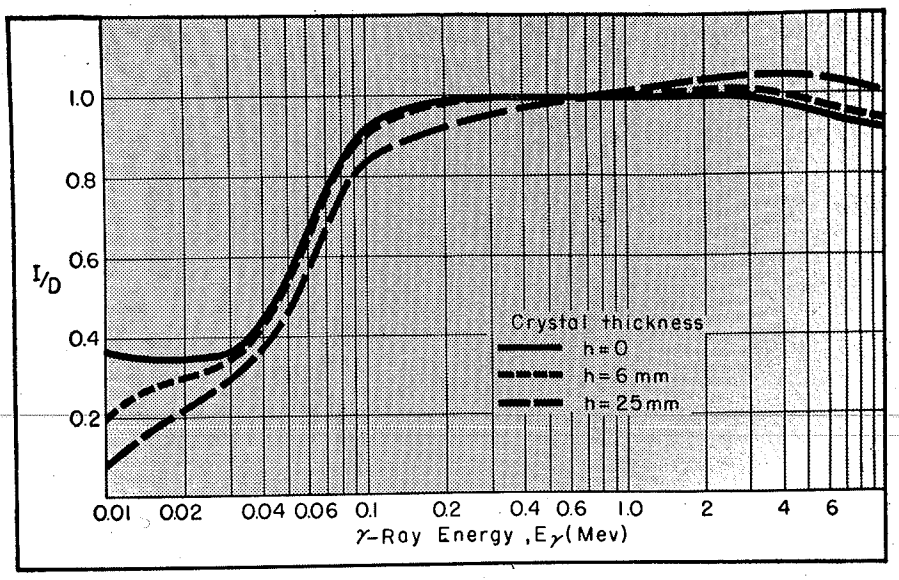


Fig. 4. Calculated anthracene to air ratio of energy absorbed per second as a function of gamma ray energy (21).

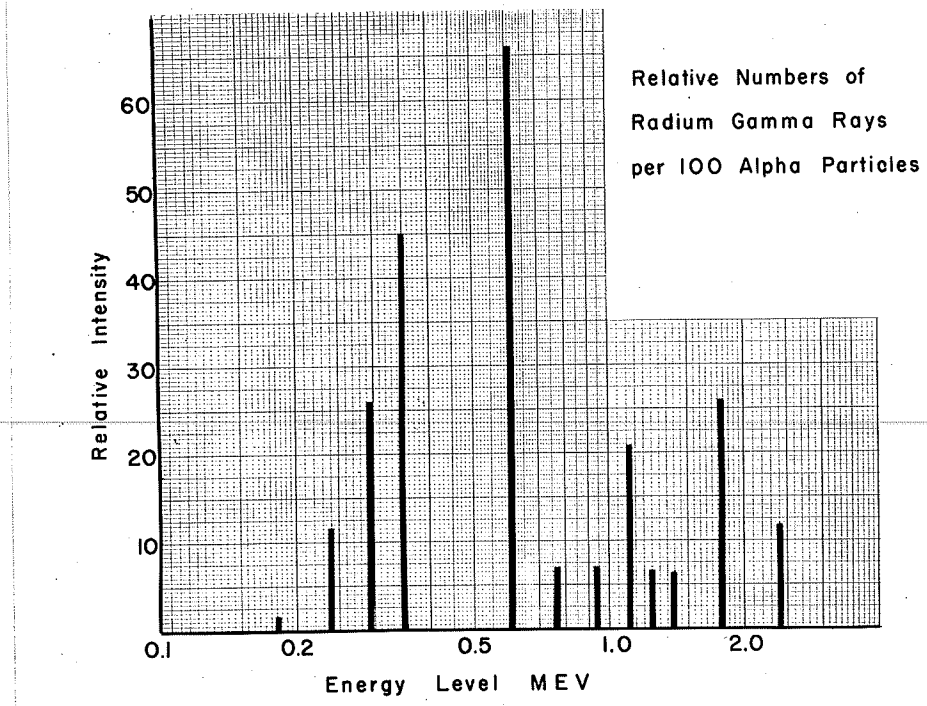


Fig. 5. Relative numbers of radium gamma rays per 100 alpha particles.

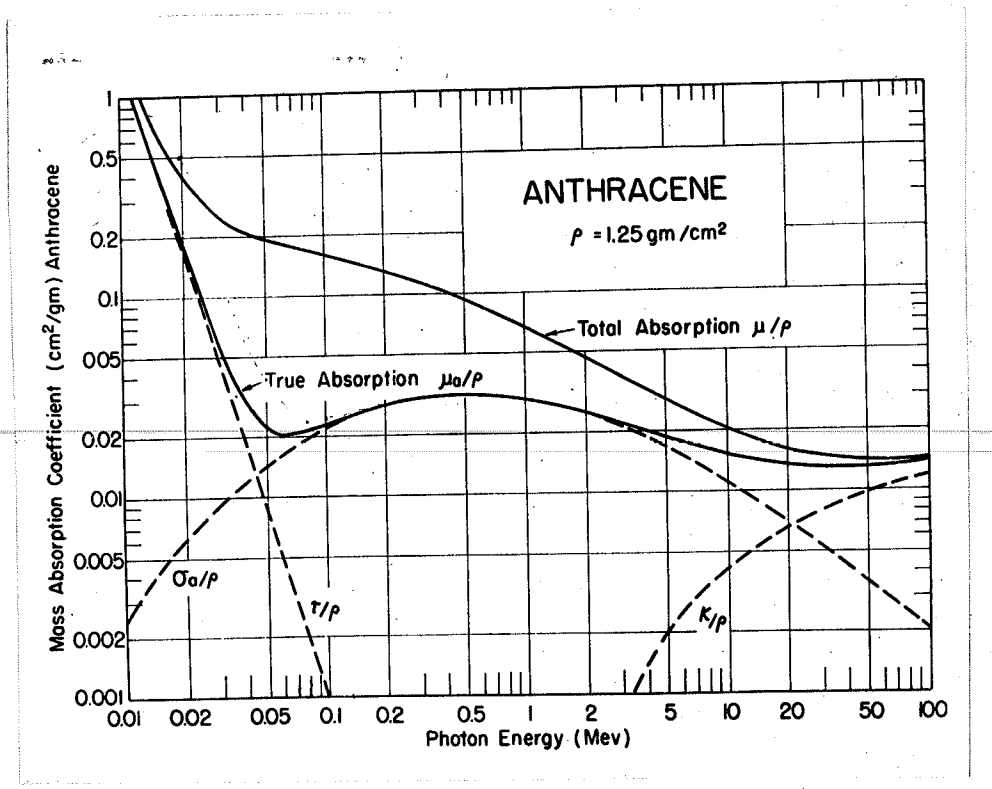


Fig. 6. Mass absorption coefficients for anthracene as a function of photon energy.

THE DOSIMETER

The dosimeter follows in external details that described by Belcher. With a fairly considerable literature available describing somewhat similar devices, it was obviously unnecessary to duplicate much of the preliminary work of other workers. Basically, the instrument consists of a scintillating crystal-phototube combination, the D.C. output of the phototube being applied to a D.C. amplifier whose output is displayed on a meter calibrated in roentgens per hour. The light from a phosphor crystal of small dimensions is conducted to the phototube through a highly polished metal tube of appropriate length. It was decided at an early stage to use anthracene as the phosphor; some plastic phosphors were made up, but they were discarded as being too insensitive. An acrylic light guide was tried, but it was found, as others had found before, that the acrylic contributed as much, or more signal as the phosphor proper. Since the phototube itself is highly sensitive to gamma radiation, provision must be made, in the form of a shutter, to cut off the light from the crystal, when the probe is in place, and balance out the effect of radiation on the phototube.

Probe. The probe is illustrated in exploded view in Fig. 7. It consists of an encapsulated anthracene crystal, cylindrical in shape, 5/32" diam., by 5/16" long. The crystal was smoked over burning magnesium, carefully dropped into the capsule, and secured in place with a thin, polished acrylic button. The capsule, of 1/4" dural is threaded and

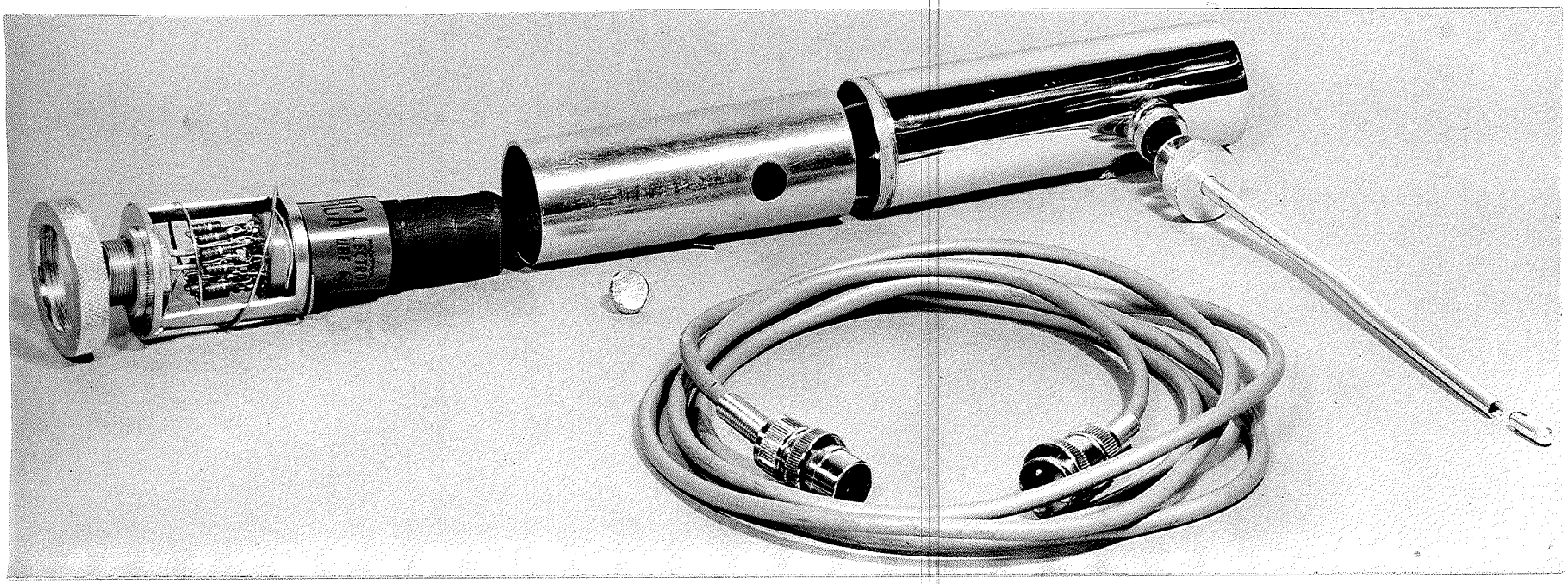


Fig. 7. Probe unit shown in exploded view.

fitted to one end of a 1/4" dural tube, 20 cm. in length, whose interior is brought to a high polish. The other end of the light pipe bears a flange which permits it to be securely fastened to the body of the probe unit by means of a threaded cap.

The probe unit consists of two concentric cylinders, the inner one being movable and spring loaded; this cylinder has an aperture, which in the normal position is opposite a corresponding hole in the outer cylinder to which the light pipe is fastened. In a radiation field, the inner cylinder, which acts as a shutter, may be moved by means of a button projecting through a slot in the outer cylinder, thus cutting off the light from the phosphor. The false signal resulting from the effect of radiation on the phototube may then be cancelled out, the shutter released, and a true reading taken.

The phototube, a photomultiplier type, may be a 1P21, 1P28, or a specially selected 931-A; it is chosen for high sensitivity and high signal to noise ratio. The phototube is mounted on a cylindrical bracket within which are mounted the dynode bleeder resistors. The base of the bracket is keyed to fit the outer cylinder and thus position the photocathode accurately with respect to the light guide. It is held in place by means of a threaded ring. A three-pin male Amphenol receptacle is fitted to the bracket base. In order to reduce leakage between the high voltage and signal pins, the bakelite interior of the standard Amphenol receptacle has been replaced with one constructed of Teflon (polytetrafluorethylene). Since the surface conductivity of the bakelite used in these fittings is lower than the volume conductivity, it was not necessary to modify the female connector, in which the con-

tacts are loosely held, and hence make only surface contact with the body of the interior. The same precaution was taken with the male connector at the other end of the inter-connecting cable.

Circuit. The instrument measures the integrated direct current from the photomultiplier. This amounts to approximately 10^{-8} A. for full scale deflection on the most sensitive scale (the Check scale, 10 roentgens per hour full scale). Electrometer techniques were indicated, without, however necessitating the use of a special electrometer tube, since many conventional tubes have grid current an order or two less than 10^{-8} A.

The requirements of any electrometer circuit are that it be linear over its full range, and be stable with respect to gain and zero. The circuit should be as simple as these factors permit. In almost all electrometer applications, the parameter measured is current. Some form of current-current transducer is indicated. Unfortunately in the present state of the art, the only current-current transducer, the transistor, is incapable of handling the minute input currents met with in practice. It might be pointed out that numerous current-current D.C. - A.C. transducers have been developed, but at the present they are either complex or expensive, or both.

The most generally used device is the vacuum tube, which of course, is a voltage- current transducer, and requires that the current to be measured pass through a very high resistance, and the voltage drop so produced applied to the control element of the tube. In order to operate satisfactorily in this application, the tube must have

a grid current of 1/10th or less of the current to be measured. Where currents of 10^{-9} A. and less are to be measured, special electrometer tubes must be used.

For a given grid bias, the plate current of any tube will depend on several factors, including tube geometry, applied electrode potentials, and contact potential. The latter may amount to 50 mv., and may not be compensated for except through the use of fairly complex circuit arrangements. This requires that a signal of about 1 volt be available if variations in zero level are to be minimized.

In the present circuit gain stability is assured by the use of a degenerative circuit giving 100% negative feedback. Variation in the plate supply is minimized by the use of voltage regulator tubes. Changes in heater current, due to variations in line voltage, produce changes in zero level, and a regulating transformer for the heaters was employed initially. However, in order to reduce weight and bulk, this was replaced by a compensating circuit which will be described later.

The basic circuit shown in Fig. 8 derives from that of Moody (6). This particular type of circuit requires that the input source be an infinite impedance, or constant-current source, such as an ionization chamber or photomultiplier.

The gain of V_1, without feedback, is

$$A_1 = \frac{\mu R_L}{R_L + R_p}$$

where μ is the amplification factor

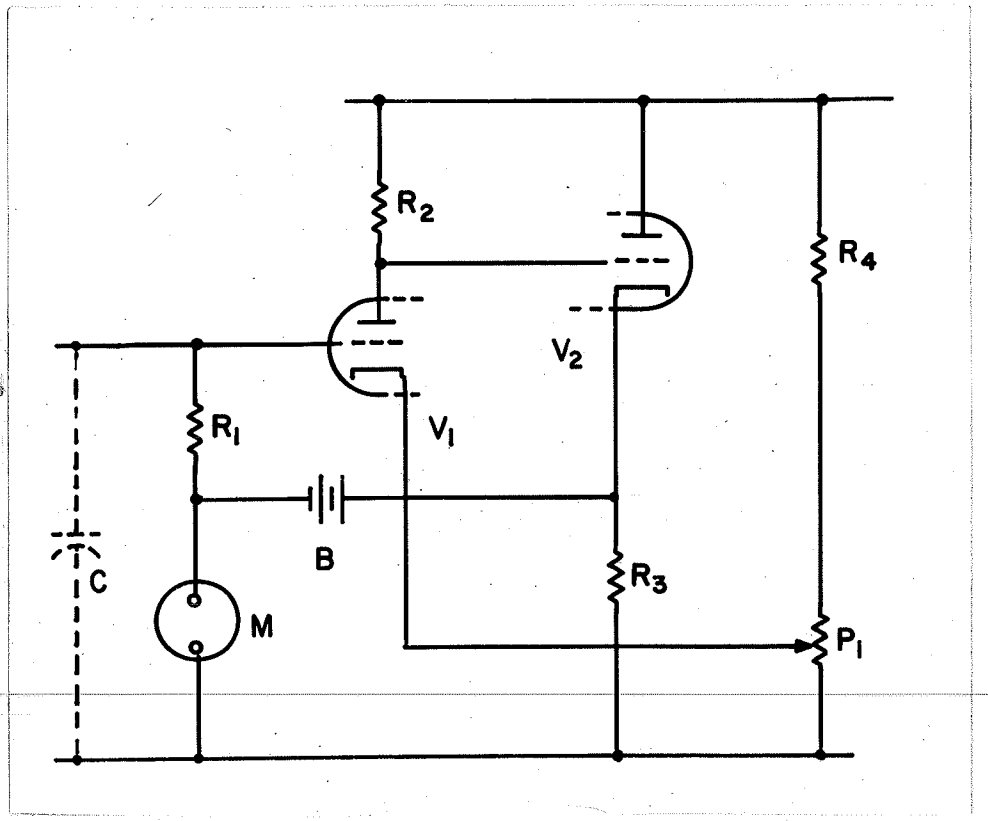


Fig. 8a. Basic electronic circuit.

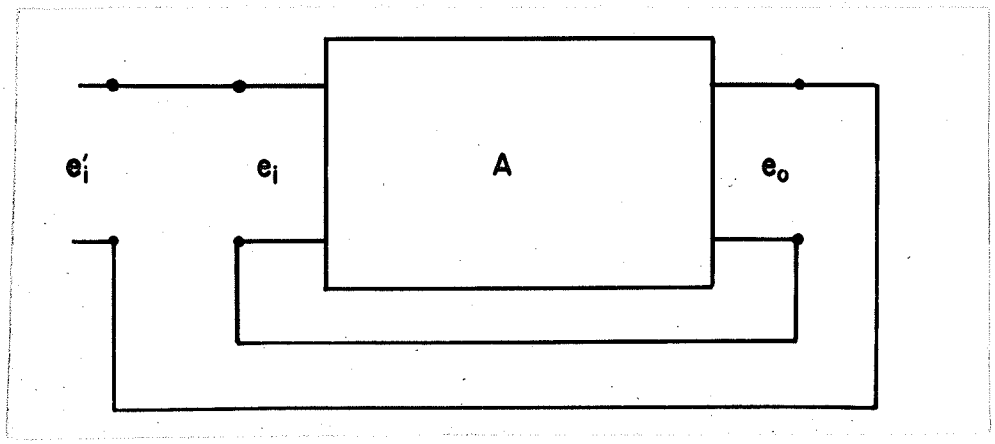


Fig. 8b. Block diagram showing feedback loop.

R_L the load resistance

R_p the plate resistance

The gain of V_2, without feedback, is

$$A_2 = \frac{\mu \ R_k}{(\mu + 1) R_k + R_p}$$

where R_k is the total cathode resistance.

The overall gain, then, is

$$A = A_1 A_2 = \frac{\mu^2 R_L R_k}{(R_L + R_p)\left[(\mu + 1) R_k + R_p\right]}$$

The circuit with feedback is shown in Fig. 8b.

$$e_o = Ae_i$$

$$e_i = e_i' - Ae_i$$

$$e_i' = e_i (1 + A)$$

$$e_o = Ae_i = \frac{A}{1+A} e_i'$$

The gain with feedback, then, is

$$A' = \frac{e_o}{e_i'} = \frac{A}{1+A}$$

For this circuit, V_{1a}, V_{1b} is a 12AT7 double triode, for which

$\mu = 60$

$R_p = 15{,}000$

$R_L = 2.2 \times 10^6$

$R_k = 5000$, the resistance of the meter,

hence
$$A = \frac{60^2 \times 2.2 \times 10^6 \times 5 \times 10^3}{(2.2 \times 10^6 + 15 \times 10^3)(61 \times 5 \times 10^3 + 15 \times 10^3)}$$

$$= 56$$

and $A' = \frac{56}{57} = 0.98$

From this it will be seen that a drop of 1.0 volt across R_1 will produce a signal of 0.98 volt at the cathode of V_{1b}. The effects of non-linearity and unwanted signals generated within the feedback loop will be reduced by a factor of 56.

The time constant of the circuit R_1C, where C is the stray capacity, is also reduced; the time constant is

$$T = R_1C / (1+A)$$

This is of considerable importance in circuits designed to measure currents less than 10^{-12} A., where R would be in excess of 10^{12} ohms. With a typical value of C taken as 20μμf, we should have a time constant of 20 seconds without feedback.

In the present circuit, with the maximum value of $R_1 = 10^8$ ohms, and assuming C = 20μμf (20×10^{-12}f),

$R_1C = 20 \times 10^{-4}$ sec. without feedback, and only

$\frac{20 \times 10^{-4}}{56} = 3.6 \times 10^{-5}$ sec. with feedback.

It should be noted that capacity across R_1, i.e., between the input grid and the feedback circuit is not affected by the feedback. Advantage is taken of this in the present circuit by inserting a capacitor across R_1, to smooth out random variations when using the check scale.

Practical Circuit. The following factors were of importance in the design: 1. The instrument was to be line operated, from the regular 117 v. 60 cycle outlet; 2. Standard, easily obtained components, in-

cluding vacuum tubes were to be used (the only component not readily available from local suppliers is the high voltage corona regulator, V3); 3. All aspects of the circuit were to be reduced to the utmost simplicity.

Sufficient regulation is furnished by the gas type voltage regulator tubes, V6 and V7, to obviate the need of a more complex, feedback regulator. The gas tetrode relaxation oscillator in the high voltage supply is, of course, well known. The 167 transformer used is a small filament transformer, driven from its low voltage winding. R13, the surge suppressor, limits the anode peak current to about 1 ampere, thus ensuring reasonable tube life. Placing the rectifier in the negative lead permitted the use of a common heater supply for all tubes except V8. The 6X4 has a heater-cathode rating of 450 volts, but since a five volt winding was available on the transformer, it was decided to take advantage of the additional factor of safety, and the reduced heater voltage contributes to longer tube life.

Measuring Circuit. Previous applications of this basic circuit had all been battery operated. Moody's original circuit used four B batteries as well as two filament batteries for the amplifier alone. V2, a 50 - 60 volt neon lamp supplies a voltage drop approximately equal to the resting potential of the anode of V1a, thus permitting the full signal from the cathode of V1b to be used for feedback.

The circuit at this point functioned adequately, but was subject to changes in zero level with changes in heater current. It was not desirable to add the considerable weight of a regulating transfor-

mer, and so the compensating circuit was devised. This circuit injects a potential into the cathode circuit of V1a sufficient to balance changes due to line voltage. The correct amount is determined by trial, and the adjustment of P3. The large filter capacitor was so chosen, that for an abrupt change in line voltage, the compensating voltage changed at the same rate as the effect of the heater temperature change. Over the normal range of line voltage variation, it was possible to compensate so that zero drift was negligible.

A standard source was constructed for maintaining the calibration of the instrument. A five milligram radium plaque was mounted in one end of a plastic cylinder, and an axial hole, just greater than the diameter of the probe drilled from the other end, deep enough to give a calculated dose rate of something less than five roentgens per hour. The plastic cylinder was then fixed into a lead shield.

The instrument was calibrated in air, using the average of several 10 milligram radium needles. Using the instrument as its own standard, with frequent checks against one of the radium needles, the hole in the plastic was deepened until the meter read five roentgens per hour with the probe fully inserted.

The accuracy of calibration of the radium needles is generally accepted to be 1% or better, so the limiting accuracy for radium in the air is probably that of the meter, which is about 3%. The scattering produced by the tissue would not introduce an error of more than 5% over the radium-probe distances met with in practice. An accuracy of ± 10% is adequate for the purposes for which the instrument was designed.

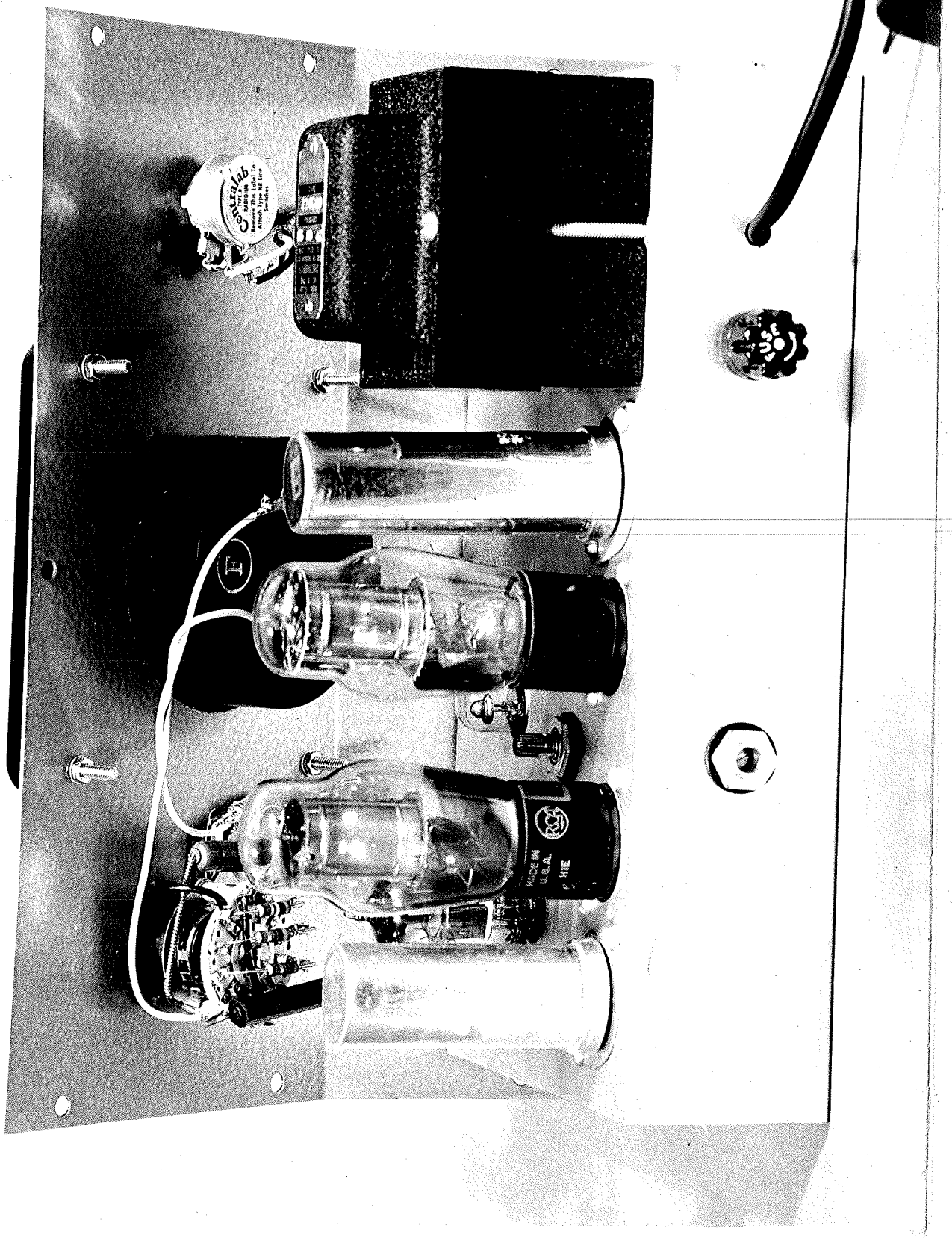


Fig. 9. Rear view of chassis.

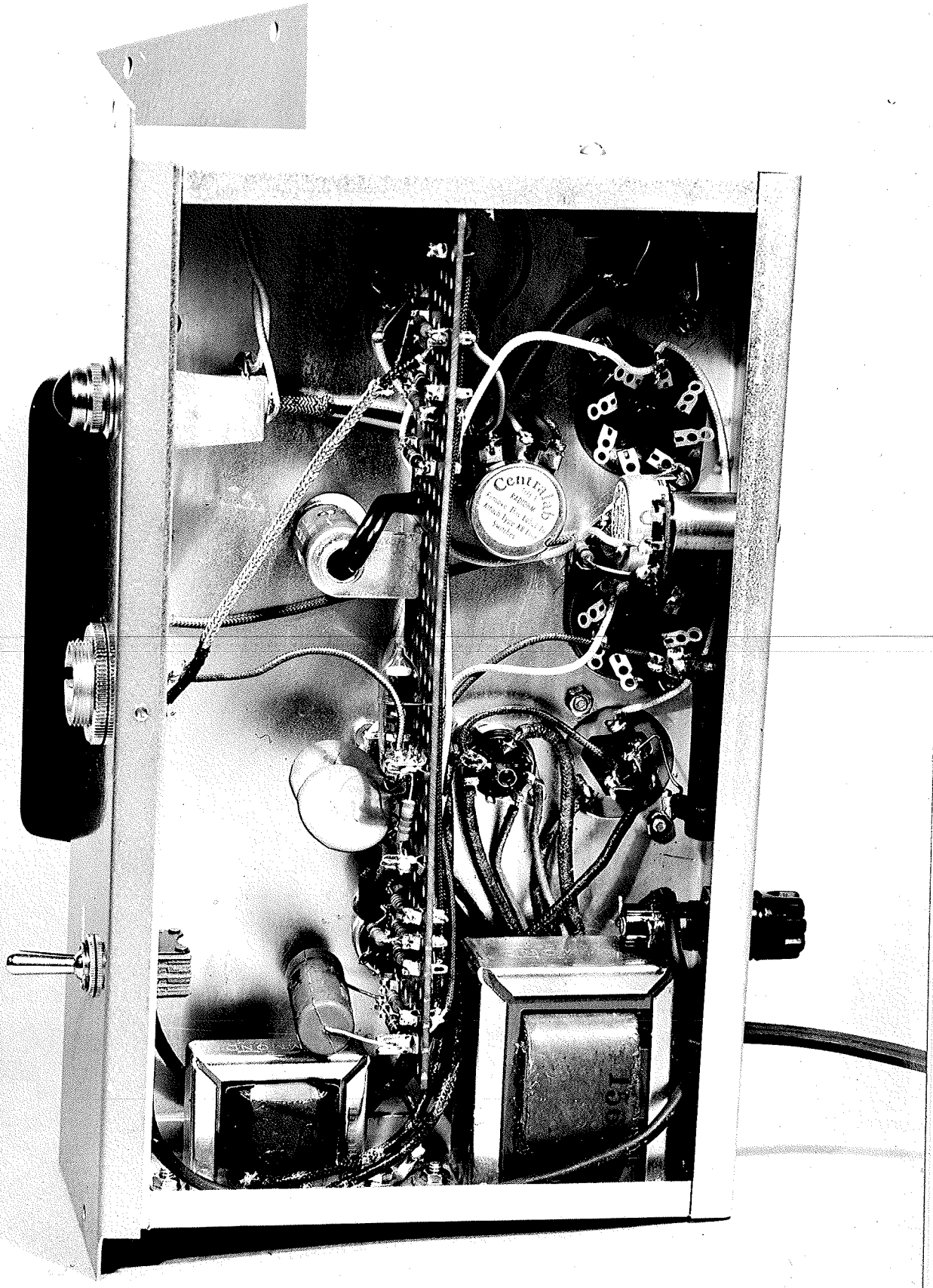


Fig. 10. Bottom view of chassis.

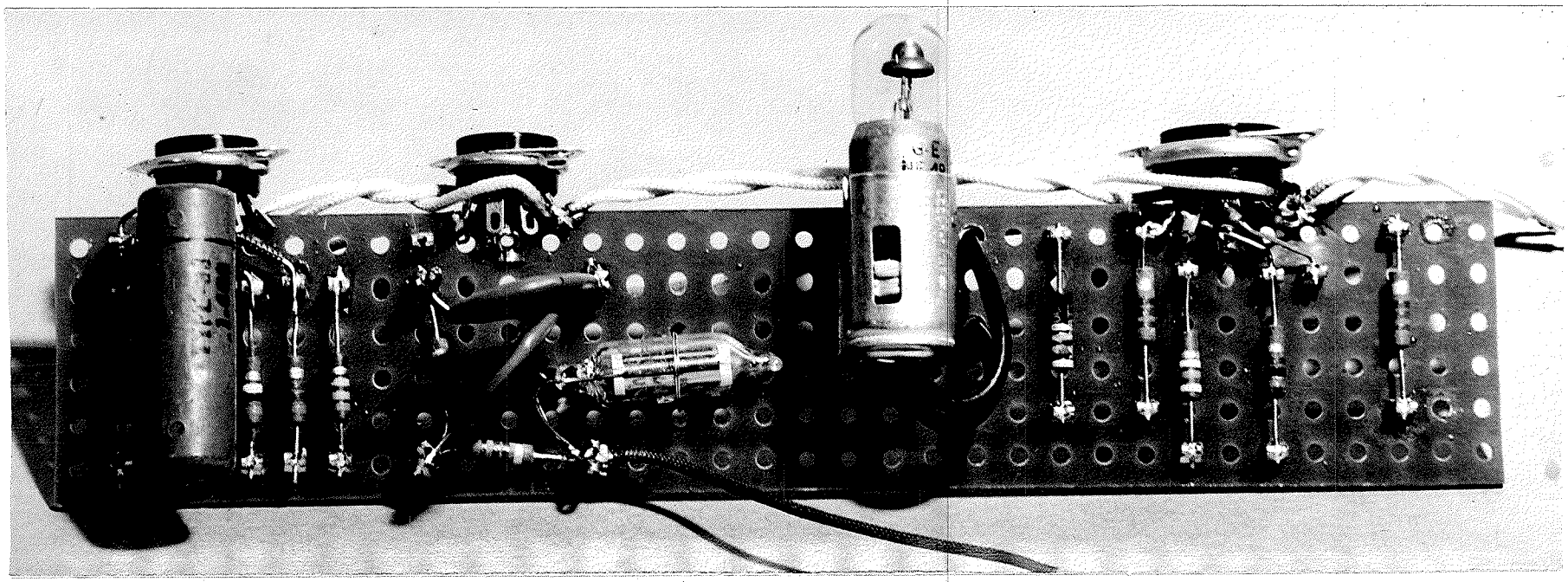


Fig. 11. Electronics component board, front.

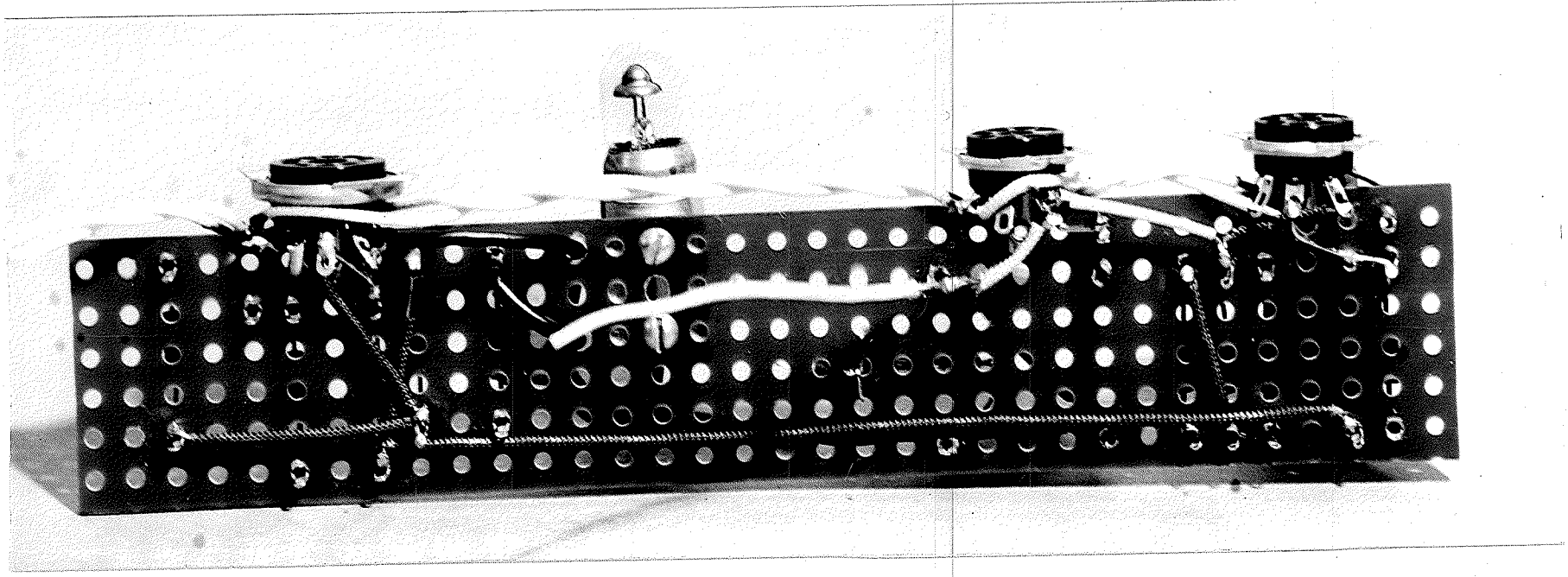
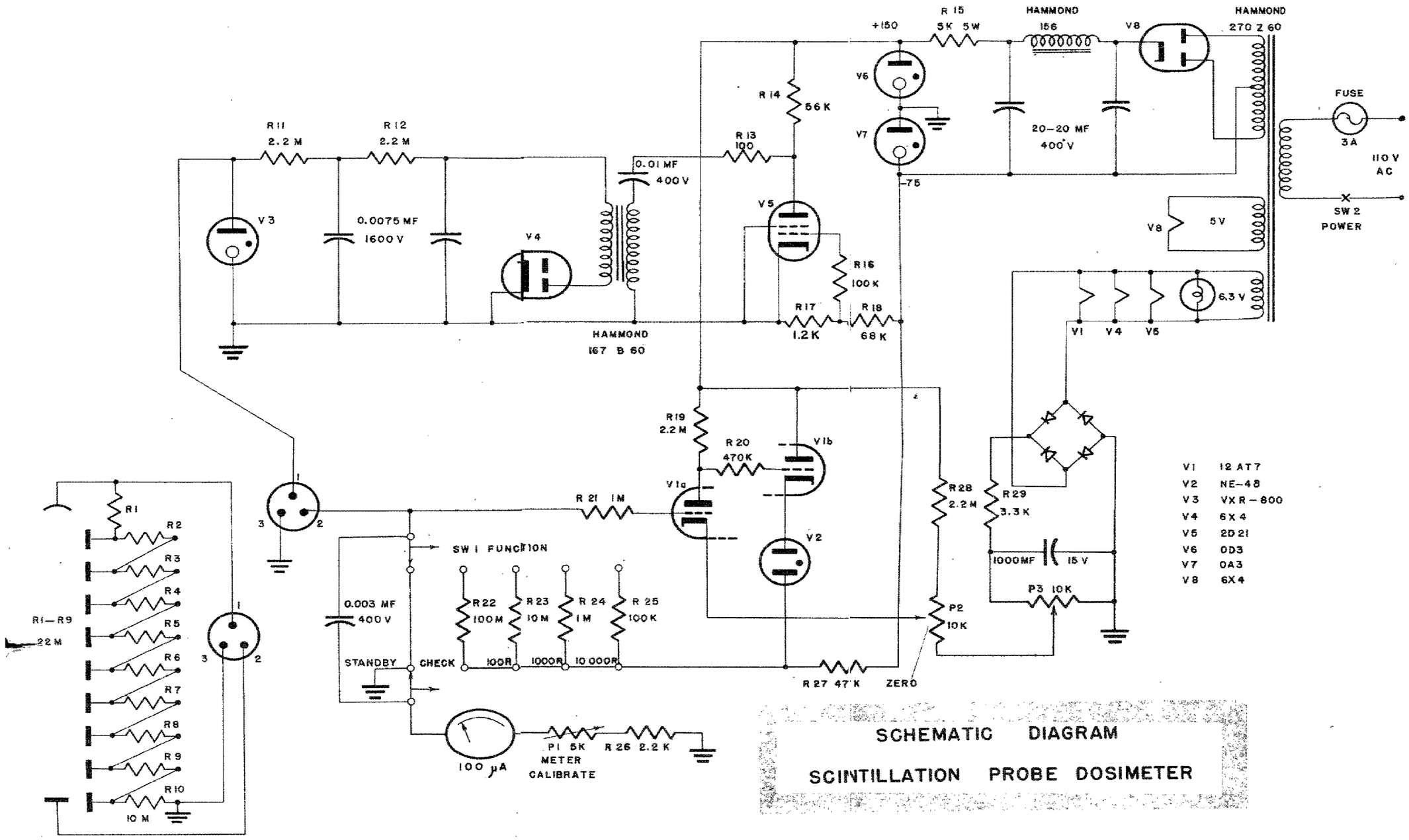


Fig. 12. Electronics component board, back.

Fig. 13. Completed instrument.



V1	12 AT 7
V2	NE-48
V3	VXR-800
V4	6X4
V5	2D21
V6	0D3
V7	0A3
V8	6X4

SCHEMATIC DIAGRAM

SCINTILLATION PROBE DOSIMETER

CLINICAL APPLICATION OF THE PROBE TYPE DOSIMETER

The problem faced by the radiologist is two-fold: He must first assess the tumour as to size and location, and determine the amount of radiation which, in his opinion will destroy it, or retard its growth; then, usually with the assistance of a clinical physicist, he must determine how best to achieve the prescribed dosage with minimum radiation applied to the surrounding healthy tissue. There are numerous sources of radiation available to the radiologist today. These include X-ray machines with energies ranging from 10 KEV, to 10 or more MEV; teletherapy units utilizing radium, cobalt 60 or other radioactive isotopes; as well as direct contact materials giving alpha, beta or gamma radiation, or a combination of them, such as radium, radioactive gold, cobalt, phosphorous, and other radioactive isotopes.

The dosage due to X-ray and teletherapy units may be calculated fairly readily with the aid of isodose curves, and other techniques, especially where fixed fields are used. This is because, with the source-tumour distances used, the field changes fairly gradually, except at the edges of the field. When the source of radiation is placed on the surface in the form of a plaque, or within the body, calculation becomes much more difficult. For an implant or insertion the field has a square-law distribution, and while the dose may be extremely high in the immediate vicinity of the source, it is attenuated very rapidly. It is of great importance, therefore, that the limits of the tumour be known with considerable accuracy, so that it will receive

adequate radiation at its boundaries. It is important too, that the location of any nearby organs be accurately known in order that they not be damaged by too much radiation.

A radiograph of the site may give the required information; usually, however, it is necessary to visualize the organ with a radio-opaque substance. This can become a complex procedure, and in many cases is not done. This is particularly true in the radium treatment for cancer of the cervix uteri. The insertion in this treatment is carried out in the operating room, with the patient under anesthesia. While radiographs are made of the pelvis as a matter of routine, generally speaking this must be done later in the diagnostic radiology department since the X-ray machines in the operating room do not have sufficient penetration. The radiologist does not normally see the radiographs until some two or more hours after the insertion. Even had the adjacent organs been visualised by means of a radio-opaque material, it would not be practicable to alter the insertion unless the radium had become grossly displaced.

It was for this reason, primarily, that the development of the dosimeter described was undertaken. With the instrument, it is possible, at the time of insertion; to determine the dose at the walls of the rectum and bladder, whose mucous membranes are particularly susceptible to radiation damage. The instrument further permits the radiologist to determine whether the radium has become displaced by the packing procedure.

Fig. 15 shows a posterior - anterior radiograph, and Fig. 16 a lateral radiograph of a typical insertion, using two 50 mg. radium

needles in tubes in the uterus, and three 10 mg. needles in each of two ovoids in the vaginal fornices. The ovoids are fixed in position by their attachment to a colpostat. In addition, the urethra and bladder have been visualized by means of a catheter, and the rectum by the admission of air.*

The areas of particular interest to the radiologist are:

1. The site of the disease itself.

2. The areas to which the disease may spread, i.e., the parametria, or regions immediately lateral to the uterus; and the lymph nodes at the side walls of the pelvis; and

3. The walls of the bladder and rectum, and the orifices of the ureters.

The probe is first passed into the bladder until it touches the base of the latter, and a reading taken; as shown in Fig. 17 then it is moved laterally and a reading taken at each side. The probe is then passed into the rectum against the anterior wall at the midline, as shown in Fig. 18 and a reading taken of the maximum. It is then moved laterally and readings taken at the left and right limits.

The readings for the case shown in the radiographs, which are typical, are given below.

Bladder

Midline	56 roentgens per hour
Left	38 roentgens per hour

*This procedure was very kindly carried out for the purpose of making these photographs by Dr. J.M. Gillies of the Manitoba Cancer Foundation.

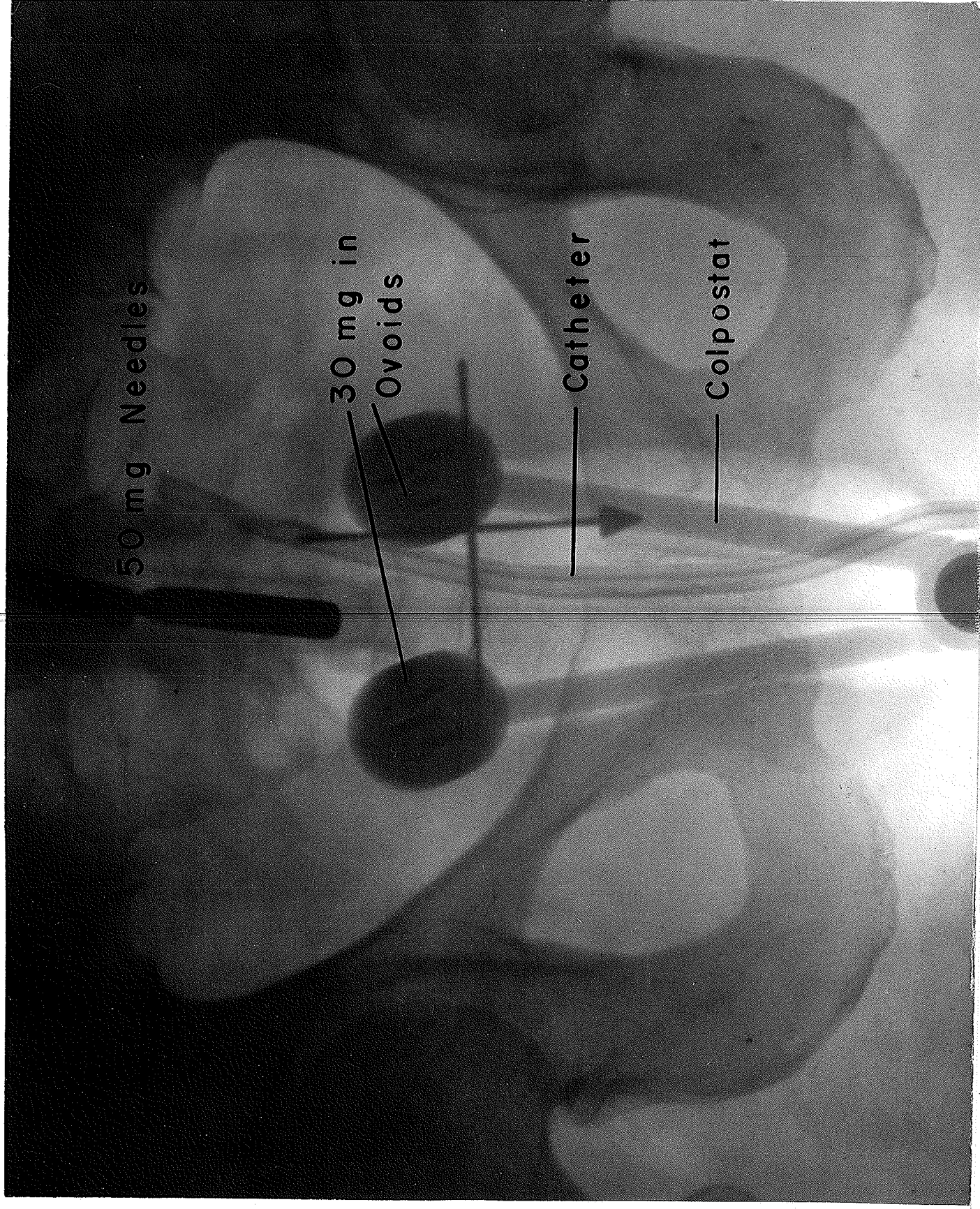


Fig. 15. Posterior - anterior radiograph, showing two 50 mg radium needles in the uterus, and two ovoids of 30 mg each in the vaginal fornices.

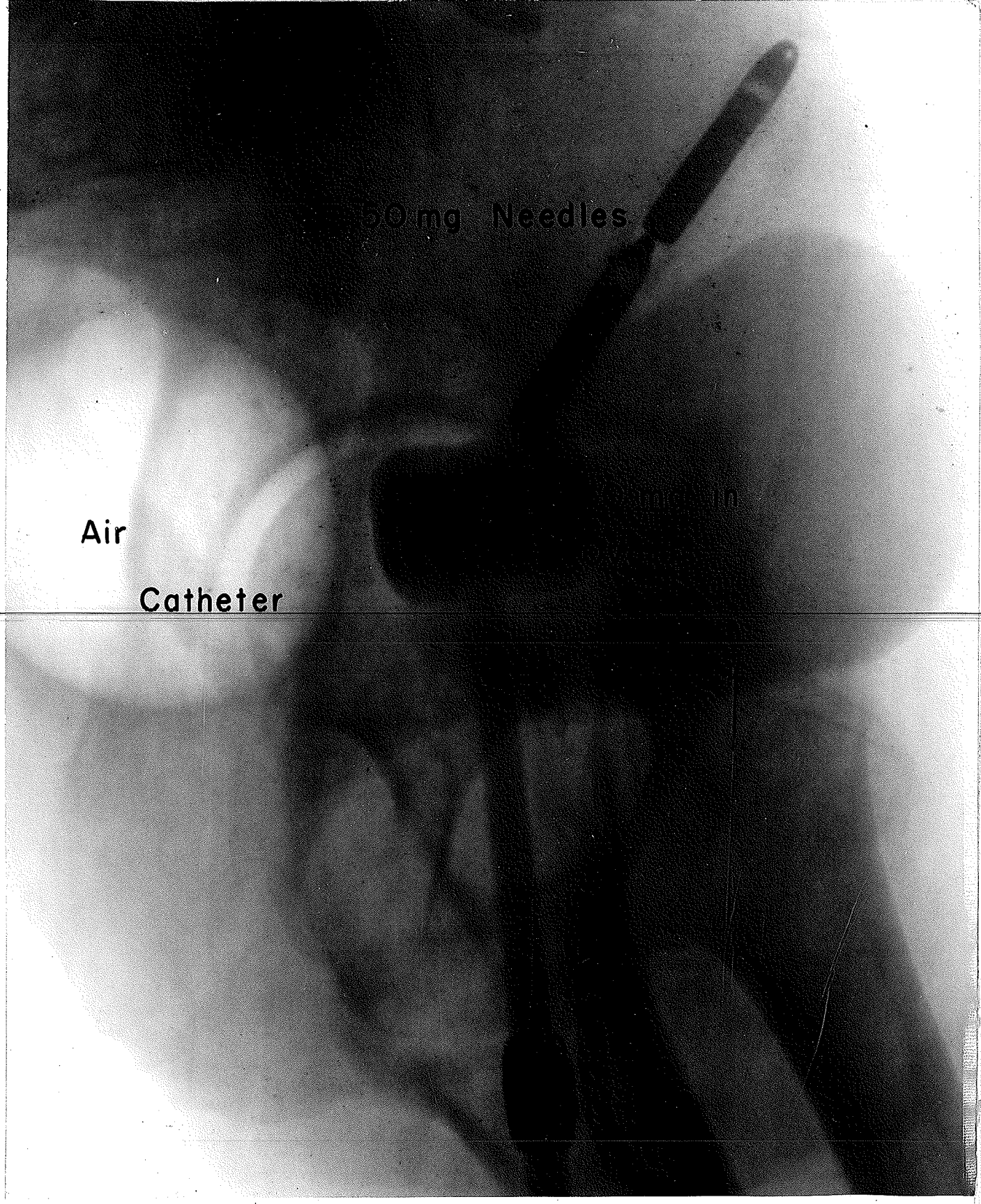


Fig. 16. Lateral radiograph of patient of Fig. 15.

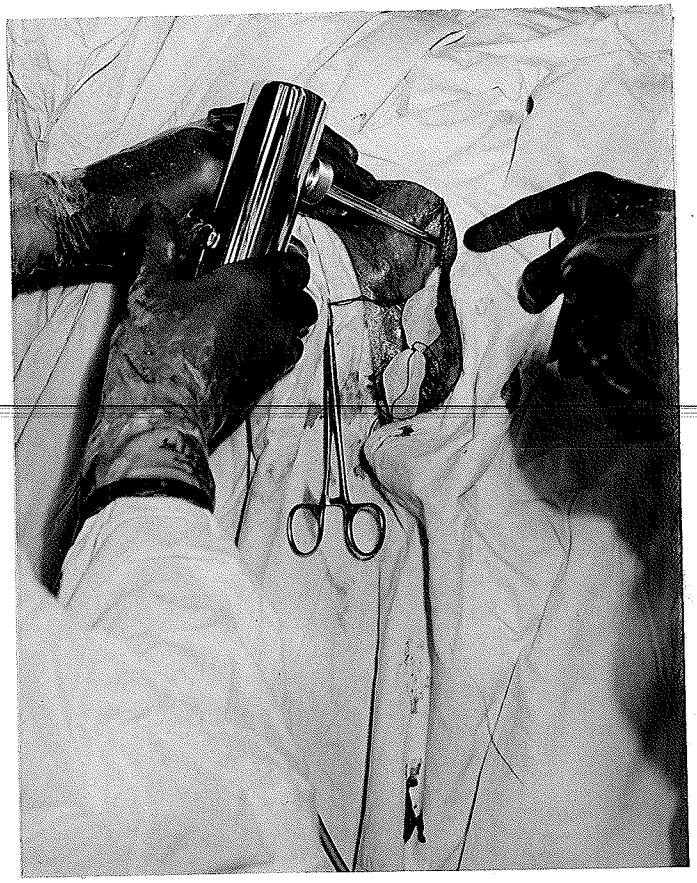


Fig. 17. Probe being inserted into bladder.

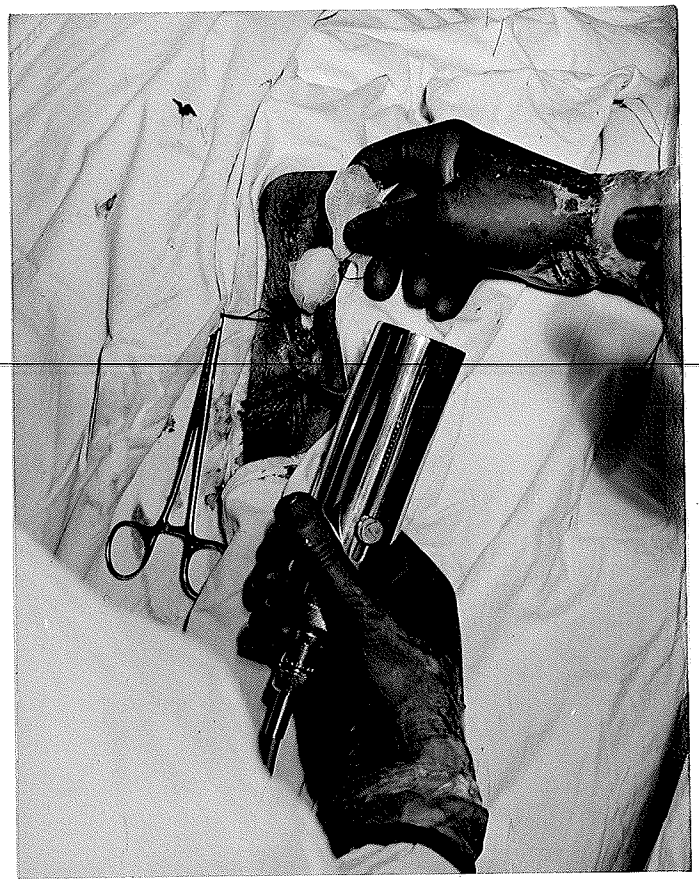


Fig. 18. Probe being inserted into rectum.

Right	56 roentgens per hour
Rectum	
Midline	50 roentgens per hour
Left	60 roentgens per hour
Right	38 roentgens per hour

"This technique measures directly the dosage of irradiation being given to bladder and rectum. Although small differences in the rate of radiation are, as yet, not recordable, gross overdosage of the base of the bladder, and anterior rectal wall can be detected. This allows immediate re-positioning of the radium.

"The dosimeter permits greater individualisation of treatment. It appears to be appreciably reducing the incidence of bladder, urethral and rectal radiation injuries.

"While its field of applicability is, at present, restricted to carcinoma of the cervix, this field may be widened as more experience is gained in its use".*

* Dr. Jean MacFarlane, personal communication.

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