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APPLICATIONS OF PROPORTIONAL CHAMBERS TO SOME

PROBLEMS IN MEDICINE AND BIOLOGY

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1. INTRODUCTION

Multiwire proportional chambers (MWPCs) and drift chambers (DCs) are now routine techniques used in nuclear and subnuclear physics.

In the field of charged particle detection, instant development followed the display in 1968 of the properties of multiwire structures^{1,2)}: high rate capability; fast resolution times; high spatial resolution, determined either by the wire spacing or by the space-time correlation of the pulses. The dramatic development in electronic techniques matched perfectly the needs of the detectors.

In the field of neutral particle detection (X-rays and neutrons) the initial investigations²⁾ already revealed promising features for the imaging of spatial distributions, such as

- self-triggering;
- two-dimensional read-out: the secondary radiation produced by gammas or neutrons is usually of a small energy and cannot cross several chambers. The read-out of the avalanche position along the wire was shown to be possible using the signal induced on cathode elements perpendicular to the anode wires.
- The electrons liberated in drift spaces attached to a chamber are easily transferred to the proportional chambers. This opens the way to different methods of increased efficiency by conversion, in appropriate drift spaces, of the neutral radiation.

The efforts of a rather small number of groups have succeeded in pioneering new fields of applications of MWPCs, and some of their results are quite encouraging.

In the last ten years a variety of detector structures have been investigated in high-energy physics laboratories. Considerable progress has been made in dealing with construction problems and in improving the reliability of these detectors. It may be that this is precisely the right time to see whether these treasures of technical know-how can be incorporated into fields where they could bring about some significant progress.

In 1975 Perez-Mendez³⁾ reviewed some applications of MWPCs. A detailed study of the present state of the art and of possible advantageous applications of MWPCs to some specific problems in the most important field of radionuclide imaging has recently been made by a group of physicists and medical doctors from MIT and the Harvard Medical School⁴⁾. Their references cover most of the efforts in this field.

This review talk aims at drawing the attention of nuclear physicists to possible applications of some familiar techniques, and discusses examples of applications.

2. APPLICATION OF PROPORTIONAL DETECTORS TO PROTON RADIOGRAPHY

I will discuss only the method of nuclear scattering radiography, leaving aside that of proton absorption.

With this type of radiography^{5,6)}, protons of about 1 GeV are scattered by the tissues. Detectors are used to determine the trajectories of the scattered and recoil protons. The density of the interaction vertices reflects the density of the matter and is also a function of interesting parameters such as the hydrogen content or the chemical composition.

The interest of the method relies on several factors:

- it gives information on the whole volume without requiring any rotation of either the object or the beam;
- it has an intrinsic resolution limited by multiple scattering and detector accuracy of about 1 mm³;
- it may display biological information that is hidden from X-rays, since the density of the vertices is not only a function of gravity but also of isotopic composition.

Despite the considerable drawback of requiring a high-energy proton beam, the investigations gained incentive from the fact that in several laboratories beams of this type are used for therapy, and it would be natural to use a fraction of the intensity for radiography if this would prove superior to X-rays.

It is a very challenging project, as is obvious from the required characteristics: spatial accuracy better than 1 mm; rate of recorded events 10⁶/sec. The construction of such an instrument will be undertaken by a CERN-Saclay Collaboration. It aims at obtaining a radiograph of a head within at most a few minutes. The preliminary results so far obtained with a much slower device ($\sim 10^3$ events/sec) on inanimate tissues are encouraging. Figure 1 shows the cross-section of a head; this was done at CERN in December 1977.

3. THE IMAGING OF X-RAYS OR γ -RAYS

The critical parameters dictating or barring the choice of MWPCs for these applications are the efficiency, the spatial accuracy, and the time resolution. The efficiency of gaseous absorption drops rapidly with energy. Figure 2 shows the mean free path for absorption in some noble gases at 1 atm.

The accuracy is rarely limited by the intrinsic accuracy in avalanche localization, which can be better than 100 μm and which can be obtained by several read-out methods. It can be limited by the range of the secondary electrons or by the free path of the fluorescent X-rays associated with the photon absorption in the gases. Figure 3 shows the range of the electrons produced in argon and xenon by X-rays of variable energies. Several Auger electrons usually follow a photoelectric absorption, and the electron with the highest range determines the accuracy.

The problems raised by the use of MWPCs to study spatial distribution of X-rays are strongly dependent on the energy of these X-rays. The fields of application are also very varied, and we will consider these different domains of energy separately.

3.1 Imaging of soft X-rays (5 keV to about 30 keV)⁷⁻¹²⁾

Above 5 keV and up to a few tens of keV a very important domain is the one connected to the structure analysis using X-ray diffraction. The study of protein structures using protein crystals is one of the most promising in this field. The replacement of photographic films or of small scanning counters as X-ray detectors offers the possibility of gains by factors of 100 or 1000 in the time so far necessary to obtain the data on molecular structure. New domains, such as rapid variations of structures, can be opened up by these techniques.

We see clearly from Fig. 2 that to obtain efficiencies close to 100% we have to go to heavy gases and to large thicknesses well above 1 cm, or to high pressures. Large thicknesses are a nuisance because of the degradation of the accuracy for inclined tracks, as shown in Fig. 4; directions inclined with respect to the chambers give rise to elongated images. For instance, this has led Cork et al.^{7,8)} to the construction of 8 mm thick chambers filled with 5 atm of xenon. However, although their chambers have a limited accuracy of a few millimetres -- a small surface -- this pioneering work may well be, to date, the best application of MWPCs outside high-energy physics. By gaining a factor of 50 over the speed of conventional X-ray diffractometers, this group has made an important advance in biological techniques.

At CERN we have been led to another solution, determined by the requirements of two users: R. Mössbauer, who needs imaging chambers for 15 keV X-rays; and a Paris group using the synchrotron radiation emitted by the electron storage rings at Orsay. Electron storage rings are modern, powerful sources of X-rays having the following characteristics:

- The spectrum is continuous and very intense. With curved crystals a narrow bandwidth (~ 2 eV) can be selected between 0.5 and 3 $\text{\AA}$ (25 keV to 4 keV),

which is an ideal wavelength for crystalized protein structures. The intensity in such a narrow bandwidth can be 10^4 higher than with a standard sealed copper X-ray tube. A realistic evaluation of the running conditions usually leads to a maximum flux of photons, diffracted by small protein crystals, of $\sim 10^6/\text{sec}$.

- The beam geometry is very good. The divergence is ~ 1 arcmin in the vertical plane and 3 arcmin in the horizontal plane. Long crystal-to-detector distances are possible, which is very important for crystals with large cells.
- The tunability of the source is a very powerful tool, since one can vary the wavelength below and above the K edges of the elements composing the sample.

As a means of making full use of the qualities of the synchrotron radiation, it appears that MWPCs are ideal, both in terms of accuracy and rate capability. To circumvent the source of error due to the converter volume thickness (see Fig. 4), conversion and drift spaces having spherical or cylindrical symmetry have been developed⁹⁻¹²). We describe here some results obtained in a large prototype spherical drift chamber¹⁰). On one side of a conventional two-dimensional imaging MWPC, a gaseous conversion volume is mounted in which a suitable set of electrodes maintains an electric field with radial symmetry, its centre being the crystal position, as shown in Fig. 5. The upper electrode consists of a thin, gas-tight aluminium cap, while the lower spherical electrode is a thin grid to let the ionization electrons, produced in the drift space by photoelectric conversion, drift smoothly to the lower imaging chamber. The chamber we have operated at CERN has a 10 cm thick radial drift space and an acceptance of 87° . The associated MWPC has an active area of $50 \times 50 \text{ cm}^2$ and 2 mm anode wire spacing. Use of the data in Fig. 2 gives the efficiency for 8 keV X-rays in argon (84%) and in xenon (100%).

For the imaging, we have used the method of measuring the centre of gravity of the induced positive signals on the cathodes with analog circuits¹⁰), requiring a few microseconds to compute the centroid position.

One of the apparently surprising results is the fact that the accuracy of the chamber, measured with a collimated beam of 8 keV X-rays, is better than $700 \mu\text{m}$ (FWHM) in all directions, while in the direction orthogonal to the anode wires it should be equivalent to their spacing (2 mm). The originally very localized primary charge, due to diffusion during the drift, reaches a size between 1 mm and 2 mm, and the energy is split between two adjacent wires thus resulting in a precise determination of the charge distribution between the wires.

This is very rewarding since, from our experience, chambers with 1 mm wire spacing and 50 cm length are difficult to operate, while 2 mm spacing is compatible with sizes of several square metres, as we have learned from our experience in high-energy physics.

The possibility of operating over large surfaces is also connected with the possibility of using large-size spherical drift spaces (up to 50 cm, for instance), which in many cases would eliminate the need to use xenon as a component of the filling gases.

At higher X-ray energies, the accuracy of a chamber is limited by the range of the photoelectrons, which at 20 keV is of the order of 1 mm in an argon-filled chamber; in a xenon-filled chamber the range is 1 mm around 40 keV. We may thus consider these energies as practical upper limits for chambers at atmospheric pressures, if 1 mm localization accuracy has to be maintained.

Examples of X-ray diffraction patterns in a spherical drift chamber have already been published¹⁰⁾. The chamber can also be used as an X-ray pinhole camera by placing an absorber with a small hole at its focal point.

Another field of application for this kind of detector is the structural study of muscular tissues. In certain conditions, muscle fibres form a regular pattern of large molecules, somewhat similar to a crystal; characteristic diffraction patterns can be produced that are sensitive to spacings down to a few tens of angstroms. The study could be static or, if the data acquisition rate is high enough, dynamic¹³⁾. Figure 6 (from Ref. 13) shows a rat-tail collagen one-dimensional pattern obtained with 10 keV X-rays. One-dimensional cylindrical chambers, with a rate capability of 100 MHz, have been produced at CERN by our group for the Orsay synchrotron radiation facility.

3.2 Imaging of γ -rays (30 keV to 1 MeV)

This is the most important field in nuclear medicine. Imaging of nuclear radiation is a widely used diagnostic tool. In most cases, suitable radioactive tracers are injected into the patients, and static or dynamic *in vivo* distribution of the tracers can give precious information. The dramatic progress introduced in X-ray imaging by computer-assisted tomography (CAT) has not reduced the interest in these methods. The tracers are usually synthesized into chemical compounds having specific metabolic properties. The knowledge of their distribution gives information on specific functions, while an X-ray transmission image gives information on structural configuration.

In a few cases, radioactive sources are used as external X-ray sources with imaging problems very similar to the case of internal sources.

The problem is to minimize patient irradiation while getting the best information. The efficiency, the spatial resolution, and the maximum counting rates are the important parameters to optimize. The instruments most used so far are based on sodium iodide scintillation counters, and the most useful and widely known tool is the Anger camera. The detector is typically a 1.3 cm thick NaI crystal of 40 cm diameter, viewed by many photomultipliers.

When a scintillation is produced in the crystal, the photomultipliers register a fraction of the total light, which is position-dependent. Computation permits the initial position to be localized within times of the order of 1 μ sec and an accuracy of the order of 0.5 cm to 1 cm (FWHM). The limiting factor in the maximum rates does not come from the electronics but from the pile-up of light pulses in the NaI. Their decay time is close to 0.6 μ sec, and rates above 3×10^5 /sec are not permitted.

3.2.1 Methods of localizing γ -emitters

The imaging is usually obtained with appropriate collimators. Figure 7 shows the three types of collimators that are suitable:

- the pinhole is useful for small objects at short range and permits magnification (Fig. 7a);
- the parallel-plate collimator has a field equal to the size of the detector (Fig. 7b);
- the diverging hole permits a larger field (Fig. 7c).

The collimators are made of thick metal plates in which holes are bored; the most commonly used collimator has parallel holes.

A typical transmission yield is of the order of 10^{-4} .

The efficiency of a NaI crystal is close to 100% for most of the radionuclides used, and the excellent energy resolution of NaI (about 16% at 140 keV) permits the rejection of scattered degraded radiation, sometimes playing an important role in the improvement of the image quality at the cost of some loss in efficiency.

Positron emitters permit a quite different type of collimation. The positron stopping in live tissue, after a range of a few millimetres at most, emits two γ -rays of 0.511 keV at nearly 180° except for a small deviation from this angle of about 0.5° . By simultaneously detecting and localizing the two γ -rays, it is possible to trace a straight line on which the emission point should be approximately located (Fig. 8). It is clear that if a great number of such straight lines are obtained, the maximum density of the intersecting points between the lines and planes at different depths will reflect the density of the tracer concentration.

The considerable advantage of positron cameras lies in the fact that no collimator is necessary and that depth information (tomography) is obtained.

An efficiency of 40% in the detection of the annihilation quanta results in a 40% yield of coincidences versus single counting rates for a point source without absorber. Taking into account the additional losses due to absorption, scattering, and geometry, the yield is still very favourable when compared to the 10^{-4} geometrical yield of a collimator, with the important advantage of giving depth information.

The transmission imaging with extended or punctual sources, as illustrated by Fig. 9, is directly in competition with the common X-ray absorption methods. It has an interest for some specific organs where low-energy X-rays are required. We will find here an example of the advantageous use of MWPCs.

Transmission imaging can also be used with positron emitters. In this case, the source is embedded in a thin container of 1 mm maximum size (Fig. 10). By requiring the impact points (x_1y_1) and (x_2y_2) to be collinear with the point source (x_0y_0) , the Compton-scattered radiation is eliminated since, as we will see, detectors with 1 mm accuracy are feasible in this range of γ energies using heavy converters in drift spaces. This allows us, in a unique way, to make radiographs with 0.5 MeV X-rays with a radical elimination of the noise from scattered radiation. This may be of great interest in cases where, for instance, a reduction of the contrast between bones and soft tissues is desired.

Typical tracers used in nuclear medicine, and selected from a few dozen examples, are listed below.

<u>Element</u>	<u>Half-life</u>	<u>γ energy</u>	
^{99m}Tc	6 h	140 keV	daughter of ^{99}Mo (67 h)
^{201}Tl	3 d	167 keV	
^{125}I	60 d	35 keV	
^{123}I	13 h	159 keV	
^{81m}Kr	13 sec	190 keV	daughter of ^{81}Rb (4.5 h)
^{133}Xe	5.3 d	81 keV	
^{67}Ga	3 d	92 keV	cyclotron produced
^{24}Na	15 h	1.37 and 2.76 MeV	

The most favourable positron emitters are the following:

^{68}Ga	68 min	511 keV	daughter of ^{68}Ge (287 d)
^{11}C	20 min	511 keV	cyclotron produced
^{13}N	10 min	511 keV	cyclotron produced
^{15}O	2 min	511 keV	cyclotron produced
^{18}F	110 min	511 keV	cyclotron produced

This short list shows that the energy range of interest for the emitted γ 's extends from 30 keV to a few MeV. The lifetime extends from a few seconds to years. This diversity of physical properties often commands a variety of solutions for the imaging.

In the choice of the isotopes, many factors play a major role: the metabolic properties of their compounds; the capability of the γ -rays to escape from the organs where they are fixed; the production mode; the half-life. A privileged role is played by ^{99m}Tc , which is the most commonly used. Its half-life of 6 h

permits easy handling as far as safety is concerned, and rapid elimination from the body is guaranteed by its decay. It is the daughter product of a longer-lived isotope that is easy to obtain. The energy is compatible with an accurate collimation by lead. This is not true, for instance, for 0.5 MeV where the thickness of the septa has to be much greater, of the order of 3 mm.

Scintillation cameras have adequately dealt with the imaging problems of this isotope, and before examining the various attempts to use MWPCs for imaging γ -rays, it is worth listing the essential properties of the scintillation cameras.

3.2.2 Characteristics of scintillation cameras for the imaging of single γ 's

We consider a camera used with a parallel-plate collimator. In a typical catalogue we find the following characteristics:

- NaI crystal (diameter: 38 cm, thickness: 1.3 cm)
- Energy resolution: 16% at 140 keV
- Space resolution: 6 mm (FWHM) for 140 keV γ -rays
- Maximum counting rate: 5×10^4 /sec
- Cost: \$100,000-\$200,000.

The limit in counting rates is sometimes due to electronics, but pile-up problems do limit the maximum rate to about 3×10^5 /sec.

The price of the crystal is \$8,000. Photomultipliers cost about \$100 a piece; the most modern cameras use about 20 PMs, and the most sophisticated use about 90. These numbers are important when making comparisons with MWPCs in domains where they can compete. They show that an important fraction of the cost is not in the detector but in the overhead, and that the choice will be guided by specific characteristics and rarely by the detector cost.

3.2.3 Characteristics of some scintillation cameras for positron imaging

We give below the characteristics of four different types of cameras. The fact that there is a field where MWPCs offer possible solutions requires a close watch to be kept on solutions based on scintillators.

a) A camera with two large-surface NaI crystals. This approach, developed by Muehllehner¹⁴⁾, is based on two Anger cameras in coincidence. Its characteristics are the following:

- $d\Omega = 9\%$ of 2π (diameter: 34 cm, thickness: 2.3 mm)
- Number of γ - γ coincidences/ μ Ci of distributed positron emitter: 200 counts per sec (source without absorber)
- Spatial accuracy: 1.3 cm
- Maximum rate: 8000 counts/sec, of which about 50% are from scattered radiation

- Dead-time: 1 μ sec
- Minimum time for a tomography: 1 minute.

If 5 mCi of ^{68}Ga are injected, 10% stay in the brain tissue. The single count rate reaches the limit of 300,000 counts/sec, of which about half is Compton-scattered 0.511 keV γ -radiation.

b) A camera with two sensitive planes made of a mosaic of NaI crystals¹⁵⁾. The prototype of this camera was developed at the Massachusetts General Hospital and commercialized by the Cyclotron Corporation.

Each sensitive plane consists of 140 NaI crystals of 2 cm diameter and 3.8 cm thickness.

The characteristics are:

- Number of γ - γ coincidences/ μ Ci of positron emitter: 100
- Number of good γ - γ coincidences/ μ Ci in the head: 5
- Spatial accuracy: 1 cm to 1.4 cm (FWHM)
- Maximum rate: 5000 counts/sec
- Minimum time for a tomography: 5 min (including mechanical motion).

c) A camera with a portion of a ring of NaI crystals¹⁶⁾. Such a camera, developed by Phelps¹⁶⁾ and commercialized by Ortec, views only a slice of a head, and a series of successive slices are needed. Its characteristics are:

- $d\Omega = 0.005$ of 2π
- Number of γ - γ coincidences/ μ Ci of positron emitters uniformly distributed in the head: 0.7 coincidences/ μ Ci
- Spatial accuracy: 1.3 cm
- Maximum rate: 10,000 counts/sec
- Time for imaging one 2 cm thick slice of a head: 3 min.

d) A very recent and ambitious ring camera developed at Donner Laboratory¹⁷⁾ should also be mentioned. Figure 11 shows the arrangement of the detector ring, consisting of 280 NaI crystals and photomultipliers. The main characteristics are:

- NaI crystal size: 8 mm $\times$ 30 mm $\times$ 50 mm deep
- Image resolution: 7.5 mm FWHM at ring centre
- Event rate per 200 μ Ci per 1 cm sections of a 20 cm diameter cylinder of water: 5000/sec
- Accidental rate: 1400/sec
- Scattered coincidence rate: 1200/sec.

The authors¹⁷⁾ expect that this camera will allow them to image the myocardial uptake of ^{82}Rb (half-life 1.3 min) and the metabolism of amino acids labelled with

^{11}C (half-life 20 min) with a resolution of 10 mm FWHM and a statistical uncertainty of 20%. It will be used to measure brain blood-flow, and here the limitations in statistics will then give a resolution of 20 mm (FWHM) with 20% uncertainty. They consider an improvement in resolution to 6 mm FWHM and an increase in sensitivity by a factor of 15 by using three rings of bismuth germanate. This would allow the visualization of five sections and permit dynamic studies that are beyond the scope of present foremost cameras.

These four cameras do not exhaust all the approaches to these problems, but their characteristics may help us to appreciate the possible contribution of MWPCs towards solving them.

In most cases these cameras do not fully exploit the energy resolution of NaI crystals; the Compton effect in NaI is too important to keep only the photo-electric peak. They just apply some cut-off between 50 keV and 200 keV to reject low-energy γ -rays scattered by the body under investigation. The great diversity of their properties -- acceptance, maximum counting rates, etc. -- is due to the fact that none of them is ideal. In some of them, those with a large acceptance, the maximum rate will be determined by the detector's dead-time, in others by the dose limit. This dose limit may vary very much with the isotope, its lifetime, or its metabolic properties. It may be that in some particular cases the characteristics of MWPC γ -cameras -- low efficiency, rate limitation by accidentals' rate, much better spatial accuracy -- may find a favourable application. This has still to be proved, and so far most users have preferred scintillation cameras.

The cost of a complete system with NaI scintillators is close to \$500,000 of which only a small part is invested in the detector, and some of the projected developments may well increase this cost, which is similar to that of a modern computer-assisted X-ray scanner.

3.3 Imaging of single γ 's absorbed in the gas of a MWPC

Bearing in mind the properties of NaI scintillation cameras, which are the routine tools in the present state of the art in nuclear medicine, gives us a measuring stick with which to judge the efforts made to replace them with other detectors.

From the low cost of a single large NaI crystal it is clear that the possible lower cost of MWPCs is of little importance. The reliability of a system based on photomultipliers is, so far, greater than the one based on MWPCs. Indeed good examples of reliability exist in large systems of MWPCs. For instance, a group at CERN, working with 70,000 wires of 2 m length, claims no broken wire in two years of operation. However, the design of a push-button system ready to be used

by unqualified personnel is much easier with PMs and is more attractive for commercial firms. In other words, unless MWPCs can, in some cases, do things that PM systems cannot do, and fill a gap in nuclear medicine, there is no chance of them superseding existing systems. In the cases where this *does* occur it is safer, at least in the first stages, if teams of high-energy physicists or technicians familiar with the MWPC technique collaborate in such projects. As we will see, several groups have found applications where they think they can achieve better results with MWPCs than with the present available techniques. In such applications, the better spatial resolution, the better energy resolution for low-energy X-rays, and/or the higher rate capability of MWPCs as compared to NaI crystals have been the determinant parameters. For instance, Bateman and Waters¹⁸⁾ have made a detailed study of the problems connected with X-ray absorptiometry. In the problem they considered -- namely the variations in the mineral content of the forearm -- the variations of the calcium content which are looked into are of the order of 1% per annum. The statistical accuracy is only a minor part of the factors affecting the accuracy: for instance, the variations in the fatty tissue in the marrow may simulate such fluctuations. It is only by taking images at different orientations and fitting images of successive scans that one can accurately evaluate bone-mass variations over long periods of time. A similar development has been done by an MIT group¹⁹⁾. The fact that low-energy γ -rays of the order of 50 keV are most favourable for this problem has favoured the use of MWPCs.

Energy resolution also plays a very important role, as can be seen from the spectra of silver fluorescent X-rays without and with an 11 mm thick Al absorber (Fig. 12). We see that a measurement based only on counting-rate variations will give a completely wrong answer concerning the absorption coefficients. In this respect, high-pressure xenon-filled MWPCs or, even better, high-pressure xenon light-emission chambers (which will be considered later), may be better than NaI crystals. Let us now consider in more detail some properties of MWPCs relevant to these applications.

3.3.1 Efficiency of the gaseous fillings in MWPCs

If the efficiency is determined by the absorption in the gas then clearly high-pressure krypton or xenon fillings are almost a must in such detectors. With a 6 cm thick chamber, filled with Xe-CO₂ at 4 atm, Kaufman et al.²⁰⁾ could obtain efficiencies around 20% for 100 keV photons. Bolon et al.²¹⁾ have built a camera filled with xenon at 11 atm, and with a drift space of 11 cm depth and a diameter of 10 cm. Figure 13 shows the efficiency curve of such detectors in two cases: Kr, 5 cm thickness, 2 atm; and Xe, 15 cm thickness, 10 atm. Also shown is the efficiency of a most common NaI crystal, 1.3 cm thick, of an Anger camera. Large thickness of drift spaces are permissible since collimators impose a parallel

geometry. The pressure is not only determined by the requirement of high efficiencies but also by the requirements of spatial resolution since this controls the range of the secondary electrons and photons.

At high pressures the MWPC seems to be competitive with NaI crystals in the region of 140 keV as far as efficiency is concerned, but serious technological difficulties are associated with the operation of such chambers, and it is doubtful whether it is worth the effort when one compares such a detector with a NaI detector, unless the better space resolution is highly desirable. In some cases, low-energy γ -rays are better adapted to specific studies. At these energies there is a considerable degradation of the NaI characteristics as far as energy resolution or position accuracy is concerned, and interesting developments with MWPCs may be foreseen.

Iodine-125, which emits 27 keV X-rays, has very interesting properties for some specific medical problems. It can be fixed to thyroid tissues and discloses metastases of thyroid tumours. It can also be fixed in pharmaceuticals that will adhere to blood clots, thus revealing these clots, which cause a considerable number of deaths after some surgical interventions. Some studies of the irrigation of the myocardium with ^{131}Cs (32 keV X-rays) also rely on low-energy X-rays where the efficiency of the chambers is adequate. Xenon-133, emitting an 80 keV γ -ray, is used for lung ventilation; this has been imaged by a high-pressure xenon-filled MWPC²²⁾. It was among the first attempts to use a MWPC in medicine, and the instrument is now in routine use in a hospital. Since the spatial accuracy is an essential advantage of MWPCs for low-energy γ imaging, let us examine what are its limiting factors.

3.3.2 The position accuracy in xenon-filled chambers

The spatial accuracy in MWPCs is controlled by several effects: the intrinsic accuracy of the chamber, often dominated by the anode wire spacings; the accuracy of the read-out methods; the range of the secondary electrons (photoelectric and Compton effect in the gas); and the propagation length of secondary X-rays emitted by the interacting gas atom.

Bateman et al.²³⁾ have made a detailed analysis of these factors for xenon-filled MWPCs.

Table 1 shows the results obtained by G. Bolon et al.²¹⁾ at 6.3 atm absolute, with a mixture of 90% xenon and 10% methane, for the three very important isotopes ^{125}I , ^{201}Tl , $^{99\text{m}}\text{Tc}$, with the characteristic values of the factors controlling the accuracies and efficiencies.

The symbols used in the column headings of Table 1 designate the following quantities: E_i , the incident γ and X-ray energy in keV; f_i , the fractional abundance per source disintegration; r_i , the photoelectric absorption, which has a value distance in atm-cm Xe; E_e , the dominant photoelectron energy in keV; r_e , the mean path length for this photoelectron in atm-cm Xe; f_x , the fractional fluorescence yield; E_x , the dominant fluorescence energy in keV; r_x , the photoelectric absorption half-value distance for this fluorescence in atm-cm Xe; f_a , the fractional photoelectron absorption for quanta reaching the detector; and E_a , the dominant observed energy. For the last of these quantities the letters in parentheses designate the process: P for total photoelectric absorption, or E for photoelectrons.

Table 1

Primary interaction properties in xenon

Source	E_i	f_i	r_i	E_e	r_e	f_x	E_x	r_x	f_a	E_a
^{125}I	27	1.05	11	22	0.4	0.10	4	0.3	0.98	27(P)
	30	0.24	13	24	0.5	0.10	4	0.3	0.96	29(P)
^{201}Tl	72	0.78	28	37	1.0	0.85	30	14.7	0.78	42(E)
	83	0.22	42	48	1.5	0.85	30	14.7	0.64	53(E)
$^{99\text{m}}\text{Tc}$	140	0.88	187	105	5.4	0.85	30	14.7	0.21	110(E)

The results are encouraging. They have reached an efficiency of 20% for technetium, with a spatial accuracy twice as good as that of the best scintillation cameras and much higher counting rate capabilities.

The disadvantage is that the photoelectric absorption does not result in a single peak but in a very complex set of peaks because of the escaping fluorescence radiation. Technetium, for instance, produces a broad 110 keV escape peak with a shoulder on the high-energy side and a well-separated 30 keV peak from fluorescence, while ^{125}I presents essentially a single total photoelectric peak.

Construction of such high-pressure cameras with diameters comparable with those of Anger cameras may be so difficult that the cost would be much higher. It would be justified, as a first step, in applications where the high spatial resolution is essential, together with a high counting rate capability.

At Rutherford Laboratory, Horsman et al.²⁴⁾ have used MWPCs for bone-mass measurements by photon absorptiometry methods, illustrated by Fig. 14. They choose

an external punctual source of ^{153}Ga (diameter = 3 mm), which decays with a half-life of 240 days emitting 42 keV europium X-rays. This energy lies above the K-edge of xenon, and the error introduced by the range of the photoelectrons is negligible. They use a total thickness of gas (xenon + isobutane) of 1.6 cm, and most of the K fluorescence X-rays escape.

They have obtained the width and mineral content of a bone in a way that is more satisfactory than with the existing techniques, which usually rely on rectilinear scanning with NaI detectors; these inflict greater radiation doses on the patients and are more subject to errors in determining precisely the site of the measured density.

3.4 MWPCs with solid radiators for γ -conversion

In order to overcome the drawback of low γ -absorption in the gas fillings, some authors have tried various combinations of solid converters, where the electrons are ejected from solids and localized in MWPCs. Two methods have been developed:

- i) A drift space is filled with solid converters providing channels for the production of ionization electrons by the electrons ejected from the converters. An electric field transfers the ionization electrons to a MWPC where they are then localized.
- ii) The cathodes of a MWPC are made of a suitable single converter, and the ejected electrons are localized in the chamber. The efficiency is obtained by stacking a sufficient number of chambers.

The practical differences between the two approaches are considerable. Although approach (i) was developed first, let us now discuss the simpler, second one.

3.4.1 Stacks of chambers with single plane converters

If the cathode is made of converting foil, a stack of chambers can, in principle, yield a good efficiency. The optimum material is dependent on the energy of the γ . The efficiency is a function of the absorption length in the cathode foil and the probability of the fast electron to escape from the foil.

Bateman and Connolly²⁵⁾ have made a detailed analysis of this approach, considering that 20-30 layers are practically feasible. They optimized the detector for 140 keV γ -rays, and since the calculations show that the optimum efficiency is reached for an atomic number of 50 they chose tin as the converting material.

Taking as a practical limit 25 chambers, they concluded that they can obtain 16.5% efficiency with 30 μm thick foils for 140 keV γ -rays.

Excluding xenon as a filling gas because of its high cost, they investigated gases giving the shortest range for the secondary electrons and concluded that 2.2 dimethyl propane (3 mg/cm^3) is almost as good. With a 4 mm cathode-to-cathode gap they should be able to resolve 2 mm bars, which would be very acceptable.

Strangely enough, the spectrum of pulse heights has some correlation to the spatial resolution. Selection of the lower pulse improves the spatial resolution; selecting simply 2/3 of the pulse significantly improves the resolution (a change of about 20% in the contrast). These authors obtain a satisfactory image of a thyroid phantom by selecting 44% of the pulses, but this brings down the poor efficiency by a substantial factor.

The image is good and the lack of energy resolution does not seem catastrophic. However, a more realistic phantom than the thyroid, with a more important diffusing medium as is the case in practical clinical use, would be necessary in order to make a comparison with the Anger camera where the rejection of scattered radiation, made possible by the good energy resolution, is usually considered as a necessity -- the more so since such a detector is considered by these authors as a candidate for whole-body imaging, where the imaging of deep-seated organs is important.

3.4.2 Gamma camera with a dense converter and a drift space

Two examples of the realizations of such converters have been described²⁶⁻³¹⁾.

One was realized by a Berkeley group²⁶⁻²⁸⁾ for the specific aim of building a positron camera for medical purposes. Its characteristics are: a size of $48 \times 48 \text{ cm}^2$ and an accuracy of 6 mm FWHM; an efficiency of somewhere around 8% is claimed for a chamber with a converter on each side.

The second one, realized by Jeavons et al.²⁹⁻³¹⁾, initially for accurately measuring the angle of the two annihilation γ 's, has different characteristics: a size of $10 \times 10 \text{ cm}^2$; an accuracy of 1.5 mm FWHM; an efficiency of 15%. We may consider that these characteristics are a reasonable guess at what can be achieved in the near future and compare the advantages of this type of γ -cameras with those of scintillation cameras.

3.4.3 Applications of MWPCs with solid converters for imaging of single γ 's

In the case of imaging single γ 's with collimators, the use of solid converters in drift spaces offers the advantage of a high spatial resolution and a high counting rate capability, at least one order of magnitude higher than that of an Anger camera. However, in most cases the dose limitation in human beings virtually forbids taking advantage of these properties. The high spatial accuracy requires more radiation per unit area, and this is aggravated by the low efficiency and

the loss in the collimator, which has to be narrower to give a better accuracy. In the case of research where highly radioactive short-lived γ -emitters, sitting at well-defined hot points, have to be localized rapidly, the situation may change.

The maximum activity that can be injected into human beings is 20 mCi for ^{99m}Tc ; this corresponds to a radiation dose of about 150 mrad. However, some people are considering, for instance, the use of ^{191m}Ir , with a half-life of 13 sec. For the same activity the dose is only 0.5% of the one delivered by ^{99m}Tc ; thus one starts with a considerable initial activity saturating an Anger camera. A mosaic of NaI crystals would be the solution, but a cheaper one will probably be that of a dense drift space. A much faster solution will be a stack of chambers with an independent read-out on each gap.

3.4.4 Applications of MWPCs for positron imaging

In the case of positron imaging for medical applications, the disadvantages of MWPCs are difficult to overcome.

In the case of drift spaces, the poor time resolution introduced by the drift time plays a drastic negative role in the coincidence measurements. The rates are then most often limited by the accidentals rate and not by the dose. In this case, the figure of merit is characterized by the factor $k = \epsilon^2/\tau$, where ϵ is the efficiency and τ the time resolution. Sodium-iodide cameras are 100-1000 times better off than high-density drift cameras in this respect. In other words, scintillation cameras are virtually not limited by the resolution time but by the data-rate limit or the dose limit, and this is very annoying since it prevents the full use of the two advantages of the MWPC: i.e. the higher space resolution needs higher counting rates for it to be exploited; the larger sizes that are possible with MWPCs lead to an increased contribution from parts of the body which are not under investigation but which have absorbed a fraction of the radioactive material.

The approach of Bateman et al. would seem justified here, since it permits an important gain in time resolution over drift chambers. Efficiencies of the order of 15% can be reached at 511 keV if one accepts large numbers of successive layers of MWPCs, but the reliability of a system of 50 MWPCs may be a serious problem in a hospital environment.

Table 2, calculated by R. Hubbard at Saclay, gives a rough comparison of the merits of the different approaches.

We have taken as an example for the NaI detector two Anger cameras in coincidence, since they offer a geometrical resemblance to the MWPC position camera.

Table 2

Limiting factors in positron cameras

	Dense drift	Sandwich of MWPCs	Large NaI crystals
ϵ	10%	10%	50%
τ	300 nsec	50 nsec	15 nsec
ϵ^2/τ	$1/3$	2	150
Maximum rate, in counts/sec	1000	6000	10,000
$\frac{\text{Signal}}{\text{Noise}}$	$1/2$	$1/2$	$1/1$
Resolution (FWHM)	5 mm	7 mm	15 mm
Limit	Accidentals	Accidentals	Pile-up
Activity	30 μCi	200 μCi	1 mCi

It is interesting to see the results obtained by the Perez-Mendez group, which has had an operating device at the San Francisco Hospital since several years. Its characteristics are:

- Sensitivity: 25 counts/ μCi
- Sensitive area: $48 \times 48 \text{ cm}^2$
- Angular acceptance: $\pm 45\%$
- Data rate at 10% accidentals: 500 counts/sec
- Spatial resolution: 6 mm FWHM in air
7 mm FWHM in 10 cm of lucite.
- Time resolution: 200 nsec FWHM.

The results presented in Ref. 28 show a brain tumour and a comparison is made with the results of an EMI scan and a scintillation camera image. However, the former results were obtained in about 30 minutes, and the performance of such a camera is so far quite inferior to a NaI system. But it has the great merit of showing which factors have to be gained in order to make it attractive.

The next speaker, A. Jeavons, will certainly analyse the possible lines of progress.

V. Perez-Mendez claims that a lead-glass converter can lead to an increase in the over-all efficiency of his present camera by a factor of 25. I will not discuss

the problem of the efficiency of converters, as this will certainly be dealt with in detail by the next speaker. If it were possible to reduce the measuring time to a few minutes, this would bring the camera to the same level as many NaI systems. Whether it can compete with these systems in some specific cases and make full use of its intrinsic qualities of spatial accuracy has still to be proved. In the discussion I have by-passed the problem of reabsorption of Compton-scattered γ -rays which introduce an error and a background dependent on the average density of the detectors. In this respect, stacks of chambers are inferior to high-density counters in drift spaces.

4. MWPCs FOR THE IMAGING OF NEUTRONS

Slow neutrons offer considerable advantages as compared to X-rays in some domains of crystalline structure studies. Whenever wavelengths between 1 and 20 Å are needed, the X-rays are strongly absorbed, while the matter is usually transparent to slow neutrons. Many biological structures belong to this field.

Two-dimensional positron-sensitive detectors offer the same advantages for neutrons as for X-rays, i.e. higher rates of data collection, possibility to study systems in rapid evolution. The detection medium is specific to the absorption properties of slow neutrons; the two filling gases, ^3He and BF_3 , are practical for the detection of slow neutrons because of the reaction

Alberi et al.³²⁾ report the construction of a chamber filled with a mixture of 6 atm ^3He , 4 atm Ar, and 0.5 atm CO_2 , with the following characteristics:

- Neutron efficiency: 70% for 2.7 Å neutrons
- Position resolution: $\sim 3 \text{ mm}$
- Integral linearity: 1%
- Event resolution time: 1 μsec
- Event processing time: 3 μsec
- γ -ray sensitivity (^{60}Co): 2.7×10^{-6}
- Sensitive area: $18 \times 18 \text{ cm}^2$.

We find here a typical case where the choice of a read-out method is guided by the physical constraints. In high-pressure chambers the signal rise-time is slow. Delay lines do not offer sufficient delay to give adequate spatial resolution, while charge division is perfectly suited to the collection time. The range of the protons resulting from the neutron absorption is 2.3 mm and the final resolution, 3 mm, is very close to this value.

Another kind of chamber³³⁾, built at the Laue-Langevin Institute at Grenoble, relies on different solutions. It is filled with 1.2 atm of BF_3 ; this moderate pressure permits a much larger surface than with the previous solution.

The characteristics of the chamber are:

- Neutron efficiency: 50% at $6 \text{ \AA}$
- Position resolution: 10 mm
- Event resolution time: 1 μsec
- Event processing time: 0.5 μsec
- γ -ray sensitivity (^{60}Co): 10^{-6}
- Sensitive area: $64 \times 64 \text{ cm}^2$.

The resolution is limited by the range of the recoil products, and corresponds to the detection cell size.

The read-out is made by digital methods, i.e. bands of 10 mm width printed on glass. Drift chambers with converting materials could also be used; with converters of gadolinium, boron, beryllium or lithium, drift chambers sensitive to thermal and epithermal neutrons can be produced^{34,35)}.

The large cross-sections for slow neutron absorption of these materials, producing highly ionizing recoil products, require mechanical solutions that are radically different from those applying to γ -rays and which constitute a challenging domain for experimentalists. The fact that large surfaces can easily be built, and the considerable technical advantage of operating with gases at atmospheric pressure, are certainly attractive features of a drift chamber approach with proper converting materials.

5. DEVELOPMENTS OF NEW DETECTORS

Some detectors are now under active development and the outcome of their investigation may well have an impact on possible applications of MWPCs. Liquid xenon wire chambers would be quite promising if all practical problems of gas amplification in liquid noble gases were mastered. Although one may think that the requirements of cryogenic installations are unattractive for an hospital environment, it should not be forgotten that the aim is not restricted to building imaging instruments that can compete with available commercial objects, but sometimes to provide a research establishment with a unique powerful instrument that is useful only for one line of research. The possibility of extracting electrons from liquid and solid noble gases has been demonstrated by Dolgoshein et al.³⁶⁾, and this opens up the possibility of combining the properties of solid xenon, as far as efficiency or energy resolution is concerned, with the properties of gaseous wire chambers as far as space resolution is concerned.

In pure gaseous detectors the development of scintillation proportional counters³⁷⁾ can also lead to solutions that are preferable to those of MWPCs in the domain where they have precisely an interest, i.e. for imaging of X-rays between 30 keV and 100 keV. They have the advantage of a better energy resolution, and the localization can be obtained by techniques similar to those used for scintillation cameras. Figures 15 to 17 illustrate some recent results obtained at the laboratory of the linear accelerator of Orsay by Dr. N'Guyen^{*)}.

6. CONCLUSIONS

The properties of proportional chambers are interesting for some problems of radiation localization encountered in various fields in medicine or biology.

For the imaging of low-energy X-rays in the region of 5 to 20 keV commonly used in the study of protein structures by diffraction methods, they offer one of the most attractive solutions amongst available techniques.

For the imaging of the lowest part of the spectrum of γ -rays used in nuclear medicine (27 keV to about 80 keV) they present a great interest when compared to scintillation cameras: a greater spatial accuracy and a higher rate capability when compared to single large NaI cameras, combined with a good energy resolution.

For higher γ energies, in the domain most common in nuclear medicine they cannot so far compete with scintillation cameras as tools for general use.

However, the efforts made to overcome the low efficiency may well succeed in finding specific applications where use can be made of their intrinsic qualities of spatial accuracy and high rate capability.

In the field of low-energy neutron imaging, important for structure studies in molecular biology, MWPCs offer interesting solutions when compared to available present techniques.

^{*)} Private communication.

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Figure captions

- Fig. 1 : Proton radiography. Protons of 1 GeV scattered by a human head stored in Formalin. Display of two adjacent slices, 1 cm thick, in the direction perpendicular to the beam, which enters through the top of the head. Eight levels of grey distant by one standard deviation. Radiation dose: 0.1 rad.
- Fig. 2 : Mean free path of absorption in some gases. Atmospheric pressure.
- Fig. 3 : Range of secondary electrons produced by the capture of X-rays in some gases at 1 atm. Several electrons may follow a capture process. The accuracy of a chamber is dominated by the electrons with the longest range.
- Fig. 4 : Degradation of spatial accuracy for inclined trajectories of X-rays. With a flat chamber, small gaps and high pressures are required to keep a good spatial accuracy.
- Fig. 5 : Spherical drift chambers. With a radialelectric field centred on source position, the accuracy is kept independent of inclination for any thickness of drift space (from Refs. 9 and 10).
- Fig. 6 : Diffraction pattern from a rat-tail collagen. X-rays of 10 keV (from Ref. 13).
- Fig. 7 : Imaging methods for γ -rays. Radioactive materials are injected into the body:
- a) Pinhole camera.
 - b) Parallel-hole collimator. Typical parameters: For 140 keV γ -rays: 1 mm thick septa, 1 mm diameter holes, 3 cm depth;
For 500 keV γ -rays: 3 mm thick septa, 3 mm diameter holes.
 - c) Diverging holes.
- Fig. 8 : Positron camera. Positron emitters are injected into the body. The range of β^+ is of a few millimetres; they annihilate into two γ 's of 0.511 keV almost collinear. Two detectors measure the positions of the γ 's.
- Fig. 9 : Imaging with external γ -sources. The source is either extended and collimated or punctual.
- Fig. 10 : Imaging with external positron sources. The source is embedded in a small, heavy-metal container. By requiring the alignment of the three points S γ_1 γ_2 , one eliminates scattered radiation.

- Fig. 11 : Detector ring for positron imaging. 280 NaI crystals. Images a slice of the body. Accuracy: 7.5 mm FWHM at ring centre (from Ref. 17).
- Fig. 12 : Effects of energy resolution in transmission measurements.
- a) Source of silver K X-rays detected with a germanium crystal.
 - b) Same source after traversal of 1 cm of Al.
- (From Ref. 18).
- Fig. 13 : Efficiency of some gaseous detectors (from Ref. 4). The fractional energy loss of Compton scattering at 90° is shown to permit evaluation of the energy resolution necessary for discrimination against Compton scattering.
- Fig. 14 : Bone-mass measurement by absorptiometry. 42 keV X-rays emitted by ^{153}Gd ; 1.6 cm thick Xe-filled MWPC (from Ref. 23).
- Fig. 15 : An imaging scintillation proportional chamber (N'Guyen, LAL Orsay).
- Fig. 16 : Resolution of a scintillation proportional chamber (N'Guyen, LAL Orsay). 60 keV γ -rays from an americium source scattered at 130° from a KI solution in water. The $K\alpha$ and $K\beta$ fluorescence radiation from iodine and the escape lines from the 50 keV Compton peak are visible.
- Fig. 17 : Images obtained with a scintillation proportional chamber and ^{131}I :
- a) Thyroid phantom
 - b) Rabbit thyroid, 2 minutes exposure.
- (N'Guyen, LAL Orsay.)

a)

b)

Fig. 1

Fig. 2

Fig. 3

Maximum error $d \tan \theta$

Fig. 4

Fig. 5

Fig. 6

Fig. 7

Fig. 8

Fig. 9

Fig. 10

Fig. 11

a)

b)

Fig. 12

E_γ , keV

Fig. 13

a)

b)

Fig. 14

G1-G2 Drift space 500 Vcm^{-1}
 G2-G3 Light space 3500 Vcm^{-1}
 Gas Xenon at 1 atm.

Fig. 15

Fig. 16

a)

b)

Fig. 17