EUROPEAN ORGANIZATION FOR NUCLEAR RESEARCH

CERN-EP/80-53
21 April 1980

A PROPORTIONAL CHAMBER POSITRON CAMERA FOR MEDICAL IMAGING

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ABSTRACT

A proportional chamber positron camera based on the principle of the high-density drift space has been constructed for medical imaging. Each of the two chambers has a 1 cm thick converter, 20×20 cm in size, with a detection efficiency of 8.5% at 0.5 MeV and a time resolution of 100 ns. With a new mode of operation, using neon gas, less than 20 ns has now been achieved. Spatial resolution for a 1 mm line source in plastic is 2 mm FWHM for ^{22}Na and 3.5 mm FWHM for ^{68}Ga . A source of 300 μCi activity, surrounded by a 20 cm diameter plastic scattering bolus, provides a count rate of 4000 coincidences per second, of which 50% are random. This will improve to 20% with the 20 ns time gate. A three-dimensional image comprising 16 slices may be obtained in 20 minutes. Bone and organ imaging is presented.

Paper presented at the
Wire Chamber Conference, Vienna, Austria
27-29 February 1980

1. INTRODUCTION

The advent of X-ray computerized tomography has greatly stimulated the desire to realize detailed three-dimensional images in the field of nuclear medicine. Positron-active tracers offer a unique advantage in this respect because the photon emission from positron annihilation consists of two rays back-to-back. By imaging these two gamma-rays in coincidence, directional as well as positional information is obtained. Thus emission computerized tomography may be performed with a ring of detectors around the patient. Various cameras of this nature have been constructed: well known is the ECAT camera¹⁾ which is available commercially.

With X-ray scanning, a maximum of information from a given number of photons is obtained because the X-rays are collimated into a beam. However, for gamma-ray emission this is not the case: the photons are emitted isotropically, and a camera with a large solid angle acceptance, to image as many photons as possible, is highly desirable. If the same quality of information is to be realized as with X-rays, a spatial resolution of 1-2 mm is required. To obtain pictures quickly and perform dynamic studies, high count rates must be handled, and this implies a short resolving time to minimize accidental coincidences. Now, proportional chambers are used extensively in particle physics research for just these reasons:

- i) good spatial resolution (~ 1 mm);
- ii) fast resolving time (~ 20 ns);
- iii) high rate capability ($\sim 10^6$ particles s^{-1} cm^{-2});
- iv) large area for low cost.

In making a positron camera, the difficulty is to retain these properties whilst efficiently converting the 0.5 MeV photons, resulting from positron annihilation, into an ionizing form suitable for detection by the proportional chamber. At CERN this problem has been solved by the addition of a high-density drift space to the proportional chamber. The drift space consists of a sandwich of lead and fibre-glass sheets perforated with a large number of holes close together. Photons interact in the lead bars and produce fast electrons which can escape to an adjacent hole. The free electrons, resulting from gas ionization in a hole, may be

extracted by an electric field and detected by the proportional chamber. At CERN and the Geneva Hospital we are investigating the application of such high-density proportional chambers to positron imaging, following their successful application to the measurement of the angular correlation of positron annihilation radiation for solid-state physics research. A positron camera consisting of two chambers is now in operation for tomographic imaging. In this paper the construction and performance of this latest camera is reported. The reader is referred to other publications for information on the basic principle and properties of the high-density proportional chamber²⁾, chamber read-out techniques^{3,4)}, the application to solid-state physics research^{5,6)}, and preliminary medical imaging results^{7,8)}. This earlier work has also been reviewed^{9,10)}.

2. CAMERA DESIGN AND PERFORMANCE

2.1 Construction

The camera consists of two 20 × 20 cm chambers 30 cm apart. The chambers follow conventional design, the important parameters being: anode wire diameter 20 μ m and spacing 2 mm, and anode-cathode spacing 3 mm.

Signals from the orthogonal cathode planes are read out by the coded centre-of-cluster technique¹¹⁾. Cathode wires are divided into groups, and into channels within the groups. Each group has an independent output, and corresponding channels may be connected together (see Fig. 1). The number of outputs is reduced by the factor $\sqrt{N/2}$, where N is the number of wires coded. This reduction in the number of outputs simplifies the subsequent decoding. The outputs may be strobed into a read-only memory, precoded with the coordinate values for all valid group and channel patterns. All invalid patterns resulting from spurious signals, e.g. noise, cross-talk, too-wide distributions, double events in the chamber, are set to return an out-of-range coordinate. Thus in one memory access, the signals are decoded, checked for validity, and the coordinate value returned.

Each chamber is equipped with a 1 cm thick converter, which consists of 48 0.1 mm thick sheets of a lead-antimony alloy, and 48 0.1 mm thick sheets of fibre-glass reinforced epoxy-resin. A converter is made by first gluing the sheets

together into four rigid 2.5 mm thick stacks, and then drilling out a set of 0.8 mm diameter holes on a 1 mm pitch, with a numerically controlled printed circuit drilling machine -- a precise construction results (Fig. 2) from a cheap, simple technique. Using a four-head machine, a $20 \times 20 \times 1$ cm converter with 50,000 holes can be drilled in about 16 hours. A chemical treatment after drilling removes any short circuits between the lead plates.

The top (i.e. remote from the proportional chamber) of the converter is covered with a sheet of lead glued to a flat glass plate to electrically "close" the converter and prevent electrons drifting backwards to the chamber window, which is at earth potential. A small gap is left between this plate and the converter to allow a gas flow.

2.2 Detection efficiency

The detection efficiency of a chamber depends not only on the converter design but on the chamber operation too. The best detection efficiency is obtained using a heavy hydrocarbon gas such as isobutane, which has a high specific ionization for the initial fast electron and gives a stable, high-gain operation to the chamber. The detection efficiency is then 8.5% for 0.5 MeV photons. As only 30% of the photons interact in the converter, the converter yield is 28%. It is now clear that the good stacking regularity obtained with the gluing and drilling converter-construction technique is important in minimizing electron losses during their drift through the holes. In this respect it is also advantageous to use high precision (1%) resistors to connect the lead sheets and to slightly focus the electric field in the direction of the drift. This may be achieved by increasing each resistor value by 1-2% over the preceding one. It is also important to minimize gas leaks, as oxygen can diffuse into the chamber and cause electron attachment. We reduce leaks to a level of $1 \text{ cm}^3 \text{ min}^{-1}$. An improvement in detection efficiency is planned for the future by the use of multiple converters.

2.3 Time resolution

The time resolution of a chamber is dominated by the drift time of the electrons through the converter. Clearly a thinner converter will give a better time

resolution, but only at the cost of a correspondingly reduced detection efficiency. A different type of camera has recently been announced^{12,13}) where wire planes are interleaved with every sheet of the converter. Time resolution is about 20 ns, but spatial resolution is ~ 6 mm as there are no holes in the converter to trap the fast electrons.

Drift time may be reduced by optimizing the gas mixture used in the chamber and the voltage applied to the converter. In Fig. 3 information is presented on electron drift velocities in various gases and mixtures¹⁴⁻¹⁸). It is immediately apparent that for moderate drift fields, methane is the fastest gas. Argon mixed with some polyatomic quenching gas, e.g. carbon dioxide, is also fast, but pure methane is always faster. Unfortunately, the specific ionization of methane is comparatively low and the maximum detection efficiency cannot be attained.

To minimize the time resolution without sacrificing detection efficiency, a mixture of 70% methane/30% 2-2 dimethylpropane is used in the camera. This latter gas, an isomer of pentane C_5H_{12} , is similar to isobutane, but about 20% faster (and ten times more expensive!). Figure 4 shows the time resolution for the two chambers in coincidence. The triangular shape arises from the correlation of two square distributions, one from each converter. The shoulders result from coincidences between a converter event in one chamber, and a direct, prompt event in the other. The full width of the triangular spectrum is 250 ns, and using a 100 ns time gate on each chamber which accepts all events in a time slot of 200 ns (i.e. ± 100 ns) only 4% of the events in the triangular spectrum are missed.

It is evident from Fig. 3 that a shorter resolving time can be obtained only by increasing the reduced electric field. A reduction in pressure would reduce the initial ionization signal and lower the detection efficiency, as reported above for methane. Thus the only option would seem to be an increase in the electric field strength. The use of carbon dioxide in this way has already been reported⁹): a drift velocity of 10^7 cm s⁻¹, equalling that of methane, was obtained; but detection efficiency fell slightly, possibly due to electron losses resulting from increased lateral diffusion at high drift fields. In an effort

to do better, we have investigated the use of neon, which also shows a high drift velocity at high field strength, see Fig. 3. A small amount of isobutane was added to stabilize the chamber operation. We discovered [as has recently been reported elsewhere¹⁹⁾] that neon with a small quantity of quenching gas ionizes very readily, and electron avalanching occurs at quite modest fields ($\sim 5 \text{ V cm}^{-1} \text{ Torr}^{-1}$). Thus a new mode of chamber operation is possible: *electron amplification in the holes of the converter* with further amplification in the proportional chamber. There are the benefits of overcoming electron losses in the converter, and having chamber operation with a safe, non-polymerizing gas mixture. But it is the effect on the resolving time that is dramatic, as the time lag in the converter is determined by electron avalanche propagation rather than electron drift. In Fig. 5, a time spectrum is shown for a 5 mm thick converter in coincidence with a fast scintillator. The FWHM is 13 ns, and a significant part of this will be due to the proportional chamber and its electronics since the pulse rise-time is $\sim 50 \text{ ns}$. In Fig. 6 the effect of varying the concentration of the quenching gas is shown. The optimum resolving time is obtained with 5% of isobutane.

A possible mechanism for the avalanche is the Penning effect. In this effect, electron collisions excite gas atoms to metastable states which cannot decay by photon emission; instead, the atoms ionize the secondary gas by collision. In neon the effect is very strong: the addition of only 0.1% of argon drastically changes the amount of ionization²⁰⁾. But the Penning effect does not seem to offer an explanation for the short avalanche propagation time. Electron drift velocity will be no higher than 10^7 cm s^{-1} (Fig. 3), implying a resolving time of 50 ns. Propagation of the avalanche due to ionization of the isobutane by photons resulting from inelastic collisions is probably involved. This mechanism has been invoked to explain similar phenomena observed in argon-acetone mixtures²¹⁾.

Work continues on this development, which makes the time resolution of this type of camera as good as that of conventional sodium iodide scintillators. More results will be published later²²⁾.

2.4 Spatial resolution

The over-all spatial resolution of the camera is determined by several factors: the chamber resolution, the parallax error due to the converter thickness, the departure from collinearity of the two gamma-rays, and the distance travelled by the positron before annihilation. We have measured this over-all resolution for a 1 mm line source located centrally between the chambers, and find that, as expected, the positron range error dominates. For ^{22}Na (maximum positron energy 0.5 MeV) in plastic we obtain 2 mm FWHM, a result slightly better than that of earlier work⁸⁾, probably due to the thinner converter. For ^{68}Ga (1.9 MeV) the resolution degrades to 3.5 mm. Replacing the plastic tube with tungsten eliminates the positron range error, and the result for ^{22}Na is nearly reproduced (see Fig. 7).

By taking the Fourier transform of the line spread functions of Fig. 7, the spatial frequency spectrum or modulation transfer function of the camera is obtained. This is compared to other cameras^{23,24)} in Fig. 8. It may be seen that much better contrast is achieved at all spatial frequencies.

2.5 Energy resolution

Although there is no correspondence between photon energy and chamber-pulse height, the camera does have an important intrinsic sensitivity to photon energy. As the photon energy falls, the interaction probability rises as the photoelectric effect dominates. This would increase the detection efficiency, but it is more than offset by the fact that the lower energy photoelectrons have lower probabilities of escaping from the lead. Down to about 200 keV detection efficiency drops slowly, after which it falls rapidly. Thus the camera is insensitive to a significant amount of the Compton-scattered radiation that exists in a clinical imaging situation²⁵⁾ and causes problems for a sodium iodide scintillation camera²⁶⁾.

2.6 Counting rate

The dominant source of noise in the camera is the occurrence of accidental coincidences; hence the importance of reducing the resolving time, as reported above. The current situation for the camera is given in Fig. 9: a rapid build-up of accidental coincidences with increasing source intensity is evident. With a 300 μCi source of ^{68}Ga inside a 20 cm diameter plastic scattering bolus, a coinci-

dence rate of 4000 s^{-1} is obtained, of which 50% are accidental ones. With the new mode of operation and a time gate of 20 ns instead of 100 ns, this accidental fraction will drop to less than 20%.

3. POSITRON IMAGING WITH ^{68}Ga

Objects are imaged by positioning them centrally between the two chambers. Operating the camera with the conditions given in the previous section implies that three-dimensional images containing more than one million true, unscattered coincidences may be obtained in 20 minutes. Limited angle tomographic reconstruction by Fourier techniques²⁷⁻²⁹⁾ of such images provides 1 cm thick slices parallel to the chambers.

The source of the positron-emitting radionuclide was a 1 mCi Ge/Ga generator eluted with EDTA. After elution, approximately one hour was needed to prepare a bone-labelling tracer ^{68}Ga -DTPMP. *In vivo* images were taken for a series of young rabbits. The ^{68}Ga -DTPMP was injected into an ear vein; about $1\frac{1}{2}$ hours afterwards, the rabbit was anaesthetized and imaging started. Typically, less than 100 μCi of activity was within the field-of-view of the camera at the start of the data acquisition. The data rate was therefore limited by source strength, and a time of about 45 minutes was required to accumulate some 2×10^6 events.

In Fig. 10, a single longitudinal slice corresponding to the midplane of a small rabbit is shown for two cases: a) the blurred image resulting from back-projection, and b) the same image after Fourier deconvolution²⁸⁾. The slice contains about 2×10^5 counts. As for all slices, the slice thickness is 1 cm and the pixel dimensions $2 \times 2 \text{ mm}$. For the grey scale assignment, black corresponds to the minimum count density (usually zero) and white to the maximum. The skeleton of the rabbit is almost entirely within the camera field-of-view, with the head tucked round, eliminating the characteristic curvature of the cervical vertebral column. The spine is clearly visible, as are individual vertebrae as far as the lower lumbar region. Higher activity, consistent with increased bone metabolism in young rabbits, can be seen at some metaphysis-epiphysis areas of the front and rear legs corresponding to shoulder, elbow, and knee. The skullcap is outside the field-of-view, but a portion of the base, including maxilla and mandibula, can be seen.

A similar slice of a larger rabbit is shown in Fig. 11. The skull, cervical and thoracic vertebral column, shoulder joint and front leg are all clearly seen. The skull outline is also visible, as is the strong isotope absorption in the shoulder joint and individual vertebrae. The high spot of activity seen at the top left originates from residual activity at the site of the intravenous injection in the left ear. The slice contains approximately 1.5×10^5 counts, and they were recorded at a higher accidental coincidence rate than for the previous case. Thus Fig. 11 exhibits more noise. However, some of the spots are due to weak activity in the ribs.

Finally, a third rabbit was imaged with the skull, shoulder joint, and front leg in the field-of-view (Fig. 12). Residual activity at the injection site is again seen at the top. The skull shows details of the cavity containing the brain and eyes, and the maxilla and mandibula bones. Activity from the shoulder joint appears just below the portion of the cervical vertebral column.

The tomographic capability of the camera is demonstrated in Fig. 13, which shows a sequence of *in vitro* images of an isolated human heart, with the cavities filled with a ^{68}Ga -EDTA solution. Sixteen sequential transverse slices are shown as viewed from above the heart. The first slice (top left) and last slice (bottom right) are situated physically below and above the heart, respectively, and show only noise. The intermediate slices show activity in the right and left ventricles and right and left atria, with the interventricular septum clearly visible in the early slices. In the first slice of the third row the cardiac border is formed by the coronary sinus, the lower limit of the left atrium and the right atrium, with the entrant portion of the inferior vena cava on the posterior and right lateral side. The right and left anterior border is formed by the right and left ventricles.

4. CONCLUSION

A large-area proportional chamber positron camera has been demonstrated for *in vivo* imaging with a high spatial resolution of 2-3 mm FWHM. Small animals can be imaged in about 30 minutes. For human studies further improvements, particularly in sensitivity, are desirable. Development is continuing with the following goals:

- 1) larger chambers (30-40 cm);
- 2) four chambers, rather than two, to provide high spatial resolution in all three dimensions;
- 3) 20-30% detection efficiency per chamber;
- 4) ten nanoseconds coincidence time resolution.

Such a camera should provide a significant contribution to the field of nuclear medicine.

REFERENCES

- 1) M.E. Phelps, E.J. Hoffman, S.C. Huang and D.E. Kuhl, J. Nucl. Med. 19 (1978) 635.
- 2) A.P. Jeavons, G. Charpak and R.J. Stubbs, Nucl. Instrum. Methods 124 (1975) 491.
- 3) C. Parkman, Z. Hajduk, A.P. Jeavons, N. Ford and B. Lindberg, Proc. 2nd Ispra Nuclear Electronics Symposium, Stresa, 1975 (Commission European Communities Report EUR 5370e, Luxembourg, 1975), p. 139.
- 4) A.P. Jeavons, N.L. Ford, B. Lindberg, C. Parkman and Z. Hajduk, IEEE Trans. Nucl. Sci. NS-23 (1976) 259.
- 5) A.A. Manuel, O. Fischer, M. Peter and A.P. Jeavons, Nucl. Instrum. Methods 156 (1978) 67.
- 6) A.A. Manuel, S. Samoilov, O. Fischer, M. Peter and A.P. Jeavons, Helv. Phys. Acta 52 (1979) 255.
- 7) A.P. Jeavons and C. Cate, IEEE Trans. Nucl. Sci. NS-23 (1976) 640.
- 8) A.P. Jeavons, D.W. Townsend, N.L. Ford, K. Kull, A. Manuel, O. Fischer and M. Peter, IEEE Trans. Nucl. Sci. NS-25 (1978) 164.
- 9) A.P. Jeavons, Nucl. Instrum. Methods 156 (1978) 41.
- 10) A.P. Jeavons, Proc. 5th Int. Conf. on Positron Annihilation, Lake Yamanaka, Japan, 1979 (Japan Inst. Metals, Tokyo, 1979), p. 355.
- 11) A.P. Jeavons, Coded centre-of-cluster read-out for wire chambers, to be submitted to Nucl. Instrum. Methods.
- 12) J.E. Bateman and J.F. Connolly, Nucl. Instrum. Methods 156 (1978) 27.
- 13) J.E. Bateman, J.F. Connolly, R. Stephenson and A.C. Flesher, The development of the Rutherford Laboratory MWPC positron camera, presented at this conference.
- 14) B. Jean-Marie, V. Lepeltier and D. L'Hote, Nucl. Instrum. Methods 159 (1979) 213.
- 15) S.C. Brown, Basic data of plasma physics, 1966 (The MIT Press, Cambridge, Mass., 1967).

- 16) W.N. English and G.C. Hanna, Can. J. Phys. (1953) 768.
- 17) O.C. Allkofer, Spark Chambers (Verlag Karl Thiemig KG, Munich, 1969).
- 18) E.W. McDaniel, Collision phenomena in ionized gases (Wiley, New York, 1964).
- 19) S.H. Williams, D.W.G.S. Leith, M. Poppe and T. Ypsilantis, SLAC-PUB-2412
(1979), and IEEE Trans. Nucl. Sci. NS-27 (1980) 91.
- 20) M.J. Druyvesteyn and F.M. Penning, Rev. Mod. Phys. 12 (1940) 88.
- 21) A. Breskin, G. Charpak, S. Majewski, G. Melchart, G. Petersen and F. Sauli,
Nucl. Instrum. Methods 161 (1979) 19.
- 22) A.P. Jeavons and P. Frey, The high-density avalanche chamber, to be sub-
mitted to Nucl. Instrum. Methods.
- 23) E.J. Hoffman, S.C. Huang and M.E. Phelps, J. Comput. Assist. Tomogr. 3
(1979) 299.
- 24) G. Bennett, BNL, private communication.
- 25) Z.H. Cho, M.B. Cohen, M. Singh, L. Eriksson, J. Chan, N. MacDonald and
L. Spolter, Proc. Int. Symposium on Medical Radionuclide Imaging,
Los Angeles, 1976 (IAEA, Vienna, 1977), vol. 2, p. 269.
- 26) G. Muehllehner, M.P. Buchin and J.H. Dudek, IEEE Trans. Nucl. Sci. NS-23
(1976) 528.
- 27) G. Chu and K.C. Tam, Phys. Med. Biol. 22 (1977) 245.
- 28) B. Schorr and D. Townsend, Submitted to J. Comput. Assist. Tomogr.
- 29) D. Townsend, B. Schorr and A.P. Jeavons, IEEE Trans. Nucl. Sci. NS-27
(1980) 463.

Figure captions

- Fig. 1 : The principle of coded centre-of-cluster read-out. Thirty-two wires are encoded into four groups of four channels.
- Fig. 2 : An example of drilling a 2.5 mm thick lead/glass-fibre sandwich. The hole diameter is 0.8 mm.
- Fig. 3 : Electron drift velocities in various gases: CH_4 ^{14,15}), C_3H_8 ¹⁴), 70% CH_4 /30% C_5H_{12} (this work), 95% Ar/5% CO_2 ¹⁶), He, Ne, Ar, Kr, Xe, CO_2 ^{15,17,18}).
- Fig. 4 : The coincidence time spectrum of the camera. Horizontal scale: 5 ns per channel.
- Fig. 5 : The coincidence time spectrum of a 5 mm thick converter with a fast scintillator. Horizontal scale: 0.5 ns per channel. FWHM: 13 ns. Gas mixture: 95% Ne/5% C_4H_{10} .
- Fig. 6 : The dependence of the coincidence time of Fig. 5 on the concentration of isobutane. The different points for each gas composition reflect different combinations of converter and chamber voltages.
- Fig. 7 : The spatial resolution of the camera for various line sources.
- Fig. 8 : The modulation transfer function (Fourier transform) of the data in Fig. 7 compared to other cameras^{23,24}).
- Fig. 9 : The dependence of the accidental coincidence rate on the total counting rate in a simulated clinical imaging situation.
- Fig. 10 : A longitudinal slice through a small rabbit: a) the blurred back-projected image and b) after Fourier deconvolution²⁸).
 2×10^5 counts. 2×2 mm bins.
- Fig. 11 : As Fig. 10 but with the head erect.
- Fig. 12 : Details of the head and front portion of a rabbit.

Fig. 13 : Sixteen transverse slices of an isolated human heart. $\sim 10^5$ counts in each of the central slices. Imaging time equivalent to one minute per slice.

MWPC 32 CHANNELS

CODED-CLUSTER READOUT

Fig. 1

Fig. 2

Fig. 3

Fig. 4

Fig. 5

Fig. 6

CAMERA SPATIAL RESOLUTION

Fig. 7

Fig. 8

Fig. 9

Fig. 10

Fig. 11

Fig. 12

Fig. 13