CHAPTER 7

ELECTRONICS RELATED TO NUCLEAR MEDICINE IMAGING DEVICES

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7.1. INTRODUCTION

Nuclear medicine imaging is generally based on the detection of X rays and γ rays emitted by radionuclides injected into a patient. In the previous chapter, the methods used to detect these photons were described, based most commonly on a scintillation counter although there are imaging devices that use either gas filled ionization detectors or semiconductors.

Whatever device is used, nuclear medicine images are produced from a very limited number of photons, due mainly to the level of radioactivity that can be safely injected into a patient. Hence, nuclear medicine images are usually made from many orders of magnitude fewer photons than X ray computed tomography (CT) images, for example. However, as the information produced is essentially functional in nature compared to the anatomical detail of CT, the apparently poorer image quality is overcome by the nature of the information produced.

The low levels of photons detected in nuclear medicine means that photon counting can be performed. Here each photon is detected and analysed individually, which is especially valuable, for example, in enabling scattered photons to be rejected. This is in contrast to X ray imaging where images are produced by integrating the flux entering the detectors. Photon counting, however, places a heavy burden on the electronics used for nuclear medicine imaging in terms of electronic noise and stability.

This chapter will discuss how the signals produced in the primary photon detection process can be converted into pulses providing spatial, energy and timing information, and how this information is used to produce both qualitative and quantitative images.

7.2. PRIMARY RADIATION DETECTION PROCESSES

As described in Chapter 6, the methods used for the detection of X ray and γ ray photons fall into three categories, namely the scintillation counter, gas filled detectors and semiconductors. Each of these techniques provides several detector types and requires different electronics to produce and utilize the signals.

7.2.1. Scintillation counters

Figure 7.1 shows a block diagram of a scintillation counter using a phosphor and photomultiplier combination, together with the basic electronics required to produce analogue and digital signals used to create an image. Table 6.3 shows that the phosphors used in nuclear medicine can produce 1500–67 000 optical photons per megaelectronvolt of energy deposited in the crystal and the light emission time can vary from less than 1 ns up to ~1 μs. Additionally, the amplification of the optical signal by a photomultiplier can vary by an order of magnitude or more depending on the photocathode quantum efficiency and the number of dynodes. From this, it can be seen that the pulses produced by the scintillation counter can vary substantially in both shape and amplitude, and that the electronic devices used to manipulate these signals must be flexible enough to account for these variations.

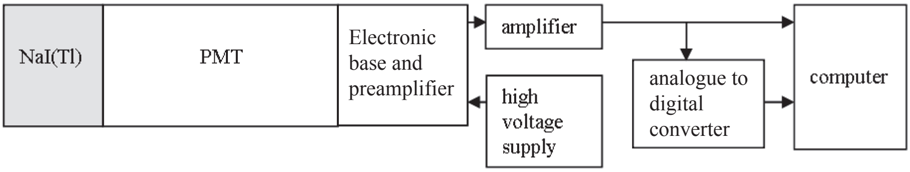

FIG. 7.1. Block diagram of a scintillation counter and associated electronics.

If the signal from the photomultiplier tube (PMT) anode is small, a preamplifier is needed prior to full amplification. This form of amplifier is usually incorporated into the PMT electronic base to minimize the noise generated prior to preamplification. Similar arguments apply to the use of solid state based light sensors such as photodiodes when coupled to phosphors.

Both PMTs and photodiodes require voltage supplies to produce signals — in the case of a PMT, this voltage supply can be 1–3 kV as each successive dynode typically requires 100–200 V to produce sufficient amplification of the electron signal. For a photodiode, the voltage required to totally deplete the device is usually a few tens of volts for a simple photodiode and more for an avalanche photodiode (APD).

7.2.2. Gas filled detection systems

Gas filled imaging systems convert the energy deposited by a γ ray photon directly into ion pairs. It takes 25–35 eV to produce a single ion pair, so the primary signal from ^{99m}Tc will be 4000–5000 electrons. This signal will be amplified in the gas detector using a high voltage (a few kilovolts) to produce an electron avalanche of typically 10^6–10^7 in a multiwire proportional chamber (MWPC) (Fig. 7.2). Typical dimensions of these devices for medical imaging are between 30 and 100 cm laterally by 10–20 cm in the direction of the γ ray.

These signals will clearly also need amplification if they are to be used as analogue and digital output pulses for image formation.

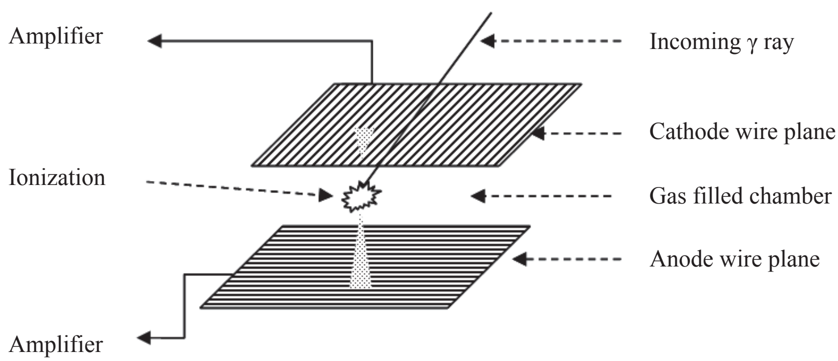

FIG. 7.2. Schematic of a two-plane multiwire proportional chamber detecting a γ ray.

7.2.3. Semiconductor detectors

A simple diode can be thought of as a solid state ionization chamber where the region between the p–n junction acts as a reservoir of electron–hole (e–h) pairs. Incoming radiation produces e–h pairs in the diode, the number of which is proportional to the energy deposited in the diode, and an array of diodes can function as a radiation imaging device. The energy needed to produce an e–h pair is ~3–5 eV (see Table 6.2) and so the size of the pulses produced varies less than for a scintillation counter. However, the signals will still require some form of amplification to produce useful analogue or digital information.

7.3. IMAGING DETECTORS

Having briefly discussed the production of signals by the three major ionizing radiation detection processes, it is necessary to understand how these methods are used to produce images in nuclear medicine. The two main imaging devices used are the gamma camera and the positron camera. For completeness, autoradiography imaging of tissue samples containing radiotracers is also described.

Generally, both gamma camera and positron camera systems use scintillation counters as the primary radiation detector because the stopping power for X rays and γ rays is good in the high density scintillating crystals used. However, there have been some examples of cameras using MWPCs and semiconductors, and a brief description is provided here.

7.3.1. The gamma camera

Invented by Hal Anger, the gamma camera is usually based on the use of a single large area (e.g. 50 cm × 40 cm of NaI(Tl)) phosphor coupled to up to a hundred PMTs. The camera (Fig. 7.3) can detect γ rays emitted by a radiotracer distributed in the body. The lead collimator placed in front of the scintillation counter selects the direction of the γ rays entering the device and allows an image of the biodistribution of the tracer to be made.

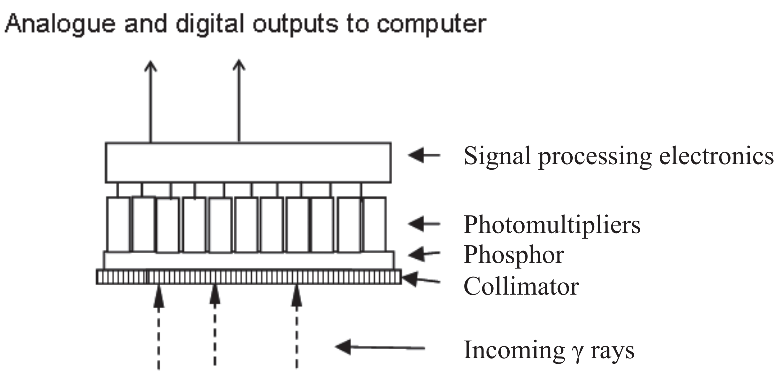

FIG. 7.3. Schematic of a gamma camera for detecting single photons.

The PMTs produce a signal that is proportional to the light generated in the crystal. Positional information can be obtained by comparing the size of the signals from different PMTs, whereas the energy information is related to the sum of the PMT signals. The accuracy of both the energy and the positional information depends on the stability of the signal production and also on the electronics used. The amplitude of analogue signals from the PMTs depends on both PMT and electronic stability, and the noise generated in the signal amplification process. The noise is kept as low as possible by amplification close to each PMT. The amplified/summed signal from the photomultiplier can be converted to a digital pulse train using an analogue to digital converter (ADC), where the number of pulses is proportional to the pulse height or charge in the signal. This signal is used to select events in which most of the energy of the γ ray is detected by the camera, a useful tool to reduce the effect on image production of γ rays scattered in the body prior to detection. It is important at this stage to include only those signals from PMTs that provide information above the intrinsic noise level of the electronics — this is done using some form of signal thresholding, such as a comparator.

Traditionally, the individual PMT pulses are digitized close to the PMTs and these signals are analysed using capacitor or resistor circuits to determine the positional information that is then sent to the computer system to form the image. The centroid of the energy/pulse height information provides a position that is closely related to the point at which the γ ray enters the crystal. Recent improvements for calculating this position based on the digital outputs from the PMTs uses nearest neighbour calculations based on a stored reference map of positional information. These methods provide more accurate estimates of the incident radiation entry point but are more demanding computationally.

Thus, the accuracy of the image production process is very much determined by the initial signal sizes and the subsequent amplification and digitization.

7.3.2. The positron camera

The positron camera is used to simultaneously detect the two annihilation photons produced by positron emitting tracers distributed in the body. The detectors are usually made of many thousands of small scintillating crystals coupled to up to a thousand PMTs. This means that there are many more amplifiers which may have to function at higher count rates than those used in a gamma camera. The detection of these two γ rays requires the addition of coincidence circuits used to select the pulses from a single annihilation event. Figure 7.4 illustrates the format of a positron camera based on multiple scintillating counters in which the signals can be read out to form a 3-D image.

The main difference in the electronics between the positron camera and the gamma camera is the large number of PMT channels involved and the count rates achieved — in both cases, factors of 10–20 are not unusual. In addition, the pulses from a positron camera must be carefully shaped to allow accurate timing information to be made in coincidence circuits to ensure that both annihilation photons from a single annihilation event are detected. Time jitter in the pulses will affect this process, allowing random photons from multiple nucleic decays to be included in the data acquisition. In addition, the recent introduction of so-called 'time of flight' cameras requires very accurate (sub-nanosecond) timing to be made between the two annihilation photons.

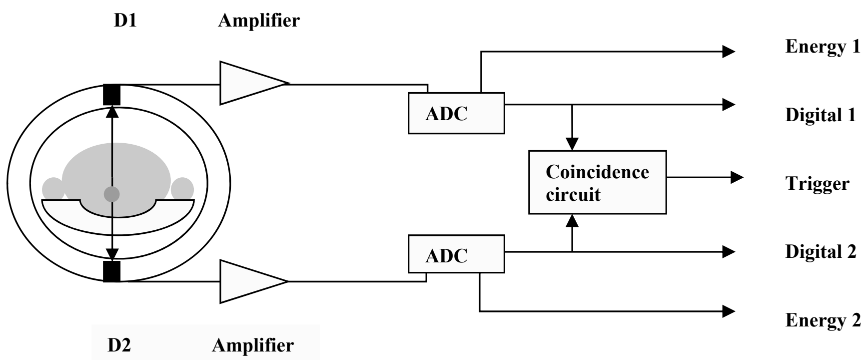

FIG. 7.4. Typical signal processing in a phosphor/photomultiplier tube based positron camera — pulses from two opposing detectors in the camera array are amplified, digitized, checked for coincidence and then used to provide positional and energy information for the event detected.

7.3.3. Multiwire proportional chamber based X ray and γ ray imagers

As shown in Fig. 7.2, an X or γ photon generates an ionization signal in the gas that is detected by the anode and cathode wire planes. The high voltages across the wire planes cause electron avalanches close to the nearest wires and these signals can be detected and amplified at either end of the wire.

In the PETRRA positron camera (Fig. 7.5), the initial γ ray detection is performed using blocks of barium fluoride. The vacuum ultraviolet produced in the crystal photo-ionizes the gas producing electrons that are subsequently amplified in the gas by a series of wire planes. The MWPC positional information is read out using delay lines coupled to the cathode wires. Signals are induced in the delay lines and detected at either end using amplifiers that produce signals with low time jitter. These signals are passed to constant fraction discriminators (CFDs) to produce fast timing signals and then to time digitizers. The time difference between the arrival of the signals at the two ends of the delay line is measured by the time to digital converters (TDCs) and provides the positional information — the accuracy of this information depends on the intrinsic properties of the delay lines and the spread of the signal at the wire planes, and in this system is ~4 mm. Pulses produced after the gas amplification region are used to provide the fast coincidence trigger to read the data into the computer — a timing resolution of ~2–3 ns is readily achievable.

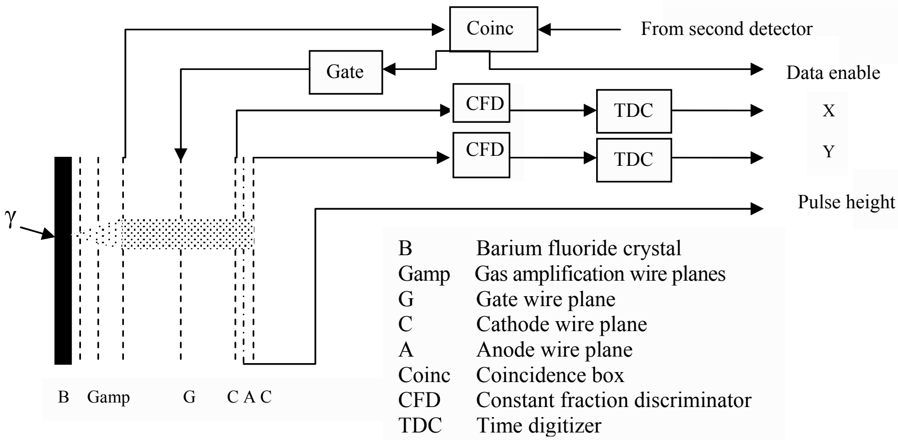

FIG. 7.5. Schematic of the pulse production and readout system for the PETRRA positron camera — wire planes are shown as dotted lines.

A further process to minimize the recording of single γ ray events in this system is the use of an electronic 'gate' in which the transport of ionization signals is controlled by an additional wire plane. This allows passage of the ionization to the anode/cathode part of the chamber only if two MWPC detectors have been triggered in fast coincidence. Anode signals can be used to measure the pulse height produced by each detected γ ray. Overall, the electronics of this camera has to manage count rates of several megahertz.

Similar detectors have been developed for imaging low energy γ rays, animal imaging and tissue autoradiography.

7.3.4. Semiconductor imagers

There have been several attempts to make nuclear medicine imaging devices using semiconductors as a γ ray camera. The need for a high *Z* material means that presently only germanium and cadmium zinc telluride (CZT) have potential as the primary γ ray detector. Germanium (in the form of GeLi) has been used as a 2-D strip detector where the signals from amplifiers at the end resistor chains could be used to determine the position of any interaction in the sensor. However, the modest stopping power of the material coupled with the need for a cryostat to reduce the intrinsic noise of the detector made this design impractical.

More practical systems based on room temperature operation of CZT have been developed by GE (the Alcyone system) and Spectrum Dynamics (the DSPECT system). In the case of the latter system designed specifically for cardiac imaging, ~1000 individual small CZT crystals are coupled to a tungsten collimator providing an intrinsic spatial resolution of 3.5–4.2 mm full width at half maximum and a sensitivity of approximately eight times that of a scintillator based camera — most of the increases in sensitivity are due to the collimator design.

Silicon photodiodes have been used as an alternative to PMTs for both gamma camera and positron camera designs. Here, APDs have been coupled to phosphors and because of their small size, a truly digital camera design is possible. In practice, due to the cost of APDs, only small systems have been developed. The recent development of silicon photomultipliers promises further improvements in nuclear medicine imaging.

7.3.5. The autoradiography imager

Autoradiography is based on the use of radioactive labels to determine the microscopic distribution of pharmaceuticals in tissues excised from humans or animals. A major use in humans is to detect areas of malignancy or tissue malfunction. In animals, the method is used to track the uptake of drugs, for instance. The pharmaceuticals are usually labelled with long lived radiotracers that have a short range β emission or low energy X ray or γ ray emission. Typical examples of tracers used are ^{3}H, ^{14}C, ^{32}P, ^{33}P and ^{125}I. Autoradiography imagers are required to detect the emissions with high efficiency as the levels of uptake in tissue samples are often very low. The gold standard for tissue radiography is film emulsion which produces a high resolution (μm) image of tissues, although these detectors have low efficiency for detecting the radiation involved. Images can take days to weeks to produce and this can be a severe limitation if diagnostic information is desired. Digital autoradiography systems based on the use of thin phosphors, gas filled detectors and silicon wafers can be 50–100 times more efficient although the spatial resolution is limited to typically a few tens of micrometres.

A phosphor based imager may use a very thin (50–100 μm) material such as GADOX or CsI(Tl) coupled to a high resolution sensor, such as a microchannel plate, a charge coupled device or a complementary metal oxide semiconductor APD (Fig. 7.6).

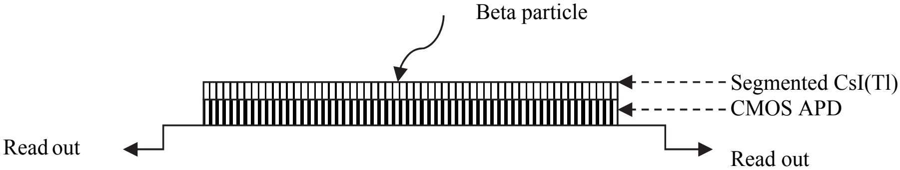

FIG. 7.6. Beta particle detection in an autoradiography system based on a segmented CsI(Tl) phosphor coupled to a complementary metal oxide semiconductor avalanche photodiode (CMOS APD).

The limitations of these devices are mostly the pixel size of the sensor and the noise in the sensor. Amplifiers with low noise are required and room temperature operation desirable. Such a device can have a resolution of <50 μm.

MWPC based autoradiography imagers have been built in which the sample is placed in intimate contact with the chamber gas — such a device will have a resolution of a few hundred micrometres. More recently, direct detection of β particles and X rays has been performed using charge coupled devices and complementary metal oxide semiconductor APDs. The advantage of such devices is that the spatial resolution can be improved further (down to a few micrometres) and ^{3}H can be imaged.

7.4. SIGNAL AMPLIFICATION

As discussed above, the primary signals from the radiation detectors are generally small and need to be amplified without the injection of high levels of noise into the signal readout system. A preamplifier is needed prior to the main amplification process if the signals from the detector are very small, for example, when a PMT has insufficient dynodes to provide a large output pulse. Preamplifiers are usually mounted immediately next to or as part of the output stage of the detector to minimize the noise produced prior to full amplification. The main amplifier can then be used to maximize and shape the signal (via current and/or voltage gain) without over-amplifying noise.

7.4.1. Typical amplifier

The output current from a PMT is directly proportional to the amount of light received from the phosphor. Although the PMT amplifies the electron signal produced at the photocathode by a large factor, the current produced at the anode is still very small. Amplifiers for PMTs are specially designed to transform this current into voltage which can be directly input into an analogue to digital converter or a comparitor. In order to achieve the optimum signal to noise ratio, the output current pulse is integrated in a capacitor, the resulting voltage forming the output signal. The capacitor is normally arranged in the feedback circuit of a wide bandwidth voltage amplifier chosen to have high input impedance and an extremely small input current. As data rates can be high (tens of kiloherz to megaherz), the operational frequency range of the amplifier must be able to cope with these rates. Ideally, the PMT anode is connected directly to the charge amplifier input, with a high value resistor providing a DC return path. Capacitative coupling can be a problem at low frequencies where the signal may be degraded. The charge amplifier integrates the current from the PMT, producing an output voltage pulse. Figure 7.7 illustrates a typical charge amplifier where the output voltage V_{out} is given by:

$$V_{out} = \frac{-1}{C} \int I(t)\, dt = \frac{-Q}{C} \tag{7.1}$$

The configuration has negative feedback that increases the effective input capacitance by a factor equal to the gain of the amplifier. This ensures that almost all of the current flows into the amplifier even though the PMT and wiring can have significant capacitance. The feedback also reduces the output impedance, so that the amplifier acts as a voltage source.

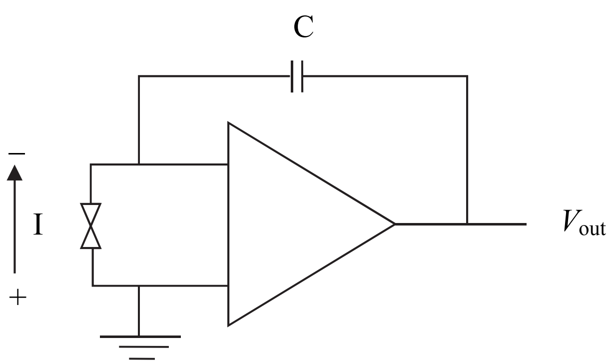

FIG. 7.7. A directly coupled charge amplifier producing a voltage output by integrating the current produced in the photomultiplier tube.

The shape of the output pulse is important for the measurement of both analogue and digital information, and is defined by the output stage of the amplifier (Fig. 7.8). The amplified signal is first passed through a CR (high pass) filter which improves the signal to noise ratio by attenuating the low frequencies, which contain a lot of noise and very little signal. The decay time of the pulse is also shortened by this filter.

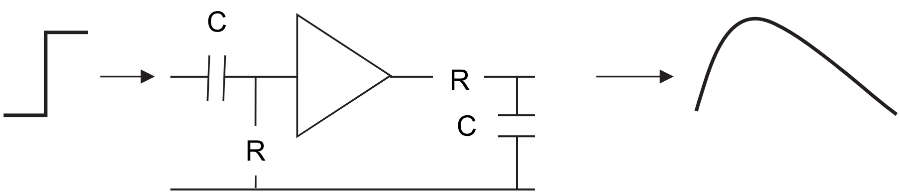

FIG 7.8. A CR–RC pulse shaping circuit.

Before the output of the amplifier, the pulse passes through an RC (low pass) filter which improves the signal to noise ratio by attenuating high frequencies, which contain excessive noise. The pulse rise time is lengthened by this filter. The combined effect produces a unipolar output pulse and with suitably chosen values, has an optimal signal to noise ratio.

7.4.2. Properties of amplifiers

The most important properties of an amplifier are gain, bandwidth, linearity, dynamic range, slew rate, rise time, ringing, overshoot, stability and noise:

— The gain of an amplifier is defined as the log ratio of the output power/voltage P_{out} to the input power/voltage P_{in} and is usually measured in decibels:

$$\text{gain [dB]} = 10\log(P_{out}/P_{in}) \tag{7.2}$$

Gain in charge amplifiers is often expressed in millivolts output per picocoulomb input charge.

— The bandwidth of an amplifier is defined as the range of frequencies that the amplifier operates and is often determined by frequencies at which the power output drops to half its normal value (the –3 dB point). This is an important feature of an amplifier attached to a high count rate detector as required in positron emission tomography (PET) imaging, for example.

— Amplifier linearity is limited when the gain of the amplifier is increased to saturation point, resulting in output pulse distortion. Clearly, this is important if the dynamic range of the pulses produced by the detector is large. Dynamic range is defined as the ratio of the smallest and largest useful output signals, with the former limited by the noise in the system and the latter by amplifier distortion.

- Rise time is often defined as the time taken for the output pulse to increase from 10–90% of its maximum and is a measure of the speed or frequency response of the amplifier.
- Slew rate is the maximum rate of change of the shape of the output signal for the whole range of input signals, usually expressed in volts per microsecond. This is very important if timing information is needed from the detector as a poor slew rate will distort the bigger signals, making them unsuitable for fast timing, as in PET. For PET applications, amplifier rise times of the order of a few nanoseconds are needed, with no shape distortion resulting from slew rate even on the biggest pulses.
- Ringing is a problem when an amplifier produces a pulse that either oscillates before reaching its maximum value or where the tail oscillates before reaching the baseline. This can be a serious problem if timing information is required or if the oscillations produce multiple triggers of the output electronics downstream of the amplifier.
- Stability is clearly an important parameter for an amplifier if the output signals are to be used for either analogue or digital purposes. It is essential that the amplifier output does not vary significantly for a given input signal as the processes used to determine positional, energy and timing information rely on the output for a given input being constant both in offset, amplitude and shape. Factors that affect stability are numerous but prime examples are variations in temperature, supply voltage and count rate as well as long term drift.
- Noise is a major impediment to the production of images using any of the devices discussed above. Examples include thermal noise caused by the thermal movement of charge carriers in resistors, shot noise caused by a random variation in the number of charge carriers and flicker or 1/*f* noise caused by the trapping or collisions of charge carriers in the structure of the silicon used in the electronics. These sources combine to produce a variation in the output signal of the combined detector/electronics system that can affect the quality of images produced. The root mean square noise of a system is defined as the square root of the absolute value of the sum of the squares of the noise variances.

For a system using PMTs, the dominant noise component is that associated with the number of photoelectrons produced at the photocathode as this is amplified by the gain of the PMT dynode chain and subsequent electronics. For a gas filled detector, the equivalent is the number of primary electrons produced at the first stage of the ionization process and for a silicon detector the important parameter is the initial number of e–h pairs produced.

7.5. SIGNAL PROCESSING

Once an amplified signal has been produced, it is then used to generate both analogue and digital information about the detected event. The analogue signal will relate to the energy deposited in the detector and is used, for example, to minimize the number of scattered γ rays accepted into the image production process. The digital signal is used to produce spatial and timing information.

7.5.1. Analogue signal utilization

The analogue information is generated by sending the pulse from the amplifier into a single or multichannel pulse height analyser. In a gamma camera, several 'energy windows' are available, whereby the pulse height or charge is compared with preset values that correspond to the known energies of the γ rays being detected. In the simplest case for imaging a single energy γ ray emission, two thresholds can be set to reject pulses that are above or below these values (Fig. 7.9).

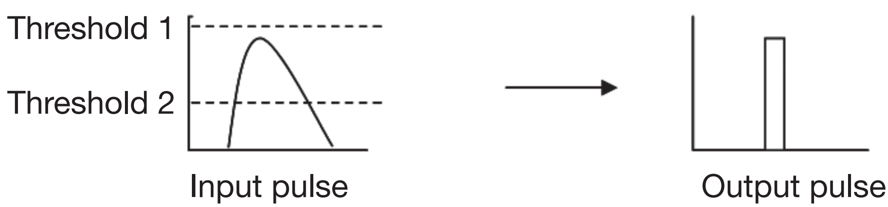

FIG. 7.9. A single channel pulse height analyser — an output pulse is produced when the input pulse is between the two thresholds. This system also functions as a single channel analogue to digital converter.

When a radiotracer that emits several different energy γ rays is being used, multiple thresholds can sort the information into several channels or images.

7.5.2. Signal digitization

Analogue signals are converted into digital signals that are subsequently used to provide spatial and temporal information about each detected event.

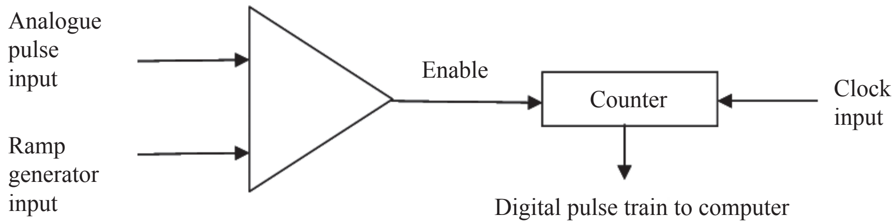

FIG. 7.10. Ramp-based single slope converter system for digitizing analogue pulses.

This is done using an ADC. The simplest method of digitizing an analogue signal is by using a single slope converter (Fig. 7.10). For this, a ramp signal is generated and at the same time a clock producing digital output pulses is started. When the ramp signal exceeds the input pulse, the clock is stopped and the

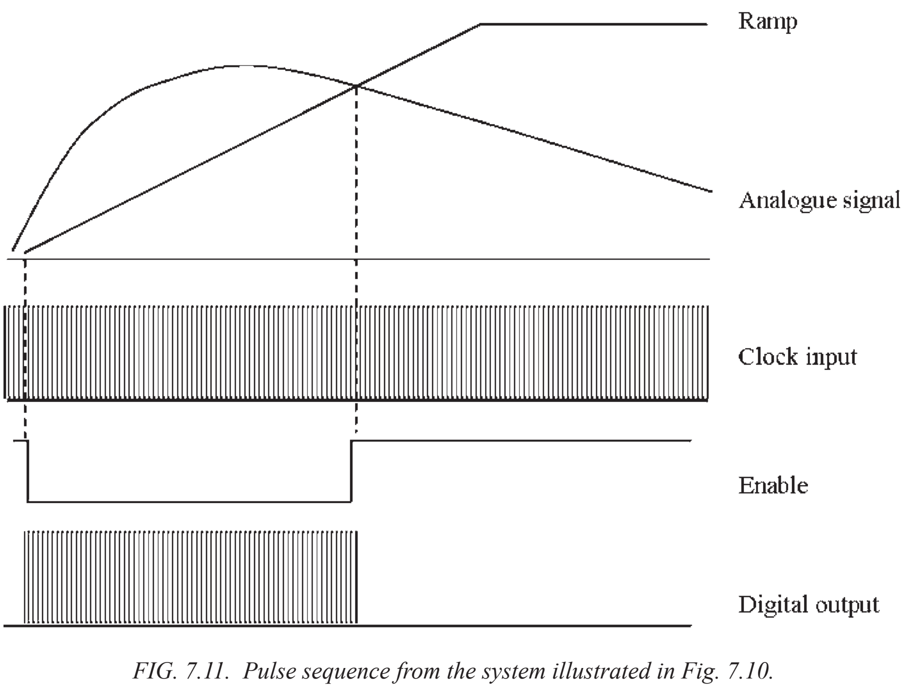

FIG. 7.11. Pulse sequence from the system illustrated in Fig. 7.10.

number of pulses generated corresponds to the amplitude of the signal. The faster the clock, the higher the accuracy of the digitization achieved. This is a relatively simple and low cost solution but is slow as the time taken to digitize the pulse is 2^N clock cycles. The pulse sequence producing the digital output is shown in Fig. 7.11. An important feature is that the analogue pulse shape must be constant to allow accurate conversion. It is clearly possible to have more than one ramp signal to provide several digitization regions if greater or lesser accuracy is needed in any region.

A faster method of analogue to digital conversion is possible by using a flash ADC. This is done using a large number of comparators (see Fig. 7.12), each having a different reference level. The output from each is the input into a logic box that produces the multiple bits of the digital signal. If an N bit output is needed, then $2^N - 1$ comparators are needed. The method is fast as conversion takes a single clock cycle but the system is complex and expensive and consumes a lot of power. Typically, between 8 and 12 bits may be needed for nuclear medicine imaging.

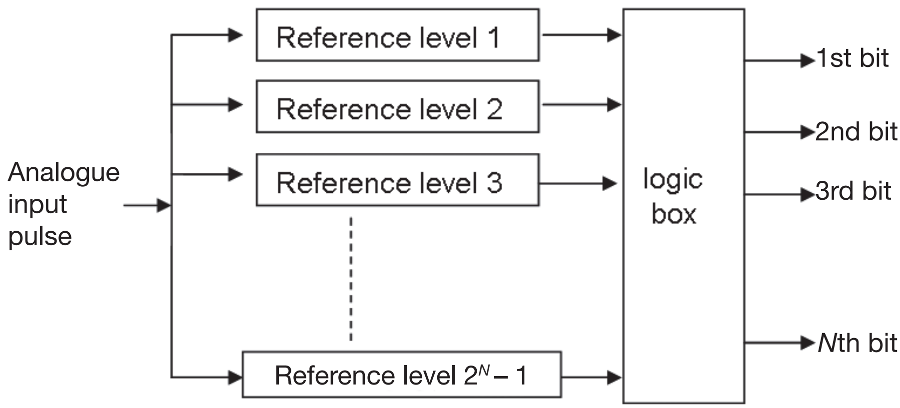

FIG. 7.12. Schematic of a FLASH analogue to digital converter producing N bits of digital data.

7.5.3. Production and use of timing information

In PET systems, the timing of events is very important as only pairs of annihilation photons from the same radioactive decay contribute positively to the image. As the single count rates in a PET scanner may be very high, fast timing is required for coincidence imaging and time of flight measurement. Coincidence timing systems can be based on the timing taken from the front edge of two pulses, from a ‘zero crossing’ point of the differentiated pulses or by using a constant fraction method. The main problem with using a simple front edge trigger is that the variation in pulse height of the analogue signals produces a large variation in timing. Figure 7.13 shows how the two pulses from detectors in a PET system are used to generate a coincidence with CFDs.

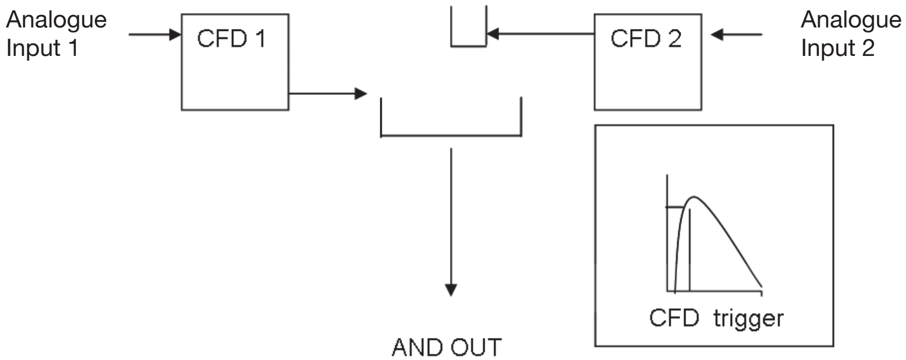

FIG. 7.13. The use of constant fraction discriminators (CFDs) to generate a fast coincidence output — the insert shows how the trigger point is set by a constant fraction of the pulse height.

The trigger points for the timing occur at a constant fraction of the shaped analogue signal, so that the timing is not affected by the different signal pulse heights. In this example, CFD1 generates a gate with a width set to more than twice the measured timing resolution of the detectors. If the pulse from CFD2 falls within this gate, a coincidence (AND) output is generated; otherwise, the event is rejected.

An alternative method of determining the timing from a pulse is to use the zero crossing technique (Fig. 7.14). In this method, the pulse is differentiated to produce a bipolar pulse — the timing is taken from the point where the pulse crosses a reference line that is usually tied to ground — hence, the 'zero crossing'. Again it is important that the pulse shapes are carefully controlled to minimize jitter in the timing information.

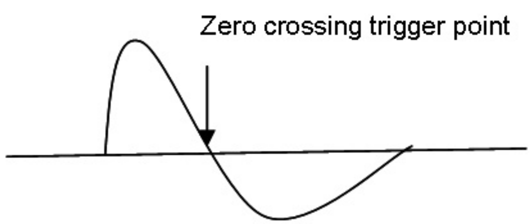

FIG. 7.14. A timing signal generated from the zero crossing point of a differentiated signal.

If the timing information is to be stored, then the two pulses from the CFDs can be input into a TDC. In this case, the first pulse starts and the second one stops a clock — the number of pulses generated is proportional to the time difference between the pulses. If a fast clock is used, excellent timing information is available for use in time of flight calculations, for example.

7.6. OTHER ELECTRONICS REQUIRED BY IMAGING SYSTEMS

7.6.1. Power supplies

Low voltage supplies are used to provide the power input for semiconductor systems where a few tens of volts are sufficient. In some cases, batteries may provide enough power but the need to maintain a constant current and voltage makes this a modest solution. Usually, a low voltage supply converts mains AC power, typically 240 V (or 110 V), into DC voltages of, for example, ±15 V and ±5 V to provide the line voltages for transistors and diodes. This is done by combining a transformer, which reduces the voltage, and a rectifier, typically a diode which allows only one half of the AC signal to pass — this is half-wave rectification (Fig. 7.15).

Full-wave rectification is achieved by using a diode bridge that allows both halves of the AC signal to be used, with one half being inverted. The oscillations are removed using a filter, usually capacitors. The smoothest DC output is provided by using a three phase AC input. The output is usually passed through a voltage regulator to stabilize the voltage and remove the last traces of ripple.

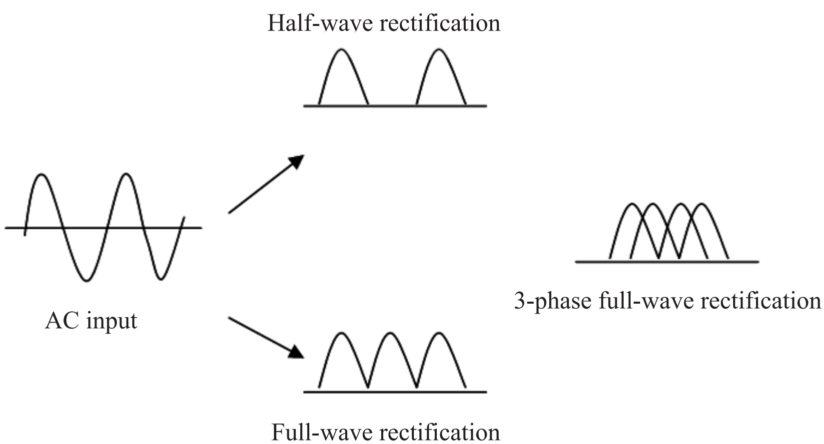

FIG. 7.15. Conversion of AC into DC using a transformer and rectifier system.

PMTs and MWPCs require power supplies that can provide voltages up to several kilovolts. For example, each pair of dynodes in a PMT usually has at least

100 V between them and even more may be used between the photocathode and the first dynode to maximize the early gain of the PMT. These power supplies usually have an oscillator and step up transformer operating at high frequency to provide the drive plus a voltage multiplier consisting of a stack of diodes and capacitors.

7.6.2. Uninterruptible power supplies

This form of support is needed for an imaging system to overcome loss of power during periods of mains supply interruption. In this case, the output power comes from the storage battery via some form of inverter. While the mains power is available, it charges the battery as well as providing power to the imaging device. If the mains supply is interrupted, the battery continues to provide support to ensure that the imaging system can continue to be used. The size of the battery support system depends directly on how long backup is needed or how long, for example, it takes the operator to save data and shut down the system. As in most imaging environments, the mains is replaced by a generator supply. The period of support is often short but usually several hours of supply is available from an uninterruptible power supply.

7.6.3. Oscilloscopes

In order to optimize the use of pulse generating equipment, an oscilloscope is essential. This type of device allows the pulses from the detectors to be displayed at various stages of generation prior to their use in image production. For example, the pulse sequences illustrated above can be displayed on an oscilloscope and this allows the equipment to be adjusted to provide the optimum analogue and digital pulse sequence, shape and size.

An oscilloscope allows the pulses to be displayed on a 2-D display, usually with the vertical axis representing voltage (pulse height) and the horizontal axis time. In addition to the amplitude of the pulses, the oscilloscope display can be used to analyse the frequency of the signals being studied and also to detect any pulse distortion such as oscillation or saturation. In an advanced form, the oscilloscope can function as a spectrum analyser over a wide range of pulse frequencies.

The original oscilloscopes were based on a cathode ray tube to display the pulses but more modern systems use liquid crystal displays connected to ADCs and other signal processing electronics. To the user, the oscilloscope will present as a box with a display screen, input connectors and various controls. The input from equipment can be done either directly using connecting cables/ sockets or through probes, often into a high impedance (e.g. 1 MΩ) or, for high frequency signals, 50 Ω. The trace on the oscilloscope screen is adjusted by various controls. The timebase control can adjust the horizontal display between, for example, 10 ns up to seconds and the pulse height control from millivolts to volts. Other controls include a beam finder, spot brightness and focus, graticule control (to provide a visual measurement grid), pulse polarity and trigger level controls, horizontal and vertical extent and position controls, selection of trigger source (particularly useful for pulse coincidence display) and sweep controls to provide single, multiple and delayed sweeps, for example. An example of an oscilloscope that can be used for examining pulses from imaging equipment is shown in Fig. 7.16.

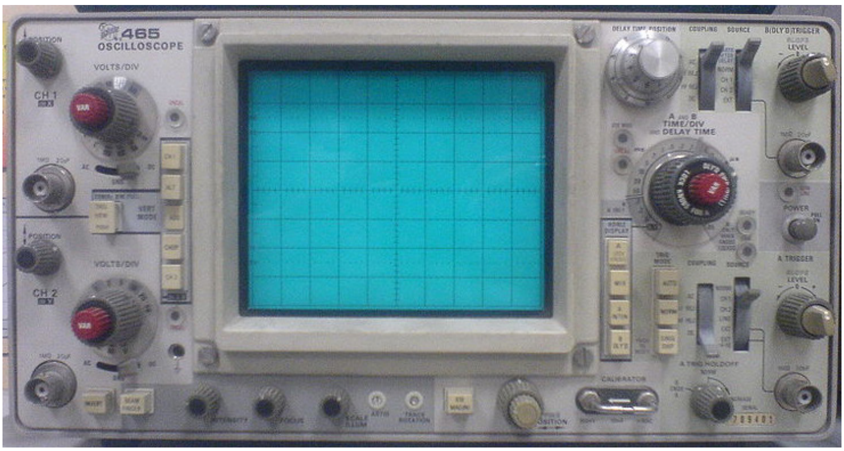

FIG. 7.16. The front panel of the Tektronix 465 oscilloscope.

More recently, it has been possible to install oscilloscope software onto a computer to provide a low cost solution for pulse display.

7.7. SUMMARY

The information provided above gives a general overview of the equipment and electronics used in nuclear medicine imaging. In some cases, manufacturers have developed special multichannel electronics readout systems tailored to the detectors. These systems include the individual electronics elements described above in a compact design that increases speed and accuracy. Such systems are usually specific to the device involved.

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