INVITED COMMENTARY

Merging the Instrumentation Evolution

After reading the interesting article by Wong et al. (*1*) in this issue of *The Journal of Nuclear Medicine* on the state-of-the-art hardware developments in nuclear medicine imaging instrumentation, it is hard to keep one's editorial comments away from historic references. The first nuclear medicine imaging device reported in the literature was the rectilinear scanner (*2*) and was used for imaging ^{131}I's single photons. That same year, time coincidence circuitry was used, allowing two opposing detectors to localize annihilation radiation detected from positron emitters (*3*). One can speculate (I believe correctly) that the front-ends (NaI crystal, photomultipliers [PMTs], preamplifiers, amplifiers, and energy windowing components) of these two devices were quite identical. Yet, since then, PET rings and single-photon planar cameras have been separated stereotypically into two species of instruments used in our profession. When and how did that evolve, if the very first two devices, aside from the coincidence circuitry itself, contained electronics that were virtually identical?

Entertain for a moment the oversimplified statement: PET rings and nuclear medicine Anger cameras evolved into their separate characteristic geometries and specialized electronics because of pulse pileup effects. I believe there is some interesting truth and historic perspective to this. The earliest PET instruments relied on two opposing NaI-PMT probes that electronically determined the line along which two back-to-back annihilation photons traveled. The position of each of the two photon interactions was determined by the physical position of the small detector itself. The detector is wide open (no collimator) and a very high number of events must be processed in the constant search for two simultaneous events. By comparison, the earliest camera built by Hal Anger required a lead collimator to be placed in front of the NaI crystal, and several PMTs behind the crystal determined to which position on the crystal (i.e., through approximately which hole in the collimator) the single photon traveled. A typical lead collimator stops >99% of the photons arriving to the camera face. Hence, these two different modes of photon detection (PET and planar imaging) defined the geometry and electronics into which these two instruments would separately develop. PET detectors evolved into smaller, more numerous elements (each with its own electronic processing channel or with minimal sharing). Anger camera crystals became larger in area over the years, whereas smaller, more numerous PMTs required that more radiation events be processed by shared electronics. Hence, the evolution of the two well-known species of instruments in nuclear medicine: multiple small-crystal PET rings and large-field-of-view SPECT cameras orbiting around the patient on sturdy gantries.

The hybrids of these two species have never been very popular. The multicrystal camera, made popular by Biard Atomic in the 1970s (*4*), found application in cardiac first-pass studies. Scinticor and thereafter Picker International (now Marconi Medical Systems) marketed this unit under the name of SIM-400 (*5*). Despite the parallel-hole collimator, counting rates in this camera could approach or exceed 1 million per second because the camera at times was clinically exposed to up to 1,000 MBq of activity. However, one does not see many of these in the field.

A 1976 publication (*6*) showed that two opposing large-field-of-view cameras could be used to acquire PET data and even suggested that the bare crystal be filtered to improve the counting rate capabilities. One major problem (it was obviously realized) with this geometry was that many nearly simultaneous events needed to be processed by the electronics behind the large-field-of-view crystals (causing pulse pileup). So, wasn't this pulse pileup problem solved earlier? Why is this article by Wong et al. (*1*) such an interesting contemporary contribution to *JNM*? Three things come to mind:

- The problem of pulse pileup was, in fact, dealt with already in the 1970s. A more recent article by Mankoff et al. (*7*) puts the development of counting rate performance in large-field-of-view cameras in good perspective. Delay line shortening of the preamplifier pulse was one of the earliest methods used for removing the long tails from the pulses, thereby minimizing the pedestals on which pulses pile up. This technique was used in the PENN-PET system (*7*) and has evolved over the years into the CPET offering from UGM Medical Systems, currently available from ADAC Laboratories. I believe that CPET's development of six "large-field-of-view" cameras designed for PET imaging was built on a vision that the major camera manufacturers had not realized earlier. This work ultimately led to the first development of Anger-type cameras by ADAC Laboratories used for clinical PET (*8*).
- Electronics was not as advanced then as it is today. In the 1970s, adding a delayed, inverted, scaled

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copy of each preamplifier pulse to itself removed (clipped) the long tails that contributed to pileup, but all pulses were clipped uniformly (some, indeed, unnecessarily). The method presented by Wong et al. (*1*) attempts to eliminate the individualized remnant signal from previous pulses so that the present pulse can be integrated correctly for calculating the Anger position and energy of each event. For those interested, the authors' earlier article (*9*) gives a more detailed description of the electronic architecture that accomplishes this. This is a state-of-the-art implementation for dealing with pulse pile.

Without a doubt, the authors have taken advantage of some of the most recent electronic innovations to make this advanced front-end possible. My only regret in reading this article is that clearly the most obvious and needed application for such acquisition hardware is for measuring 511-keV annihilation radiation. Although it is a reasonable proof of principle to operate this circuitry at 140 keV, I believe it is a nontrivial step to confirm that it also behaves as expected at higher energies. Because the authors state that ". . .the count-rate capability of the HYPER electronics for ^{18}F is expected to be similar or better," clearly some adjustments must be made to accommodate the differences that will appear at higher energies. Indeed, perhaps some preliminary tests could have been presented using 511 keV. This would have relieved some apprehension as to how difficult it will be to operate this hardware at higher energies. This is akin to designing a supersonic jet and testing it at subsonic speeds. But I am confident that the investigators will break this barrier soon.

- In the 1970s and 1980s, there was little interest in imaging PET radiopharmaceuticals using Anger-type cameras. We need to remember that in the late 1970s, research groups were enhancing the electronics for dedicated PET instruments to measure the time of flight (TOF). The difference in arrival times of the two annihilation photons could be used for a better determination of where the positron annihilation actually occurred. This led to higher quality PET image reconstructions. In the early 1980s, two groups (Washington University and CEA-LETI) were building the first TOF tomographs. The first system put into operation for patient scans was the Super PETT I built at Washington University by Ter-Pogossian et al. (*10*). The Washington University group designed two additional versions of TOF systems. A more detailed history of PET instrument development has been aptly reported by Lewellen (*11*). Needless to say, in this era, detecting positron emitters with TOF PET machines represented a high point in instrument complexity and cost. Little general interest was shown at this time for adapting large-field-of-view cameras for PET applications. Alas, TOF did not survive.

Remember also that at this same time, SPECT imaging with Anger-type cameras was first seriously being developed. The original single-head orbiting camera could not supply sufficient counts within a reasonable imaging time for acceptably low-noise SPECT images. The first triple-head camera was described in 1980 (*12*), and the first images using this design were reported in 1985 (*13*). I think it is safe to say that much effort was spent in the early to middle 1980s by the major camera manufacturers on improving the gantry designs, on securely affixing Anger-type cameras to gantries, and on designing the mechanics necessary to accurately orbit the cameras around the patient.

Finally, I will make one nontechnical point regarding nuclear medicine instruments. In the last decade, the medical care suppliers (insurance companies) have exerted a strong financial pressure to deliver clinical diagnoses as cheaply as possible. I believe this has created a strong impetus to simplify and merge imaging modalities. I reread Dr. Wagner's Highlights of the 1995 Society of Nuclear Medicine annual meeting (*14*) with great interest and retrospective insight, in which he ". . . asked the assembled technologists at this meeting to try to stop using the terms PET and SPECT, replacing these terms with emission tomography (ET). . . . Five years from now, we may not be referring to SPECT or PET scanners, only ET scanners." I believe that his prediction has to a great extent come true. Multiple-head Anger-type cameras are described as PET capable, as operating in coincidence mode, and one manufacturer uses a particularly interesting product name, "gamma-PET." Is it a gamma camera? Or is it a PET instrument? The line has been blurred.

Hence, another oversimplified statement would be: The merger of PET scanners and Anger-type large-field-of-view cameras is being accomplished by solving the pulse pileup problem. And this is exactly the problem on which Wong et al. (*1*) focus. Clearly, other improvements, such as development of axial filters, implementation of multizone independent processing areas on the crystal face, thicker crystals, and even the national network for delivering. ^{18}F-FDG to most hospitals, have contributed to the possibility of imaging positron emitters using upgraded nuclear medicine cameras. From an instrumentation point of view, solving the pulse pileup problem continues to be the central issue. From this point of view, the article by Wong et al. is quite central to the new developments occurring in nuclear medicine imaging and should be remembered as one of the central contributions to merging PET and SPECT into an ET scanner.

It is obvious to everyone that computers and software have been leading the improvements in nuclear medicine imaging, but faster acquisition elec-

tronics (albeit less highlighted) are making just as important an impact on the evolution of our imaging systems. Maximum counting rates of Anger-type cameras have increased from <100,000 counts per second (in the 1960s–1970s) and approached 500,000 in the 1990s. The type of acquisition hardware development presented here promises to make counting rates of a few or several millions per second routinely obtainable in the near future. It is exactly this type of engineering that promises to complete the cycle of nuclear medicine instrumentation evolution, whereby the same front-end electronics is used to detect either 511-keV coincident annihilation radiation or 140-keV single-photon events. I do not mean to say that dedicated PET rings and dedicated collimated large-field-of-view SPECT cameras will disappear. They will not. However, I do believe that perhaps, finally, one hybrid of these two species has found an environment in which it can thrive.

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