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Is LSO the future of PET?

For

If one were to list the ideal combination of properties needed in a detector material to be used for positron emission tomography (PET), the list would include, in no particular order, high speed, high light output, and sufficient density to "stop" gamma rays. Lutetium oxyorthosilicate – LSO – very closely approaches being that ideal material.

The quest for the ideal scintillator for PET imaging has been an ongoing process over the past three decades, and to date, there are only four materials in use. Beginning with the early laboratory PET scanners made in 1976 (and for the following few years), sodium iodide (NaI) was the only reasonable choice for use in PET; in fact, the first commercial PET tomograph was built using NaI detector material [1].

In the early 1980s, bismuth germanate (BGO) and gadolinium oxyorthosilicate (GSO) were both introduced as PET detector materials by EG&G Ortec [1] and Scanditronix [2]. From 1980 through the year 2000, BGO was the dominant detector material used for PET, while NaI and GSO were seldom used in commercial PET detectors.

In 1990, LSO became available in research quantities and immediately caught the interest of the PET imaging community. After an initial development period, the first commercial clinical LSO PET scanners were produced by CPS Innovations in 2001. This new detector material has already made significant contributions to PET, and promises to deliver breakthrough performance for PET tomographs in the next 1–2 years.

Why is LSO considered an advanced detector material? The scientific answer to this question is found in three fundamental detector material parameters – light output, decay time, and density. Table 1 summarizes these parameters for the four detector materials.

The light output of the detector material defines the amount of signal that is available to perform the gamma ray positioning, the energy discrimination (or scatter rejection), and the time window. The time window establishes the noise level resulting from the random events. Therefore, the light output needs to be high to obtain optimal image resolution and minimal noise.

The faster decay time of LSO translates into faster data processing. Also, the faster decay time translates into a shorter time window, and thus a lower level of random noise.

Table 1. Light output, decay time, and density for the four PET detector materials

Characteristic	LSO	BGO	GSO	NaI
Light output (%)	75	15	25	100
Decay time (ns)	40	300	60	230
Density (g/cc)	7.4	7.1	6.8	3.7

Density is a first-order determinant of a detector's ability to stop the 511-keV gamma ray. This parameter, in turn, establishes detection efficiency – the higher the stopping power of the crystal (which is directly related to its density), the greater the detector's efficiency.

In summary, the light output should be high, the decay time should be short, and the density should be high. A figure of merit (F.O.M.) that is used at CPS Innovations for comparing the relative performance of various detectors in PET tomographs is as follows:

$$\text{F.O.M.} = Ep \times (1/\tau) \times (\text{Light output}) \tag{1}$$

where Ep is the PET detection efficiency, τ is the lifetime, and light output is the total number of photons generated in the detector for a single gamma ray event. This in-house figure of merit (Fig. 1) for LSO is 35 times better than BGO and 10 times better than GSO.

Commercialization of LSO technology is no longer an issue, as more than 100 crystal pullers are now dedicated to growing this important detector material. The existing LSO capacity is sufficient to produce material for more than 200 tomographs per year.

LSO has already had a tremendous impact on the speed and image quality of PET and PET/CT designs and made possible a very effective animal research tomograph. This tomograph was designed at UCLA and is manufactured by Concorde Microsystems. In current dedicated PET and PET/CT scanner designs, LSO takes true advantage of 3D imaging. The fast decay time of LSO permits a narrower (6 ns) coincidence window and therefore a dramatic reduction in randoms. The excellent

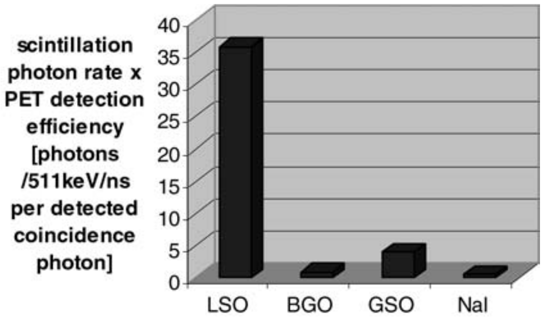

Fig. 1. PET scintillator quality (figure of merit; "scintillation photon efficiency") for the four detector materials (data from Derenzo and Moses [3])

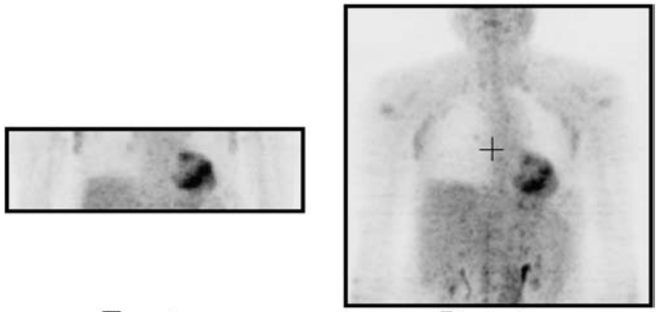

Fig. 2. Comparison of a 15-cm image acquired in 10 min on a conventional BGO scanner (Exact; full 3D OSEM reconstruction) (*left*) and a 52-cm image acquired in 2.5 min on the new LSO prototype (*right*), in a patient investigated in one bed position for recurrence of breast cancer. The image acquired with the prototype illustrates its potential to reduce imaging time

energy resolution (approximately 12% full-width at half-maximum) permits a narrow energy window, which decreases the scatter in the image.

This is particularly important for 3D PET acquisitions. Traditional 2D acquisitions employ lead septa, which essentially collimate PET events to a narrow 2D band, but also serve to block "scatter" events. Removal of the septa for 3D acquisitions takes full advantage of the 360º nature of PET emissions by accepting a far wider angle, thus improving sensitivity approximately fivefold. More information is gathered in the 3D process, and it is LSO that allows this additional information to be processed faster. The integration of LSO with current technology allows CPS Innovations to manufacture PET scanners that deliver the highest image quality with the fastest scan times in the industry.

Our continued research into the design of PET systems will take greater advantage of the characteristics of LSO. CPS Innovations is now completing work on the next generation PET tomograph [4]. Key specifications and features are: patient port 70 cm, axial length 52 cm, pixel size 4×4 mm, noise equivalent count rate 140,000 cts/s, coincidence window 6 ns, and simultaneous emission/transmission. The noise equivalent count rate for a state-of-the-art PET tomograph today is in the range of 50,000–60,000 cts/s.

Because of the aforementioned advantages of LSO technology, the whole-body imaging time for the torso can be reduced to approximately 5 min in commercial scanners within the next year or two, a milestone that will represent a major operational advance. Figure 2 shows the first set of patient images acquired while examining for recurrent breast cancer.

The panel-design tomograph is a fully 3D unit built to take advantage of the unique properties of LSO. The goal is to achieve high count rate sensitivity while performing simultaneous emission/transmission acquisitions. This will result in significantly shorter whole body scan times (compared with those achievable on current scanners), while image quality is maintained. The detector heads of the panel-design tomograph have been designed to take advantage of the superior properties of LSO: high light output, short scintillation light decay time, and high stopping power. The high light output and short scintillation decay time of LSO allow the scintillators to be arranged in panels of size 370 mm (transaxial) by 530 mm (axial) comprising 10,080 individual crystals 4×4×20 mm^3 coupled to 88 photomultiplier tubes. Five panels in a hexagonal array continuously revolve around the subject within a closed gantry with an aperture of 700 mm and an axial coverage of 500 mm. The sixth side holds $^{68}Ge/^{68}Ga$ point sources in a design that allows simultaneous emission and transmission acquisitions.

This new 3D tomograph, designed to capitalize on the advantages of LSO, outperforms current tomographs and offers the potential to reduce whole-body acquisition times from more than 30 min to less than 5 min while maintaining optimal image quality. Other benefits, such as reduced patient motion and minimized effects of non-steady state isotope distribution, also contribute significantly to improved image quality.

In summary, LSO offers the best combination of properties for PET of any scintillator known today. The superiority of LSO over previous detector technologies means that this new generation of PET scanners will change the nature of PET scanning. In the very near future, the medical community will look to PET and PET/CT for the most reliable, efficient, and cost-effective diagnostic imaging available.

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Against

Yes and no. The answer is *yes* if the question is interpreted as whether or not LSO will be an important scintillator for positron emission tomography (PET) scanners in the future. The answer is *no* if the question is interpreted as whether or not LSO will be the *only* scintillator for PET scanners in the future.

LSO improves the performance of 3-D PET

Certainly lutetium oxyorthosilicate (LSO) has drawn considerable attention since it was first reported on by Melcher and Schweitzer [1] in 1992. It has a very favorable combination of properties for PET imaging compared with the other scintillators then (and still) in use in PET, namely bismuth germanate (BGO) and sodium iodide (NaI) (Table 1). LSO was first used in the microPET animal scanner from UCLA [2] and the research brain HRRT scanner from Siemens/CTI [3], but it was only in 2001 that LSO was offered in a commercial whole-body scanner intended for clinical use – the Siemens/CTI Accel.

The advantages of LSO are best appreciated for 3-D imaging without septa. For 2-D imaging with septa it is hard to beat BGO, since it has the highest stopping power of all scintillators commonly used in PET. The low light output of BGO leads to poor energy resolution, but this is relatively unimportant in 2-D PET since the septa limit scatter and randoms. The advantage of 3-D imaging is the large gain in true sensitivity, but this is partially offset by a gain in scatter and random coincidences, both of which decrease contrast, often leading to decreased lesion detectability in fluorine-18 fluorodeoxyglucose (FDG) studies. A common way to evaluate the trade-off of signal (true coincidences) vs noise (scatter and randoms) is to measure the noise-equivalent count rate (NEC) [4]. The NEC accounts for the additional noise that scatter and randoms contribute to the image, even though correction methods are used to compensate for the bias from scatter and randoms. In comparisons using BGO scanners with retractable septa, the NEC is higher in 3-D (septa out) for small activity distributions [5] such as those encountered in brain studies, but the gain is modest for whole-body studies, since scatter and randoms increase and there is unshielded activity from outside the field of view (FOV). In fact, comparisons of patient studies using BGO scanners indicate better image quality for 2-D for whole-body oncologic studies [6]. But, BGO is definitely not the best scintillator for 3-D. 3-D PET demands a fast scintillator to reduce dead time and randoms, and one that has good energy resolution to reduce scatter and randoms from inside *and* outside the FOV. In both speed and energy resolution, LSO has a clear advantage over BGO, and the NEC for the Accel in 3-D [Dr. M. Casey, personal communication] – as measured with the 70-cm-long phantom [7] which is relevant for whole-body studies – is higher than the EXACT, which is very similar in overall design, but which uses BGO instead of LSO. In clinical practice, the Accel scanner performs well in 3-D for both brain and whole-body studies.

GSO as an alternative to LSO

The Allegro scanner introduced in 2001 by Philips Medical Systems uses gadolinium oxyorthosilicate (GSO), an alternative to LSO. The GSO Allegro operates exclusive-

Table 1. Comparison of properties for scintillators currently used in PET. Energy resolution measured at 662 keV for a single crystal coupled to a single PMT; thus the measured value does not include effects of detector design. Data from [10] and [11]

Scintillator	NaI(Tl)	BGO	GSO	LSO	LuAP	LPS	LaBr
τ (ns)	230	300	60	40	18	30	35
μ (cm^{-1})	0.35	0.95	0.70	0.86	0.95	0.70	0.47
$\Delta E/E$ (%)	6.6	10.2	8.5	10.0	~15	~10	2.9
Rel. light output (%)	100	15	25	70	30	73	150