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Suitability of Epitaxial Ga-As for X-Ray Imaging

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□ ABSTRACT □

The rate of indirect photon-electron conversion for scintillator materials coupled with arrays of photodiodes is at least 25 times smaller than the rate of direct conversion. We examine the conditions to be fulfilled by semiconductors undergoing such direct conversion to be applied to X-ray imaging. Bulk grown materials are not well-suited to this application because large defect concentrations give rise to strongly non-uniform electronic properties. We argue that epitaxial layers are suitable for use as imaging devices and we illustrate our argument using the case of thin epitaxial GaAs layers. Detectors made with such layers exhibit a good energy resolution, and a charge collection efficiency which approaches one.

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ملاءمة وصلاحية الكواشف من النوع Ga-As من أجل التصوير بأشعة X

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□ الملخّص □

إن نسبة التحويل غير المباشر لـ "فوتون – الكترون" في المواد الوميضية – المقرونة بترتيب معين لثنائيات ضوئية أقل بحوالي 25 مرة من نسبة التحويل المباشر. وسوف نتفحص الشروط التي يجب أن تتوفر في أنصاف النواقل عند خضوعها لظاهرة التحويل المباشر، وذلك عند استخدامها للتصوير بواسطة أشعة **x**.

إن الصفائح المتوفرة من المركب **Ga-As** غير ملائمة لهذا النوع من التطبيقات، بسبب ارتفاع كثافة العيوب التي تؤدي إلى عدم انتظام في خواصها الالكترونية.

وسوف نحاول أن نبرهن على أن الكواشف، من نوع **Ga-As** التي تعتمد صناعتها على تقنية التنضيد، أي توضع الطبقات فوق بعضها البعض بالتبخر هي الطريقة الوحيدة الملائمة للاستخدام في أجهزة التصوير بأشعة **X**، وسندعم حجتنا هذه باستخدام طبقات رقيقة من المركب **Ga-As**. وتظهر الكواشف المصنعة من مثل هذه الطبقات الرقيقة ميز عالي للطاقة، ومردود وتجميع شحنة فعلية قريب من الواحد.

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INTRODUCTION:

The advantage provided by digital imaging methods recently gave an impulse to the development of new X-ray imagers for medical applications. These new imagers are flat panel devices based on a combination of a scintillator material coupled to an array of photodiodes (for a review see ref.l). Such imagers present only a limited improvement over the photographic film in terms of dose received by the patient, which is reduced typically by a factor of 2 [2]. A photon of 50 keV produces about 10^3 visible photons with a typical commercially-available flat panel [3], i.e. a maximum of 500 electrons, corresponding to 10 electrons/keV.

In medicine, the need for digital detectors is further increased by the requirement to drastically lower the dose received by the patient (and the medical staff). Dose reduction can be achieved using semiconductor detectors because they allow direct photon-electron conversion, with a conversion efficiency considerably higher than that of flat panels. Indeed, for a material characterized by a gap of 1.5 eV, the number of electron-hole pairs created by a 50 keV photon is $l.l \times l0^4$, i.e. 220 electron/keV, which is therefore much higher than for a scintillator.

X-ray semiconductor detectors working at room temperature are made of high Z, atoms to absorb efficiently X-ray photons, also they must have a forbidden gap of about 1.5 eV (large enough to lower the electronic noise and small enough to produce a maximum number of electron-hole pairs). Among the few materials that fulfil these requirements, only CdTe, or CdZnTe (CZT), are currently available in large enough quantities. CZT detectors exhibit a charge collection efficiency which approaches one and a good energy resolution even for photons of low energies [4]. However, CZT materials are expensive, of small dimensions, their electronic properties are not uniform and there is no standard technology to produce pixel detector structures. Because they contain a large concentration of defects [5, 6], CZT detectors show polarization (reduction of the electric field due to charge trapping on defects) and afterglow (transient currents due to the return to equilibrium of the charge distribution of the defect levels) effects [7]. Hence, CZT is not well adapted to X-ray imaging. An alternative material is GaAs: Ga and As are high Z atoms, the gap of GaAs is 1.42 eV, the technology is standard and this material can now be obtained as large area wafers (up to 6 inches in diameter). It has been demonstrated that high energy resolution detectors can be made with it [8-10]: for X-ray detection, GaAs has in practice similar performances as Ge, with the advantage of operating at room temperature. Unfortunately the GaAs material that is available contains a large concentration of defects (the so called EL2 defects) which act as very efficient non radiative recombination centers and are inhomogeneously distributed [11].

A suitable material for X-ray imaging should be large area, free of defects, with uniform electronic properties and for which a standard technology exists. Up to now, the only available materials with the thickness required to absorb X-ray photons efficiently are bulk ones, which are neither free of defects nor with uniform electronic properties. The only way to obtain a material with a low defect concentration and uniform properties is to use epitaxial growth.

In this paper we present results obtained from thin epitaxial GaAs layers grown by a novel method explained in reference [12]. The growth technique, which is low cost and non-polluting, produces thick layers (several hundred micrometers) in a reasonable time (few hours). Photoluminescence mapping has shown [13] that these layers exhibit a very

homogeneous minority carrier lifetime over the whole area of the layers (grown on 4 inches wafers). Since the lifetime is very sensitive to deep level defects that lower the charge collection efficiency, this implies that these defects are homogeneously distributed. A Deep Level Transient Spectroscopy observation has shown [13] that the concentration of deep defects is 7×10^{13} cm^3. Detectors composed of $p^+/i/n^+$ structure in which the region i is the non-intentionally doped thick epitaxial layer have been fabricated [14]. Some of their detection properties have been evaluated, such as the energy resolution which is found to be similar to that of Ge [15-17].

The aim of this communication is to describe some of the properties of these detectors that are important for imaging, namely the linearity of the response, the charge induced polarization and the afterglow effect.

EXPERIMENTAL AND RESULTS:

The detector used in this work has a size of 1 mm^2 and the thickness of the i layer is 125 μm. The photocurrent induced by a commercial medical imaging X-ray source (Cu tube) has been measured over a resistance in series with the detector, with an anode voltage of 40 kV and a bias of 40 V.

The linearity of photocurrent versus X-ray flux is shown by the value (one) of the slope of the log-log plot given in fig.1, in which the photon flux is adjusted by varying the anode current (10 to 50 mA) and the detector distance (11 to 137 cm). The vertical shift between the data lines corresponding to different values of distance is a result of increasing X-ray absorption by air. The linearity has been monitored over three orders of magnitude, limited by the maximum available X-ray flux. No saturation is expected up to a significantly higher flux, when radiative recombination will occur. No limitation due to space charge effects exists since the concentration of deep defects is negligible and the generation of electron-hole pairs is uniform.

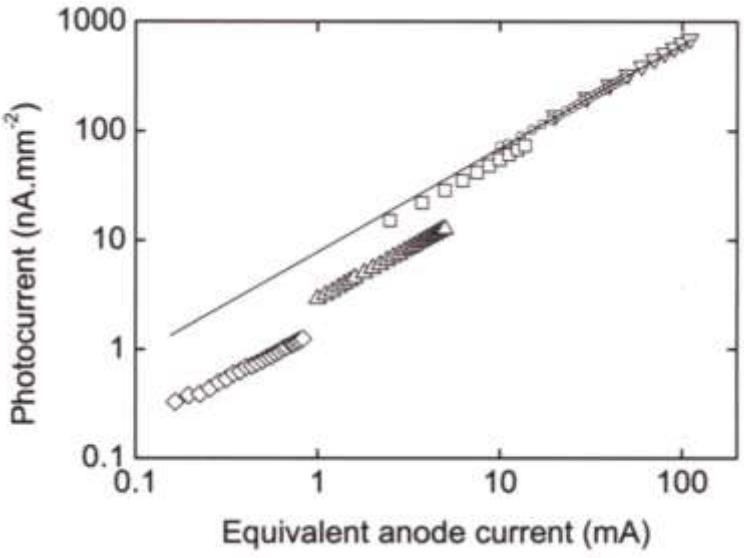

Figure 1. Linearity of the photocurrent versus X-ray flux, which is adjusted by varying the anode current and the detector distance (∇: 11 cm, O: 17 cm, □: 34 cm. Δ: 51 cm, ◊: 137 cm). Taking the anode current at 17 cm distance as the reference, the equivalent anode currents are calculated according to the relation between the flux and the distance. The line corresponds to the data once corrected for the absorption in air.

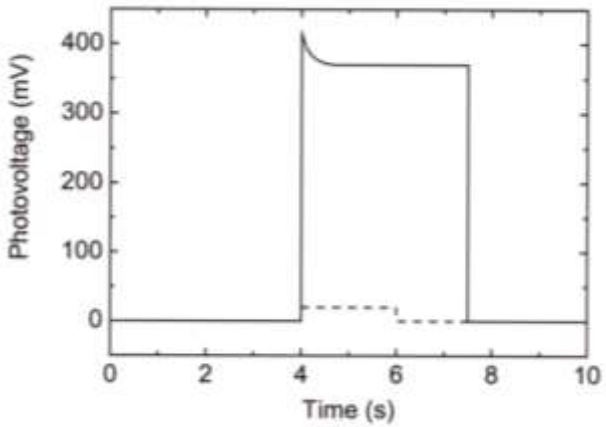

Figure 2. Time response of the photocurrent signal (measured on 1 MΩ resistance) for a distance of 17 cm and anode currents of 50 mA (solid line) and 10 mA (dash line). The X-ray pulse is obtained by opening and closing the shutter of the X-ray machine.

The time response of the detector is given in fig.2. The signal is obtained by applying the X-ray pulse for a few seconds. It demonstrates that a polarization effect, recognized by the decrease of the signal amplitude with time, exists. The time to reach stability is of the order of 0.5 second. This decrease of signal amplitude is limited to 10% while it reaches 80% in the case of CZT detectors [7]. It is detected only at high fluxes. typically higher than 50 mA equivalent current (see fig.2). No remaining signal is observed after stopping the X-ray pulse, which means that the afterglow, associated with the return of the defect states to equilibrium, is negligible.

An attempt to measure the decay time has also been performed using a chopper system equipped with a 3 mm Pb screen to modulate the X-ray beam. As shown in fig.3, the signal is interrupted when the Pb screen covers the detector. The fall and rise times are of the order of 20 μs, corresponding to the time for the Pb screen to cover the detector. This implies that the intrinsic time constant of the detector is shorter than 20 μs. This time constant should actually be the minority carrier lifetime, which is expected to be of the order of 10 to 100 ns in such material.

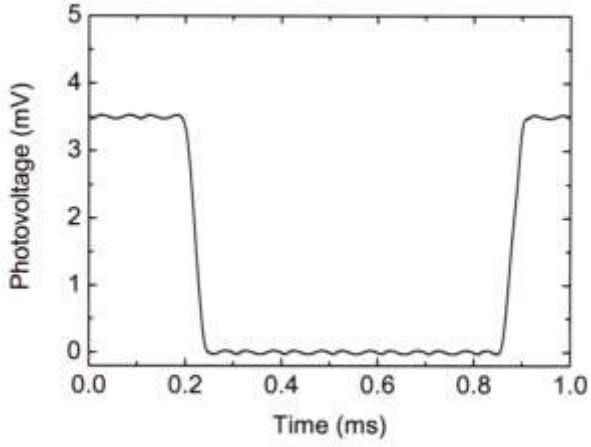

Figure 3. Photocurrent signal (measured on 10 kΩ resistance) chopped by a 3 mm Pb screen, for a distance of 17 cm and an anode current of 55 mA.

CONCLUSION:

In conclusion, the performance has been demonstrated of epitaxial layers of GaAs with a thickness adapted to the photon energies of medical X-rays. This material will allow the production of direct-conversion digital imaging devices which will offer a dose reduction factor of at least 10, as compared to the currently available flat panel devices. These devices will also have the potential production of colour images, due to the good energy resolution of the material. Finally, because of the absence of defects, no afterglow is present which should permit a fast acquisition (important in particular for CT scanners). The only advantage of CZT, a larger absorption coefficient at high energy, is at least partially compensated by non-existence of polarization effects. The only actual limitation we are aware for such epitaxial GaAs layers is that the residual doping impurities, being not compensated because of the low defect concentration, limits the width of the depleted zone in which the created electron-hole pairs are collected. However there are ways to overcome this difficulty [18].

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