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The design of a system for coloured digital radiology with VLSI circuits and GaAs pixel detectors

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Abstract

We describe the design of a digital radiology system with GaAs pixel detectors, based on the energy selection of the non absorbed X-ray spectrum. We present a general layout; we show that simulation and experimental data corroborate the idea for the feasibility of the system.

1. Introduction

The use of GaAs as a high-efficiency X-ray imaging detector is now well known [1].

An idea for a novel digital radiology device with inherent high spatial and energy resolution has been proposed [2]. It is based on a GaAs pixel detector and two custom-made VLSI electronic circuits which process each photon pulse and classify it according to its energy.

The circuits store the counting and the energy information, allowing the acquisition of *coloured* X-ray images. The artificial elaboration of colour can be performed on line in a scale proportional to the energy of the detected photon. The scale resolution will be limited by the electronic circuits design.

The method, generally applicable, has been illustrated before [3]. Here we describe the design for a digital imaging system with an X-ray source, a GaAs pixel detector and the VLSI circuits. The paper concentrates on the general layout in a realistic configuration. We address the compromises between the performances of the system, versus the fabrication technologies risks and the costs.

We have at present a first version of the detectors with 32 pixels and the new chip with 8 energy intervals.

We show data and compare with the simulation proving that the design idea is well based.

2. The basic principle

In radiology, the use of the energy and counting information is performed simultaneously in *dual energy systems*. There, two X-ray beams with

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different energies are used to acquire two images: their comparison allows a better contrast of the resulting radiography. However, this leads to relatively complex and expensive devices with the images not being available *on the spot*.

The system we propose, uses a commercial X-ray tube as a source and, as a detector, a column of squared GaAs pixel detectors, and two custom VLSI circuits. The principle is to create n images corresponding to n energy windows from a pulse height analysis of the detected photons performed on the chips. This correponds to a comparison between the emission spectrum of the X-ray tube and the spectrum after traversing the object. By subdividing the spectrum into several small energy intervals, one obtains images for each one of these. Thus the image of, e.g., human soft tissues, is enhanced by the lower-energy part of the spectrum, while hard components, like bones, are visible via the higher energy part. This method allows to distinguish materials with different effective atomic numbers by separating the different absorption cross-section components as much as possible.

We have designed a detector module made of three basic components (Fig. 1): the GaAs pixels, to be wire bonded to a first VLSI custom chip, performing the analog operations and wire bonded to a second one mainly dedicated to the peculiar energy selection via digital operations. In the following, we present a comparison between the simulated and the experimental data for the absorption of hard and soft materials after addressing the design specifications and describing a realistic layout.

3. The design of the system

We have adopted the approach of separating the detectors from the VLSI circuitry and the analog from the digital functions. Highly integrated systems can then be approached at a later stage, but only with regard to more satisfactory performances. This choice sacrifices only very little of the integration advantages (mainly the physical occupation of the device and its handiness), saving all of

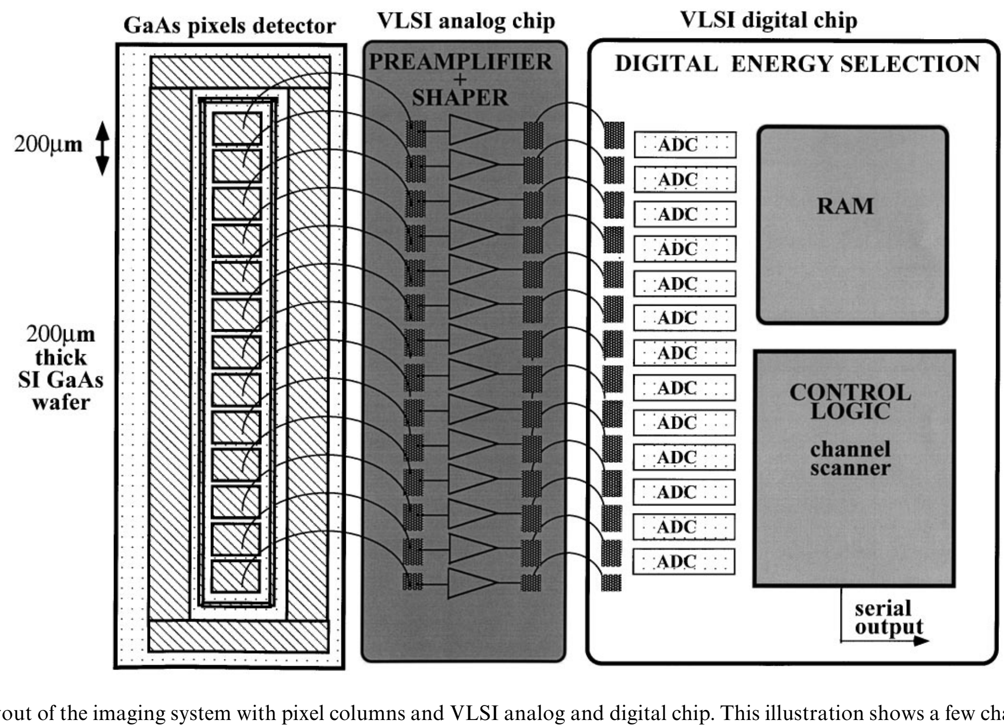

Fig. 1. Layout of the imaging system with pixel columns and VLSI analog and digital chip. This illustration shows a few channels only.

the expected digital performances. On the other hand only very stringent requirements on the foreseen applications should actually lead to high integration. Our choice therefore leaves also open many imaging applications and will allow us a deep understanding of the performances before a dedicated system can be built in series.

Although the system can generally be applied to imaging, in order to explain our design choice, we limit our considerations here to diagnostic digital radiology. By using GaAs, this should allow for a reduction of the dose applied to the patient. The energy selection can give a better contrast and new diagnostic criteria. In this design we had to face the VLSI design constraints, using both full-custom and the standard-cell approach, wherever suitable, and the requirements of low noise and speed for the above mentioned low-dose requirements.

We present a system to perform a full scan on a $\approx$ 10–20 cm object. In Fig. 2 the set-up is shown schematically: a precision motor moves a rigid structure with the X-ray tube by 100 μm steps. This structure consists of the collimator slits right after the tube and in front of the sensors and VLSI electronics chips boards. The whole structure is movable along the direction orthogonal to the pixels column. The object under diagnose is kept fixed, held on top of a plastic table. The X-ray tube with a tungsten anode uses a high voltage of 50–100 kV with anodic currents of the order of 10^{-2}–10^{-3} A. In front of it, a 100 μm collimator slit directs the beam to the object at a distance of 50 cm. In front of the detector, fixed to the structure, the other 200 μm collimator is placed in the direction of the detector column, over the whole length (which in the final configuration is 256 cm corresponding to 128 pixels).

3.1. The GaAs detectors

The detectors have been manufactured at Alenia,[1] on a joint INFN design, on a semi-insulating GaAs wafer, 200 μm thick. The metalization is gold and titanium. Fig. 1 shows the layout of the mask, the pixel size is $200 \times 200\,\mu m^2$. The pitch size suits the current detector and assembly technologies. A finer pitch would require a very challenging technology in terms of detector fabrication, assembly and VLSI integration. On the other hand, the 100 μm step scanning, can, by dithering, give a better image contrast [4], comparable to the one of 100 μm pitch. This has to be considered more or less a limit in these imaging applications, since the estimated diffusion of the photoelectrons, with respect to the impact point inside the detector material, can be as large as a few tenths of a μm.

The two guard rings should guarantee a more stable operation. In the final configuration, this is relevant since it introduces dead zones. An adequate scanning procedure can be adopted to avoid this problem. This leads however to a longer exposure time, which is a crucial parameter for the application of this system in diagnostic imaging on patients; the design we have adopted for the detector is a general one and it is aimed to demonstrate the validity of the method. The detector layout can be adequately tuned for specific applications.

3.2. The VLSI chips

For the electronic functions, we have simply separated the analog from the digital part in order to minimize the risk of high integration.

For the low-noise analog chip we have adopted a 0.8 μm BiCMOS design of a charge pre-amplifier and a $CR\text{-}RC^4$ shaper. The performances, with a detector capacitance of the order of 1 pF and a peaking time of the shaper of 1 μs, should guarantee a sufficient figure of ENC $\approx 200e^-$ in a 10 μs readout time, with a power consumption less than 7 mW/ch.

The digital chip consists essentially of a series of very compact pulse height analyzers. Every channel consists of an ADC and one memory which stores the energy information. At the end of each acquisition cycle, a histogramming circuit produces the spectrum of the detected photons, grouped in several energy windows.

The resolving power of the image colours depends on how many energy intervals can be resolved and how well they are defined with respect to each other. This is defined by the different discriminator

[1] Alenia S.p.A., Direzione Ricerche, via Tiburtina 12.4 km, Roma, Italy.

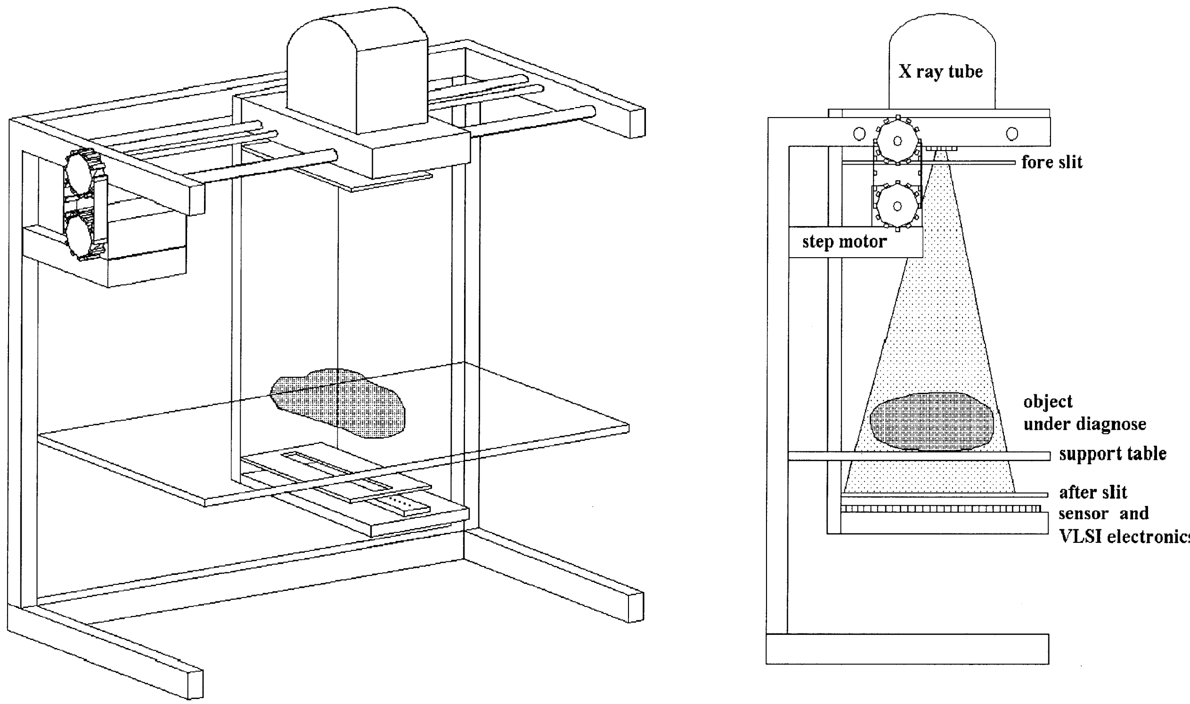

Fig. 2. View of the full imaging system. The step motor moves the X-ray, the collimators and the detector, scanning a fixed object in a direction orthogonal to the pixel line.

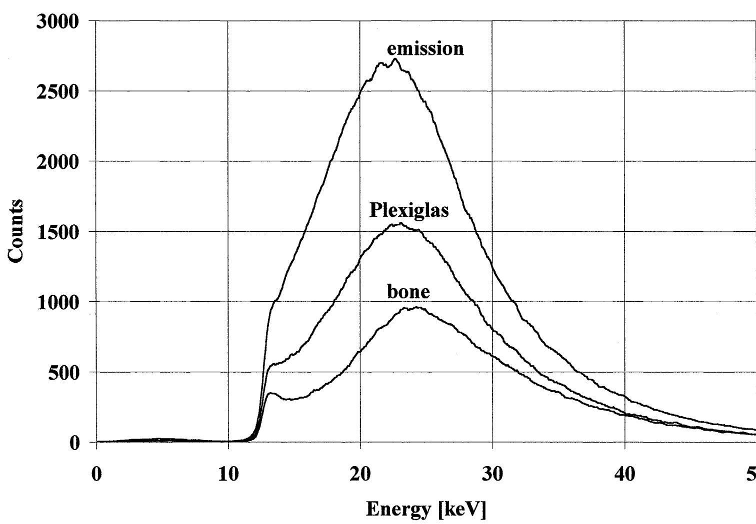

Fig. 3. The spectra of the non-absorbed photons, after tranversing a 1.6 cm Plexiglas and a 0.2 cm bone slab. The corresponding emission spectrum without any material in front, is also plotted.

thresholds. Fluctuations of these thresholds should be limited. The adopted design shall guarantee a safe operation. We have realized an 8 levels discrimination chip and we are designing a new one with 64 levels of discrimination: in both cases we foresee a threshold variation well within the intervals set.

4. Simulation and experimental results

In the radiological energy interval under consideration the difference between the photoelectric and Compton cross sections is more pronounced for heavy materials than for softer tissues. To verify that, we have irradiated two different material slabs. The spectrum of the photons traversing them without being absorbed, is illustrated in Fig. 3. There the energy spectrum of the photons is plotted after traversing a Plexiglas and a bone slab, of 1.6 and 0.2 cm thickness, respectively. The two are clearly distinct, in particular in the interval 15–30 keV. The corresponding ratios of the two spectra, with respect to the emission obtained without any material, is plotted in Fig. 4, to be compared with the corresponding simulation in Fig. 5. The simulation has been done on the equivalent

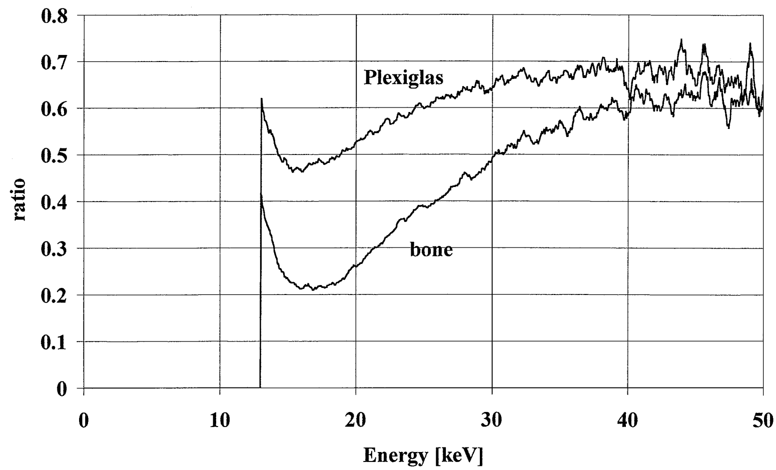

Fig. 4. Plots of the ratios of the spectrum for the Plexiglas and bone slabs.

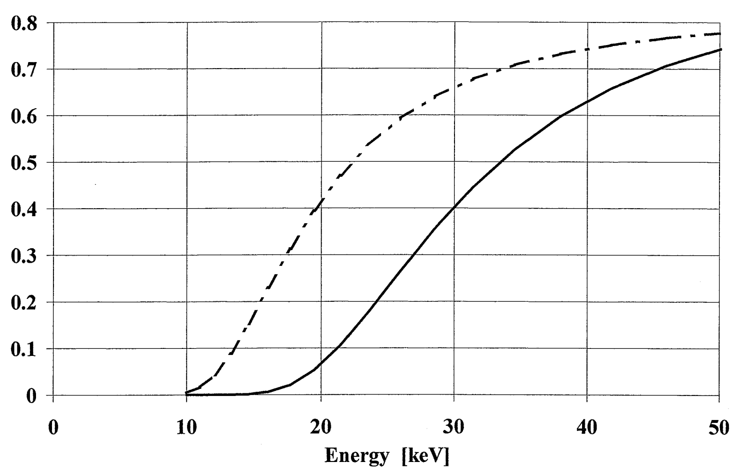

Fig. 5. Spectrum ratios of simulated X-rays absorbed in water (similar to soft tissue) and calcium (similar to bones).

corresponding effective-Z materials: water and calcium. The behaviour of the two curves agrees well and, most important, the two materials are neatly distinguishable. This confirms the expected difference of the cross sections and proves that the principle of the design of the system for coloured digital radiography based on the energy intervals separation is sound.

5. Conclusions

We have described the design guidelines for a digital radiology system with GaAs pixel detectors, which, by an energy selection of each detected photon, produces colour digital images. This very promising idea is now turning into a real project. The simulation results and the conservative, although original, design choices give us confidence in the system designed for many applications.

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