Performance comparison of CMOS-based photodiodes for high-resolution and high-sensitivity digital mammography

This article has been downloaded from IOPscience. Please scroll down to see the full text article.
2011 JINST 6 C12046
(http://iopscience.iop.org/1748-0221/6/12/C12046)

View the table of contents for this issue, or go to the journal homepage for more

Download details:
IP Address: 143.248.118.104
The article was downloaded on 28/03/2012 at 02:29

Please note that terms and conditions apply.
Performance comparison of CMOS-based photodiodes for high-resolution and high-sensitivity digital mammography


Dept. of Nuclear & Quantum Engineering, Korea Advanced Institute of Science and Technology, 291 Daehak-ro, Yuseong-gu, Daejeon 305-701, Republic of Korea

E-mail: kuria@kaist.ac.kr

ABSTRACT: In order to develop a high-resolution and high-sensitivity digital mammographic detector, to use a commercially-available and well-developed CMOS image sensor (CIS) process can be a cost-effective way. However, in any commercial CIS process, several different types of n- or p-layers can be used so that various pn-junction structures could be formed depending on the choice of n- and p-layer combination. We performed a comparative analysis on the characteristics of three types of photodiodes formed on a high-resistivity p-type epitaxial wafer by applying three available n-layer processes in order to develop the high-sensitivity photodiode for a scintillator-based X-ray imaging detector. As a preliminar study, a small test-version CIS chip with an 80 × 80 pixel array of a 3-transistor active pixel sensor structure, 50 µm pitch and 80% fill factor was fabricated. The pixel area is subdivided into four 40 × 40 sub-arrays and 3 different types of photodides are designed for each sub-array by using n⁺, n⁻ and n-well layers. All other components are designed to be identical for impartial comparison of the photodiodes only. Among 3 types, the n⁻/p-epi photodiode exhibited high charge-to-voltage gain (0.86 µV/e⁻⁻), high quantum efficiency (49 % at 532 nm wavelength) and low dark current (294 pA/cm²). The test CIS chip was coupled to a phosphor screen, Lanex Fine or Lanex Regular, both composed of Gd₂O₂S:Tb, and was tested using X-rays in a mammography setting. Among 6 cases, n⁻/p-epi photodiode coupled with the Lanex Regular also showed the highest sensitivity of 30.5 mV/mR.

KEYWORDS: X-ray detectors; Photon detectors for UV, visible and IR photons (solid-state); X-ray mammography and scintillator- and MRI-mammography

© 2011 IOP Publishing Ltd and SISSA
doi:10.1088/1748-0221/6/12/C12046
1 Introduction

The main purpose of X-ray mammography is to diagnose breast cancers in their early stage, so the ability to detect fine structures with lower patient dose is one of the most important requirements for a mammographic detector. Although conventional film-screen systems still produce high quality X-ray images, mammography is very rapidly progressing toward digital X-ray systems due to their well-known merits: the fast image acquisition and display, the versatile post image processing by using software, the efficient storage and transport of the huge size of image data, and the potential to reduce the dose and so on [1, 2]. Amorphous-silicon (a-Si) and amorphous-selenium (a-Se) based flat-panel detectors and charge coupled device (CCD) with fine optical lens assembly are the commonly used detector types for general digital radiography. However, these detectors are not easily applicable to high-resolution and high-sensitivity digital mammography, since the minimum-feature size in a large a-Si flat-panel fabrication process limits their spatial resolution and the use of demagnifying optics in a CCD system causes lower X-ray sensitivity, or poorer image quality for the given patient dose.

Recently, large-area CMOS image sensor (CIS) has been in the spotlight as a powerful candidate detector for digital mammography. As great advances are made in CMOS fabrication processes for mainstream memory and logic, the minimum-feature size is getting smaller and smaller so that the pixel pitch of the CIS can be extremely small, less than a few micrometers. Thus, the CIS pixel pitch for vision applications such as a digital camera tends to be smaller and smaller to get better spatial resolution. This advanced CMOS technology can also be applied to fabricate a high-resolution X-ray detector for medical diagnostic applications where much larger pixels are required. In addition, the CMOS X-ray detector benefits from the inherited features of well-developed and standardized CMOS technology such as low power consumption, high speed, low...
production cost, and above all the full integration of signal processing circuitry in the same sensor chip. Application of the commercially-available standard CMOS fabrication process to X-ray mammography is restricted by reticle and wafer size, so stitching and tiling techniques were tried in order to overcome these limitations and have been under active development [3–6].

While a few digital mammographic detectors fabricated in the CMOS fabrication process were already commercialized recently, the need remains to be able to develop a high-sensitivity and high-resolution detector with lower production cost. In X-ray imaging, the CIS should be coupled with a phosphor or a scintillator where incident X-ray is converted into visible light photons, so the resolution and the sensitivity of the CIS are not just about the property of the CIS itself. Generally, if the scintillator gets thicker, the X-ray sensitivity increases owing to the increased number of visible light photons generated in the scintillator but the effective spatial resolution decreases because of the lateral spreading of the photons. Moreover, depending on the type of scintillation materials selected, the visible light photons emitted from the scintillator have different wavelengths and thus have different absorption length in silicon. For example, Gd2O2S:Tb, one of the most popular scintillation materials, generates visible light photons with a wavelength range between 380 nm and 600 nm and with the peak intensity at 532 nm. The absorption length ranges from 0.1 µm to 1.7 µm and is approximately 1.2 µm for the wavelength of 532 nm. Therefore, the performance parameters of every critical component of the X-ray imaging system should be well-matched or independently improved with new technologies so as to increase both the resolution and the sensitivity at the same time.

In this study, we have focused on possible methods of implementing various structures of photodiodes within the given boundary conditions of utilizing the standard CMOS process and coupling to widely available scintillator screens for the cost-effective mammographic-detector fabrication with the optimal spatial resolution and X-ray sensitivity. In line of this thinking, for vision and space applications, several prior works have evaluated and compared the imaging performance of CMOS-based photodiodes. In [7], it was reported that n⁺/p-sub and n-well/p-sub photodiodes have approximately the same response for white light. In [8] and [9], it was shown that the optical properties of a photodiode significantly vary with a photodiode structure and a fabrication process. In [10], it was described that the spectral response and the dark current of an active pixel strongly depend on the junction depth and the implanted doping concentration of the photodiode. From the results of these works, it is evident that characterizing the photodiode with the variation of photodiode structures for the given fabrication process has to be preceded before developing the CIS for high-sensitivity digital mammography. In addition, to our knowledge, there are few studies on optimizing the imaging performance of the CIS in combination with the scintillator for high-resolution full-field digital mammography where a 2-D photodiodes array with the order of 50 × 50 µm² pixel size is necessary [11].

We designed small test-version 3-transistor active pixel sensor (3-tr APS) arrays with three different types of photodiodes formed by every n-doped layer available in a typical CIS process on a high-resistive p-type epitaxial (p-epi) wafer. The prototype chip was fabricated in a 0.18 µm 1-poly/4-metal CIS process. The optical properties of the three photodiodes were evaluated and analyzed comparatively in the viewpoint of optimal matching to scintillator screens for future fabrication of the full-field high-sensitivity and high-resolution digital mammographic detector. For this purpose, we simply measured and estimated three key imaging parameters: charge-to-
Figure 1. (a) Circuit diagram of the 3-transistor active pixel sensor. Cross-sectional diagrams of (b) Type A: n−/p-epi, (c) Type B: n-well/p-epi and (d) Type C: n+/p-epi photodiode.

voltage conversion gain (CVCG), quantum efficiency (QE) and dark current (DC). Furthermore, to investigate X-ray sensitivity of the photodiodes, two screen type scintillators, Lanex Fine (LF) and Lanex Regular (LR), based on Gd$_2$O$_2$S:Tb were coupled with the chip and tested under practical mammographic X-ray beam conditions.

2 Materials and methods

2.1 Photodiode structures and pixel design

In order to evaluate and compare the characteristics of all types of CMOS-based pn photodiodes achievable in a typical CIS process, we thoroughly investigated all possible n-layer processes on a p-type wafer, and designated three relevant processes. These are an n− layer used in a pinned photodiode process [12, 13] as well as an n-well and an n+ implant layer used in a standard CMOS logic process. These three distinct processes could produce three different pn junctions with different doping profiles and junction depths which are normally defined by the distance from the silicon surface. The cross-sectional structures of all these three photodiodes (Type A, Type B and Type C) are shown in figure 1(b), (c) and (d), respectively. For all the photodiodes, junctions were formed in a lightly-doped p-epi layer in common which has a resistivity of 30 Ω•cm and a thickness of 5.5 μm. The junction depth of the Type A photodiode is estimated to be approximately 0.7 μm and the peak doping concentration is in the order of 10$^{16}$ cm$^{-3}$. For the Type B and Type C structures, the
Figure 2. (a) Chip layout and (b) packaged test chip on a printed circuit board. Test chips coupled with (c) Lanex Fine and (d) Lanex Regular which have a thickness of 46 µm and 91 µm, respectively.

Junction depths are about 1.3 µm and 0.5 µm, and the peak doping levels are in the order of $10^{17}$ cm$^{-3}$ and $10^{20}$ cm$^{-3}$, respectively.

For impartial comparison, the same pixel structure, the 3-tr APS, was applied to designing the test pixels for these three photodiodes and its circuit diagram is shown in figure 1(a). The 3-tr APS consists of a photodiode, a reset transistor (M1), a source follower (M2) and a selection transistor (M3). The operation of the 3-tr APS is composed of 3 phases; the reset, integration and readout phase. Operation begins with turning on M1 by a row reset control signal so the D node voltage resets to the level equal to $V_{DDA} - V_{THM1}$, where $V_{DDA}$ is the bias voltage and $V_{THM1}$ is the threshold voltage of M1. After M1 is turned off, the second phase or X-ray signal integration time begins. During this time, the electron-hole pairs generated by the light photons in the photodiode junction start to discharge the D node voltage continuously for the integration time. After the integration time, M3 is turned on by the second row select control signal, and then M2 transfers the decreased amount of the D node voltage as a pixel output to the corresponding sample-and-hold channel of the readout circuit. At this time, every pixel output in the row is transferred at the same time. Then the readout circuit digitizes and transmits all pixel signals in a row sequentially to the memory of a main computer. This three-phase procedure is repeated row by row until all pixel signals are read out. All the test pixels were designed to have a size of $50 \times 50 \mu m^2$. 
Table 1. Design parameters of the three test pixels and specifications for the fabrication process.

<table>
<thead>
<tr>
<th></th>
<th>Type A</th>
<th>Type B</th>
<th>Type C</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Original function of n-layer</strong></td>
<td>n− for pinned photodiode</td>
<td>n-well for logic</td>
<td>n+ implant for logic</td>
</tr>
<tr>
<td><strong>Doping concentration (cm⁻³)</strong></td>
<td>~ 10¹⁶</td>
<td>~ 10¹⁷</td>
<td>~ 10²⁰</td>
</tr>
<tr>
<td><strong>Junction depth (µm)</strong></td>
<td>~ 0.7</td>
<td>~ 1.3</td>
<td>~ 0.5</td>
</tr>
<tr>
<td><strong>Epitaxial layer</strong></td>
<td>p-type, resistivity: 30Ω·cm, thickness: 5.5 µm</td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Pixel structure</strong></td>
<td>3-transistor active pixel sensor</td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Fill factor (%)</strong></td>
<td>80</td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Pixel size (µm²)</strong></td>
<td>50 × 50</td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Pixel sub-array size</strong></td>
<td>40 × 40</td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Technology</strong></td>
<td>0.18 µm 1P4M CMOS Image Sensor process</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

2.2 Fabrication of a test chip

A test chip was fabricated in the 0.18 µm 1P4M CMOS process which is specialized in image sensor technology. The test chip is composed of an 80 × 80 pixel array, row/column addressing blocks and an output buffer. The pixel array is subdivided into four 40 × 40 sub-arrays, however, only three sub-arrays are used for this study. A chip layout and a packaged test chip on a printed circuit board is depicted in figure 2(a) and (b), respectively. In the fabrication, various schemes including thin back-end fabrication process, passivation planarization and anti-reflection coating as well as the lightly-doped epitaxial wafer are allowed without additional cost for the purpose of maximizing the image sensor performance. By using the passivation planarization and the thin back-end techniques, the total thickness of the back-end layers could be reduced to ~4.5 µm (compared with a thickness of ~7 µm in a standard logic process) minimizing the photon loss due to the oxide and passivation above the photodiode active area. The design parameters of the three test pixels and specifications for the fabrication process are summarized in table 1.

To measure the sensitivities of the test pixels under practical mammographic X-ray beam conditions, two screen type scintillators, LF and LR, were combined with the test chip using BC-630 optical grease as shown in figure 2(c) and (d). Both screens are frequently used in X-ray radiography and are composed of Gd₂O₂S:Tb which generates visible light photons with wavelength range between 380 nm and 600 nm and with the peak intensity at 532 nm [14]. The mass thickness of LF is 34 mg/cm² (equivalent thickness: 46 µm) and that of LR is 67 mg/cm² (equivalent thickness: 91 µm), so LF is better for the high-resolution application and LR is for the high-sensitivity application. The average particle size of these two screens is approximately 5 µm, and the physical density is 7.34 g/cm³. BC-630 has a very flat transmission of about 95% for wavelengths between 280 nm and 700 nm that covers the entire spectrum of Gd₂O₂S:Tb.

2.3 Performance evaluation methods

In this study, we measured the CVCG, the QE and the DC to evaluate the optical properties of the designed pixels. The CVCG of the pixel was estimated by means of the photon transfer method which is based on Poisson statistics of photon shot noise [15].
independent experiments, pixel output variance was plotted with respect to averaged effective pixel output voltage, and consequently the CVCG could be obtained in $\mu V/e^-$ from the slope of the graph. With the measured CVCG, the pixel QE was calculated for a given wavelength, $\lambda$, using the following equation:

$$QE(\lambda) = \frac{N_{SIG}}{N_{PH}(\lambda)} = \frac{1}{N_{PH}(\lambda)} \cdot \frac{V_{SIG}[V] \times 10^6}{\text{CVCG}[V/e^-]} \times 100 \% \quad (2.1)$$

where $N_{SIG}$ is the number of generated signal charges in the photodiode and $V_{SIG}$ is the amount of output voltage change corresponding to $N_{SIG}$ at the end of the integration period, $T$. $N_{PH}(\lambda)$ is the number of incident photons on the pixel and can be computed by

$$N_{PH}(\lambda) = \frac{I_0 \times T \times A}{E_{PH}(\lambda)} \quad (e^-) \quad (2.2)$$

where $I_0$ is the intensity of incident visible light in W/$\mu m^2$ and was measured using a reference photodiode and a calibrated optical power-meter, Thorlabs S120C and PM100D, respectively. $E_{PH}(\lambda)$ is the energy of the incident photon for the corresponding wavelength in joules (J). $A$ is the area of the pixel in $\mu m^2$.

The DC was estimated from the dark rate which refers to the number of generated dark electrons per second for each pixel. To measure the dark rate, two dark values were taken at the end of two different integration periods, and then the difference of two dark levels were divided by the difference of two integration times, the CVCG and the photodiode area. In turn, the DC could be calculated in pA/cm$^2$ by multiplying the dark rate and the elementary charge, $1.602 \times 10^{-19} \text{C}$.

On the other hand, the X-ray sensitivity of the photodiode-Lanex combination is defined, in this paper, as the ratio of the pixel output voltage to the incident X-ray exposure. Varying the tube current of the X-ray source at a fixed integration period, or the X-ray exposure, the pixel outputs were recorded and plotted at each exposure level. After all, we could get the X-ray sensitivity in mV/mR from the slope of the graph. To make this experiment suitable for mammography conditions, an X-ray source (Varian Rad70-B) having a Mo/Mo target/filter combination and a 300 $\mu m$ focal spot was used. At this focal spot size, the source could be controlled to have the peak voltage of 32 kVp and the maximum tube current of 160 mA. Each incident X-ray exposure was measured using the ionization chamber, RAD CHEK$^{TM}$ PLUS, at the entrance window of the test chip.

To perform all the experiments accurately and promptly, we developed a data acquisition system (DAS) that is divided into three parts: a sensor board, an interface board and a graphic user interface (GUI) software. The interface board included the sensor board, a 16-bit analog-to-digital converter (ADC), a field-programmable gate array (FPGA), voltage regulators and bias circuits. The interface board played the role of controlling the test chip and the ADC through the FPGA, communicating with the GUI software by means of an NI PCI-6254 board. The sensor board installed on top of the interface board is presented in figure 3(a) and the GUI software programmed by using NI LabVIEW is shown in figure 3(b).

During the experiments, supply voltage for the pixels (VDDA in figure 1(a)) was set to 3.3 V, and thus the reset voltage of the $D_\text{node}$ was 2.5 V. Bias current for each pixel was adjusted to 850 nA and temperature was regulated to 25$^\circ$C.
3 Results and discussion

3.1 Optical properties

First of all, the CVCG of each pixel was measured by employing the aforementioned photon transfer method. As shown in figure 4(a), (b) and (c), the obtained CVCGs are 0.86, 0.53 and 0.93 $\mu$V/e$^-$ for Type A, Type B and Type C, respectively. Since the small signal gain of the in-pixel source follower is approximately 0.8 V/V, the CVCGs of the D$_{\text{node}}$ in figure 1(a) can be estimated to 1.08, 0.66 and 1.16 $\mu$V/e$^-$, respectively.

Once the CVCG is known, it is possible to calculate the D$_{\text{node}}$ capacitance by using the reciprocal of the CVCG. The calculated D$_{\text{node}}$ capacitances are 148 fF for Type A, 242 fF for Type B and 138 fF for Type C. The capacitance of Type C is smaller than the other two photodiodes, although the n$^+$ junction has the highest doping concentration. As far as the n-well and the n$^+$ junctions are concerned, the periphery junction capacitance would be the main cause of this result. The doping level of the epitaxial layer is very low compared with those of the n$^+$ and the n-well layers, so that the depletion width of the n$^+$ junction is comparable to that of the n-well junction. Therefore, assuming that the area capacitances of both junctions are the same, we can conclude that the capacitance of the n-well junction which is deeper than the n$^+$ junction is higher on account of the periphery capacitance. On the other hand, the doping concentration and the depth of the n$^-$ layer are lower than those of the n-well layer, so the area capacitance of the n$^-$ junction is lower than that of the n$^+$ and n-well junctions while the periphery capacitance is higher than that of the n$^+$ junction.

Taking the above outcomes into account, we measured the QE of the three photodiodes for the range of 420 nm to 700 nm wavelength and presented the results in figure 5. The graph shows that Type A and Type B have almost the same characteristics while the Type C has lower QE than those two photodiodes on a whole range of wavelengths. Especially, the shorter the wavelength is, the poorer the QE of Type C is. As the n$^+$ layer is doped with high concentration, the diffusion length of minority carriers is very short, so generated carriers in the n$^+$ neutral region can be easily recombined. Hence, the QE of the n$^+$/p-epi photodiode is lower and this phenomenon is more prominent in the region of short wavelengths because of their short absorption lengths. On the
Figure 4. Photon transfer curves for (a) Type A, (b) Type B and (c) Type C structures.

Figure 5. Quantum efficiency of the photodiode for the range of 420 nm to 700 nm wavelength.

contrary, the life time of carriers generated in the low-doped epitaxial layer is long enough to be brought to the depletion region and contribute the signal output. Therefore, the difference between the QEs is narrowed in the region of long wavelengths.
Finally, to estimate the DC, we measured the output voltage under dark conditions as a function of integration time as shown in figure 6. The slopes are 31.63, 17.01 and 38.73 mV/s for Type A, Type B and Type C, respectively. Using the measured CVCGs and the photodiode area, 2,000 $\mu$m$^2$, the DC for each photodiode can be obtained: 294, 257 and 333 pA/cm$^2$, respectively. Generally, the DC is known to be mainly caused by STI stress around the junction and surface damage due to the implantation of high doping concentration [10]. Since the n$^+$ junction is formed close to STI regions and has high doping concentration, Type C generates more DC than the other two photodiodes.

3.2 Sensitivity to mammographic X-ray

Keeping the distance between the X-ray source and the test chip 65 cm, the pixel output voltage was measured as a function of entrance exposure as depicted in figure 7(a) and (b). The tube voltage of the X-ray source was 28 kVp. The same experiment was repeatedly performed for the three photodiodes coupled with the two types of scintillators: LF and LR. The maximum X-ray sensitivity is obtained from Type A coupled with LR and the identical value is 30.5 mV/mR. Type A, when compared the other two photodiodes, have relatively large CVCG and QE in the vicinity of 532 nm wavelength, and consequently, it generates the greatest number of electric charges in response to scintillation light and the voltage of the D$_{node}$ is rapidly discharged.

4 Conclusion and further study

The optical properties and the X-ray sensitivities of three different photodiodes fabricated by using a 0.18 $\mu$m 1P4M standard CIS process have been measured and compared to choose the most advantageous photodiode structure for high-resolution and high-sensitivity digital mammography. Use of the n$^-$ layer which forms a relatively shallow and lightly doped junction have many benefits for the optical properties of the photodiode. The low junction capacitance and the long carrier life time of the n$^-$/p-epi (Type A) photodiode assure comparatively high CVCG, high QE and low DC, although each property is not the best. This competitiveness also goes for the mammography conditions, as a result, the n$^-$/p-epi photodiode combined with either LF or LR exhibits the highest

---

**Figure 6.** Pixel output voltage under dark condition as a function of integration time.
Figure 7. Sensitivities to mammographic X-ray for the three structures coupled with (a) Lanex Fine and (b) Lanex Regular.

Table 2. Summary and comparison of the characteristic for the three photodiodes.

<table>
<thead>
<tr>
<th></th>
<th>Type A</th>
<th>Type B</th>
<th>Type C</th>
<th>RadEye 1</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pixel supply voltage (V)</td>
<td>3.3</td>
<td>5</td>
<td></td>
<td>5</td>
</tr>
<tr>
<td>Temperature (°C)</td>
<td>25</td>
<td>23</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pixel size (µm²)</td>
<td>50 × 50</td>
<td>48 × 48</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Fill factor (%)</td>
<td>80</td>
<td>80</td>
<td></td>
<td></td>
</tr>
<tr>
<td>CVCG (µV/e⁻)</td>
<td>0.86</td>
<td>0.53</td>
<td>0.93</td>
<td>0.51</td>
</tr>
<tr>
<td>QE (%) @ 532 nm</td>
<td>49</td>
<td>50</td>
<td>45</td>
<td>35</td>
</tr>
<tr>
<td>DC (pA/cm²)</td>
<td>294</td>
<td>257</td>
<td>333</td>
<td>&lt; 69.5</td>
</tr>
<tr>
<td>Sensitivity to X-ray (mV/mR) @ 28 kVp</td>
<td>LF 8.9</td>
<td>5.2</td>
<td>8.4</td>
<td>3.9 @ 30 kVp</td>
</tr>
<tr>
<td></td>
<td>LR 30.5</td>
<td>18.9</td>
<td>29.4</td>
<td>NA</td>
</tr>
</tbody>
</table>

X-ray sensitivity. On the other hand, the n-well/p-epi (Type B) has poor X-ray sensitivity due to the low CVCG and the n⁺/p-epi (Type C) photodiode shows the largest DC in spite of pretty good X-ray sensitivity.

The experimental results are summarized in table 2 and compared with a commercial CIS, RadEye 1, manufactured by Rad-Icon in U.S.A. in order to verify the feasibility of the designed pixels for high-sensitivity digital mammography [16, 17]. The RadEye 1 has been developed for digital X-ray imaging and has the pixel size of 48 × 48 µm² with 80 % fill factor similar to that of the pixels presented in this paper. Except for the DC, the properties of the n⁻/p-epi photodiode are superior to those of the RadEye 1. In this study, not enough effort was put into reducing the DC, so this comparison definitely provides directions on the future work for further reduction of the DC.

Acknowledgments

This work was supported by Nuclear Research & Development Program of the Korea Science and Engineering Foundation (KOSEF) grant funded by the Korean government (MEST). (grant code: N01110445)
References


