

Introduction to the MOSFET Dosimeter

Introduction

MOSFET dosimeters are now used as clinical dosimeters for radiotherapy radiation beams.¹ The main advantages of MOSFET devices are:

1. The MOSFET dosimeter is direct reading with a very thin active area (less than 2 μm).
2. The physical size of the MOSFET when packaged is less than 4 mm².
3. The post radiation signal is permanently stored and is dose rate independent.²

This paper reviews the basic MOSFET structure and how it is used as a dosimeter.

The MOSFET Dosimeter

The basic MOSFET structure is depicted in Figure 1. The type shown is a P channel enhancement MOSFET which is built on a negatively doped (n-type) silicon. When a sufficiently large negative voltage is applied to the polysilicon gate a significant number of minority carriers (holes) will be attracted to the oxide/silicon surface from both the bulk silicon substrate and the source and drain regions.

Once a sufficient concentration of holes have accumulated there, a conduction channel is formed, allowing current to flow between the source and drain (I_{ds}). The voltage necessary to initiate current flow is known as the device threshold voltage (V_{TH}).

When a MOSFET device is irradiated, three things happen within the silicon dioxide layer (sensitive region): a build-up of trapped charge in the oxide; the increase in the number of interface traps; and the increase in the number of bulk oxide traps.

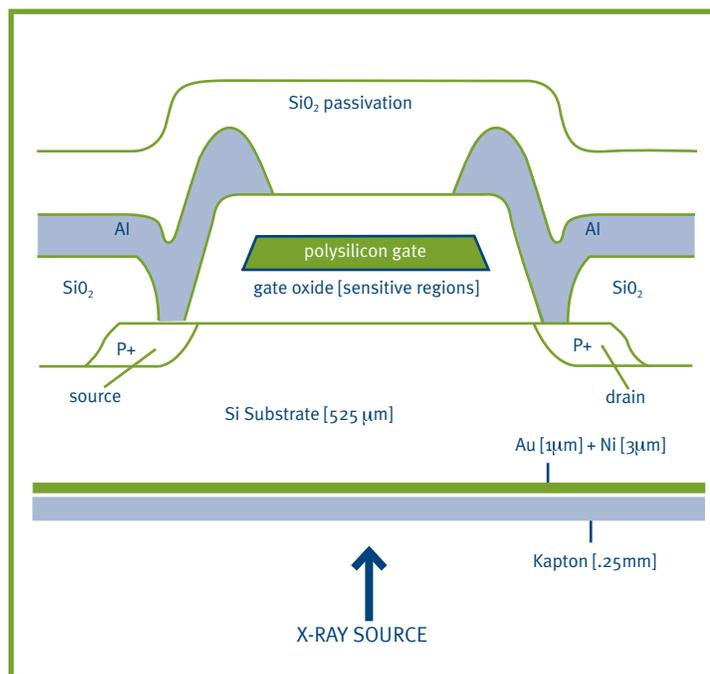


FIGURE 1. Schematic cross section of a P-channel MOSFET

Electron-hole pairs are generated within the silicon dioxide by the incident radiation. Electrons, whose mobility in SiO₂ at room temperature is about 4 orders of magnitude greater than holes, quickly move out of the gate electrode while holes move in a stochastic fashion towards the Si/SiO₂ interface where they become trapped in long term sites, causing a negative threshold voltage shift (ΔV_{TH}), which can persist for years.

The difference in voltage shift before and after exposure can be measured, and is proportional to dose.

At energies below 50 keV photoelectric absorption is the dominant process while between 60 keV and 90 keV both photoelectric and Compton scattering become important. For both of these mechanisms the energy deposited to the oxide (sensitive region) of the MOSFET is delivered mostly by secondary electrons.

The photoelectric cross-section depends strongly on the target material, approximately Z^4 . The result of this is that interface dose enhancement occurs because the photoelectric cross section is different in the Si than in the SiO₂. More electrons will be produced in the Si, the higher cross section material, so more will cross from Si into SiO₂ than will cross the interface in the other direction. Dose enhancement and the increasing dominance of photoelectric absorption, as the incident energy decreases from 150 keV to 10 keV, results in an increase in device sensitivity. This energy dependence is typical of most dosimeters.

The range of photon energies used in fluoroscopy is narrow and the observed MOSFET energy dependence is 10%. This is small compared to other errors inherent in skin dose measurements, making it possible to use the dosimeter in the fluorographic energy range 40kVp to 140kVp.

Oncology procedures consist of irradiating an immobilized patient with a high energy photon (1 to 20 MeV) or electron beam (6 to 20 MeV). In these energy ranges the observed energy dependence does not exceed 5% and the MOSFETs respond identically to electrons and photons.^{3,4}

Conclusions

The physics of MOSFET dosimeters is now well understood so that these devices can be characterized and used for many applications in radiation dosimetry.

Small size and radio-transparency make the device an excellent choice for dosimetry in fluorography.

Low energy dependence, high sensitivity and immediate read out make the MOSFET a good replacement for TLD in radiation therapy dosimetry.

References

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