A study of a triple GEM detector as real time dosimeter in external photon beam radiotherapy

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Acknowledgments
Introduction

Radiotherapy with external photon or electron beams represents one of the most widespread methods in the treatment of cancer disease. Actually new methods based on other particles are increasing. Among the most used particles there are protons, carbon ions and neutrons. All these techniques goes under the name of hadrontherapy which represents the frontier in the treatment of cancer. Despite this, radiotherapy is not outdated because the new techniques can be considered as a supplement to conventional radiotherapy for some types of tumors. Recently the irradiation techniques have reached an accuracy of few mm. Intensity-Modulated-Radiation-Therapy (IMRT) is one of the most advanced irradiation techniques with external X-ray beams. It allows to conform in more detail the beam to the tumor volume adjusting a set of programmable multi-leaf collimator which can assume very different shapes. Then if there is the possibility to obtain very accurate fields, there must be also an instrument able to verify the real released doses. A typically radiotherapy treatment session follows a precise protocol made of a sequence of steps: data gathering and medical consulting, target identification and prescription of doses, treatment planning calculation, pre-treatment quality assurance, on-treatment dose monitoring and post-treatment analysis. Among these pre-treatment quality assurance play a central role because guarantees the success of a radiotherapy treatment. The devices devoted to the quality assurance are dosimeter with different forms and dimensions. In particular an important category is represented by 2D device which are able to provide a measure of the dose distribution on a given area. The most common are ionization chamber arrays, films and Electronic Portal Imaging Devices (EPID). In the following work, triple-GEM detectors are presented as valid alternatives to 2D dosimeters for external beams radiotherapy. These type of detector shows some interesting characteristic which can overcome the typical drawbacks observed in the traditional 2D dosimeters. Actually the ionization chamber arrays are the most widely used. They can provide a dose measure point by point with an electronic read-out. However their spatial resolution is limited to few mm and even if this is an optimal value for great uniform fields, it cannot provide correct measure on area with high dose gradient. Film dosimeters, on the contrary, can provide an higher spatial resolution with the drawback that it must be processed after taking into account other parameters which can influence the correct dose measure like fading, UV sensitivity, temperature and so on. A possible alternative is represented by EPIDs, even if they have been not conceived as dosimeters. They are made of matrix of silicon sensors which can have an area minor than 1 mm and are used for high energy X-rays radiographies to set up the patient in the correct position. Then in order to convert the signal registered by an EPID into dose values, an algorithm software is needed to reconstruct the real dose distributions. This work will demonstrate that detectors based on the Gas Electron Multiplier (GEM) technology can be a valid alternative for 2D dosimetry respect to the traditional devices. In particular the last presented prototype, called “GEMpix”, can perform 2D dose distribution measurements with a spatial resolution even higher than the gafchromic films and in real time acquisitions. All the works is presented in four chapters.

Chapter 1 treats the main elements of a radiotherapy treatment unit with particular emphasis to the particular used sources (linacs) and to the irradiation techniques with external photon beams (standard and IMRT). The main mechanism of photon-matter interaction are resumed to introduce some important dosimetric concepts like kerma, absorbed dose and charge particle equilibrium. At
the end, a radiotherapy treatment unit is presented from a physical point of view and the concepts of absolute and relative dosimetry are introduced together with the gamma index acceptance test.

Chapter 2 introduces the widespread cavities theories and the theoretical definition of a dosimeter. Then the main characteristic parameter of a dosimeter are introduced and a brief description of the main 1D and 2D dosimeters is presented with particular attention to film dosimeter which represents the reference device for the new proposed detectors.

Chapter 3 presents the triple-GEM detector in details and the problem of his use as a 2D dosimeter will be discussed. At first some preliminary measurements performed with a usual triple-GEM detector are presented. These will provide important guidelines for the construction of the new GEM detector which will be used in scanning mode. At the end some measure with GEM detector are compared with those obtained with gafchromic EBTs applying the gamma index acceptance test and some important information are derived.

Chapter 4 introduce the technology of Timepix Chip with the description of its main working parameters. A description of the new prototype GEMpix is presented with some simulation of expected results. At the end a series of measures made on irradiating different fields will be analyzed comparing them to those obtained with gafchromic EBTs. It will be shown how working on some characteristic parameters of the GEMpix detector. The obtained dose distributions are in optimal accordance with analogous measures performed with gafchromics. This will be demonstrated also for some IMRT fields.

The following work will demonstrate that GEM technology with an appropriate read-out electronic able to provide a measure of the charge may represent a valid alternative to the actual devices used for 2D dosimetry in radiotherapy. It can work in real time and at a spatial resolution and a sensitivity higher than films. Acquisitions in real time mode are much faster than actual electronic devices with frame rate that can reach the pulse frequency of a medical Linac accelerator (hundreds of Hz).
Chapter 1

Introduction

More than 2/3 of cancer diseases are treated with radiotherapy [1]. It is a widespread technique that is used for treating the tumor itself or matching to other treatments (like chemotherapy, for example). Often it is used also as a palliative treatment. The following chapter will presents the main elements of a typical radiotherapy equipment especially from physical point of view. It begins describing the medical Linac which is the most diffused X-ray source applied in radiotherapy: more than 80 % of radiation sources are accelerators. Then a brief description of the principal interaction mechanisms of high energy X-rays with matter follows with a particular attention to the Compton scattering which represents the main interaction for the typical energy spectra applied in radiotherapy treatments. These basic concepts will allow to define some important dosimetric quantities like kerma and dose. The second part of the Chapter introduces the main elements which constitutes a radiotherapy unit and the typical steps followed in a radiotherapy treatment session. These arguments will frame the role played by dosimetry and in particular by 2D dosimeters. The last paragraph is devoted to the definition of gamma index acceptance test which is the main comparison method between two dose distributions and that will be widely used in the present work.

1.1 External beam radiotherapy

Actually radiotherapy techniques can be distinguished in external and internal. The most important ones belonging to the second group there are Brachytherapy and Intra-Operative Radio Therapy (IORT). In the first case the radiation source is placed inside or next to the area requiring treatment. In the second case, the target area is irradiated when it is exposed during surgery. External Beam Radio Therapy (EBRT) or Teletherapy, instead, uses external sources which irradiate the patient in order to deliver the dose to the target volume [2,3]. This technique is the most common and can use particles like photons, electrons and, recently, also protons, heavy ions and neutrons. This work is focalized on photons, in particular, those produced by medical LINear ACcelerators (LINACs), a widespread device in many radiotherapy centers. A traditional EBRT treatment with photons is performed irradiating the tumor target with a series of consecutive fields. Once the target volume (defined Planning Target Volume, PTV) has been identified, the amount of dose to deliver to the target and dose tolerance to the neighboring healthy organs (defined Organs At Risk, OAR) is defined by the oncologist. How these level of doses are realized on the target volume is a task of the medical physicists. They dispose of a Treatment Planning System (TPS) control unit by which it is possible planning the treatment. It permits to set different beam configurations with different irradiating directions. At the same time, a calculation on the PTV and OAR is produced so that the operator can change the beams setting until the required dose are reached within the allowed errors.
like physiological organs motion, patient positioning and so on. Standard radiotherapy follows this scheme and the irradiation is executed rotating the accelerator gantry around the patient and shaping the beam sent on the target (fig. 1.1).

**Figure 1.1**: left: patient layout during a treatment session; right: example of standard radiotherapy with four external fields for a prostate treatment [4].

The beams available in the standard external radiotherapy have square, rectangular or other simple geometries. In addition some wedged filters can be used to modulate photon fluence. Usually these are made of lead or other materials. Actually a new technique allows to obtain more accurate dose distribution with best conforming geometries on the target volume and a reduced dose on the OAR. This is known as IMRT radiotherapy and will be discussed in par. 1.1.2. First of all, it is important to describe the irradiating machine: the medical LINAC.

1.1.1 Medical LINACs

X rays used in radiation therapy are produced by a linear accelerator (LINAC) through electron bremsstrahlung. An electron beam is accelerated by MV potential and is directed on a metallic target. High-Z materials are used because they convert a larger fraction of the electron energy into bremsstrahlung X-rays [5]. Tungsten (Z = 74) is a common choice as a target. The block diagram illustrates the main elements of a medical LINAC and allows for a brief description of its working operation.

**Figure 1.2**: schematic layout of a medical linac accelerator [3].
A power supply provides a continuous power to a pulsed modulator which consists of a pulse-forming-network (PFN) and a switch tube, known as “thyatron”. Modulator, by means of PFN, produces high frequency rectangular short pulses (few microseconds) which are sent either to the microwave source (klystron or magnetron) or to the electron generation circuit (“electron gun”). Microwave are sent, through a wave guide, to the accelerating tube. Electrons produced by the electron gun are injected in phase with the electromagnetic field created in the accelerating tube. In these conditions, due to the principle of phase stability [6,7,8], electrons will cluster in bunches and are accelerated. Usually the accelerating tube is made of copper and is divided in a sequence of cavities separated by copper irises (multi-cells structure). Energy provided to the bunches can be delivered through stationary waves or traveling waves. In the first case, the accelerating structure must be terminated by a reflecting system in order to obtain stationary waves. In the second case, it is necessary to have a terminator which absorbs the travelling wave avoiding the reflection wave. In addition, there are some coils around the tube (“focusing coils”) and at its ends (“steering coils”). Focusing coils have the function of preserving the compactness of electron bunches which can expand, due to electrostatic repulsion. Steering coils are used to center the electron beam on the axis of the tube. Coil magnetic fields are created by preset currents which can be varied by appropriate retroactive circuits which receive a signal from a monitor ionization chamber in the acceleration head. Electrons coming from the accelerating tube typically appear as a narrow beam of few mm in diameter. A set of magnet in the accelerator head has the function of bending the electron trajectory (“bending magnet”) and focusing the electron beam (“focusing magnet”). The bend angle can have different values depending on the accelerator. X-ray are produced sending this electron beam on a target. A fraction of the electron energy is converted in heat, the other produces bremsstrahlung photons. Because the photon fluence angular distribution is strongly tilted at the center, a flattening filter is used in order to obtain a uniform fluence. Typically, it consists of a small steel or copper cone. Its effect is illustrated in fig. 1.3.

This explains also the role of steering coils: if the electron beam is not centered, it can hit the target in a different position producing an highly asymmetric beam after the flattening filter. It is possible to use directly also the electron beam. In this case there is no target and the electron beam is expanded through a diffuser sheet which typically is made of copper. Diffuser sheets and flattening filters are located on a rotating carousel which can change the various elements depending on the used beam and on its energy. Immediately below the carousel, there are the two ionization monitor chambers. They are calibrated in Monitor Units (MU). These units are defined so that the monitor chamber reads 100 MU when an absorbed dose of 1 Gray is delivered by a 10x10 cm² field to a
point at a given depth in a phantom and with the surface of the phantom positioned so that the specified point is at the isocenter of the machine. For the Synergy Linac of this work, a water phantom is used and its surface is at 95 cm from the source, while the measuring point is at the isocenter 5 cm in depth. The energy spectrum is that corresponding to 6 MV. Beyond the monitor ionization chambers, there is a secondary collimation system made of two couples of blocks, usually made of tungsten. It allows to obtain fields of different dimensions, but they are limited to square or rectangular fields. Modern Linacs are equipped also with a multileaf collimator which allows to define a specific field shape. Technology of multileafs is the basis of the new irradiating technique, known as IMRT radiotherapy. Elekta Synergy Linac is based only on a multileaf collimator without secondary collimator. Each leaf is made of tungsten, has a height of about 7 cm (along the beam trajectory) and a thickness of 3 mm. It is a modern accelerator which can perform IMRT treatments. Like all the other medical linacs, it has a photon beam with a pulsed temporal structure (fig. 1.4).

![Figure 1.4: temporal structure of the pulsed beam coming from the Elekta Synergy LINAC working at a dose rate of 100 MU/min.](image)

Each pulse last for 3 µs and the number of pulses per second, known as the “Pulse Repetition Frequency (PRF)”, can assume only 7 values: 6, 12, 25, 50, 100, 200 and 400 Hz. The PRF value is numerical equal to the delivered MU/min.

### 1.1.2 Intensity Modulated Radiation Therapy

Intensity-modulated radiation therapy (IMRT) is an advanced mode of high-precision radiotherapy [9] that uses computer-controlled linear accelerators to deliver precise radiation doses to a malignant tumor or specific areas within the tumor. IMRT allows for the radiation dose to conform more precisely to the three-dimensional (3-D) shape of the tumor by modulating or controlling the intensity of the radiation beam in multiple small volumes. Each IMRT field consists of a sequence of conformal small field, called segments, which overlap in order to realized the prescribed dose distribution. A schematic example of an IMRT field is shown in the fig. 1.4.

![Figure 1.4: schematic representation of a sequence of 3 IMRT segments and the relative delivered dose distribution [4].](image)
Irradiation can be released in two different modes: step and shoot or continuous. In the first one the accelerator deliver a segment and remain in that configuration once it has finished. Then it change configuration and irradiates the next segment. It continues in this way until all segments have been delivered, covering the prescribed dose distribution on the target. In the second mode of irradiation, accelerator moves continuously together with the multileaf collimator which is set on the configuration of the segment delivered and that time. Treatment is carefully planned by using 3-D computed tomography (CT) or magnetic resonance (MRI) images of the patient in conjunction with computerized dose calculations to determine the dose intensity pattern that will best conform to the tumor shape. Typically, combinations of multiple intensity-modulated fields coming from different beam directions produce a custom tailored radiation dose that maximizes tumor dose (PTV) while also minimizing the dose to adjacent normal tissues (OAR). As the ratio of normal tissue dose to tumor dose is reduced to a minimum by means of IMRT approach (fig. 1.4), higher and more effective radiation doses can safely be delivered to tumors with fewer side effects compared with conventional radiotherapy techniques. IMRT also has the potential to reduce treatment toxicity, even when delivered doses are increased. Due to its complexity, IMRT does require slightly longer daily treatment times and additional planning and safety checks before the patient can start the treatment than conventional radiotherapy. Currently, IMRT is being used most extensively to treat cancers of the prostate, head and neck, and central nervous system. IMRT has also been used in limited situations to treat breast, thyroid, lung, as well as in gastrointestinal, gynecologic malignancies and certain types of sarcomas. IMRT may also be beneficial for treating pediatric malignancies.

1.2 Photon interactions with matter

For the topics covered in this thesis, it is important to consider the following interaction mechanisms of X and γ ray photons in matter [10,11]:

- Compton effect
- Photoelectric effect
- Pair production
- Rayleigh (coherent) scattering
- Photonuclear interactions

The first three are the most important for dosimetry, because these interactions produce secondary electrons which release energy inside matter. In Rayleigh scattering, a photon is merely redirected through a small angle with no energy loss. Photonuclear interactions are significant for photon energies above a few MeV and can create radiation protection problems through the (γ,n) production of neutrons. The relative importance of the first three interactions is related to the atomic number Z of the material and to the photon quantum energy \(E_\gamma = h\nu\). Fig. 1.6 shows the regions of Z and \(E_\gamma\) in which each interaction predominates.
Curves indicate where two kinds of interactions have the same probability, in terms of cross sections. It is clear that for low Z media (e.g., carbon, air, water, human tissue) the region of Compton-effect is very broad and extends from \( \approx 20 \text{ keV} \) to \( \approx 30 \text{ MeV} \). For higher Z values this region is narrow.

### 1.2.1 Compton scattering

Compton interaction can be described from two points of view: kinematics and probabilistic [10]. The first relates energies and angles of the participating angles, the second describes the cross sections that a Compton interaction has in matter. In the following calculations, it is assumed that the electron struck by the incoming photon is initially unbound and stationary. Even if these assumptions do not represent real conditions, the resulting errors remain negligible for radiological physics applications.

\[ E_{\gamma} = h\nu \]  
\[ \text{mom.} = h\nu/c \]

\[ E'_{\gamma} = h\nu' \]  
\[ \text{mom.} = h\nu'/c \]

\[ k.e = T \]  
\[ \text{mom.} = p \]

\[ \phi \]
\[ \theta \]

Applying energy and momentum conservation laws, it is possible obtain the following three simultaneous equations in five parameters: \( h\nu, h\nu', T, \theta \) and \( \phi \):

\[
h\nu' = \frac{h\nu}{1+(h\nu/m_0c^2)(1-cos\phi)} \tag{1.1}
\]
\[ T = h\nu - h\nu' \quad (1.2) \]
\[ \cot\theta = \left(1 + \frac{h\nu}{m_0c^2}\right)\tan\left(\frac{\theta}{2}\right) \quad (1.3) \]

in which \(m_0c^2\) (the rest energy of the electron) is 0.511 MeV, and \(h\nu, h\nu'\) and \(T\) are expressed in MeV. Previous relations tell nothing about the probability of a photon or an electron being scattered in a particular direction. The differential cross section was derived by Klein and Nishina applying the Dirac’s relativistic theory of electron to the Compton effect. According to their calculations, the differential cross section for photon scattering at angle \(\varphi\), per unit solid angle and per electron, may be written as

\[
d\sigma \over d\Omega_{\varphi} = \frac{r_0^2}{2} \left(\frac{h\nu'}{h\nu}\right)^2 \left(\frac{h\nu'}{h\nu} + \frac{h\nu'}{h\nu} \cdot \sin^2 \varphi\right) \quad (1.4)
\]

in which \(h\nu'\) is given by eq. 1.1. The total K-N cross section per electron \(e\sigma\) can be gotten from an integration of eq. 1.4 over all photon scattering angles \(\varphi\):

\[ e\sigma = 2\pi \int_0^{\pi} \frac{d\sigma}{d\Omega_{\varphi}} \sin \varphi \, d\varphi \]

\(e\sigma\) is independent of the atomic number \(Z\), since the electron binding energy has been assumed to be zero, so the cross section per atom of any \(Z\) is given by

\[ e\sigma = Z \cdot e\sigma \quad (cm^2/\text{atom}) \]

while the total cross section per unit mass, \(\sigma/\rho\), which is also called the Compton mass attenuation coefficient is given by

\[ \frac{\sigma}{\rho} = \frac{N_AZ}{A} \cdot e\sigma \quad (cm^2/\text{gr}) \]

where \(N_A = 6.022 \cdot 10^{23} \text{ mole}^{-1}\) is the Avogadro’s constant, \(Z\) = number of electrons per atom (or molecule) of an element (or compound), \(A\) = number of grams per mole of material, \(\rho\) = density of the material in gr/cm\(^3\) and \(N_AZ/A\) = number of electrons per gram of material. In radiological physics, it is also important to know the overall fraction of \(h\nu\) that is given to the electrons, averaged over all scattering angles. As described later, this energy contributes to the kerma and thence to the dose. For this purpose, the quantity \(T/h\nu\) must be calculated, where \(T\) is the average kinetic energy of the recoiling electrons. Multiplying the differential K-N cross section by \(T/h\nu\), it is obtained a new quantity called “differential K-N energy-transfer cross section, \(d_e\sigma_{tr}/d\Omega_{\varphi}\) :

\[
\frac{d_e\sigma_{tr}}{d\Omega_{\varphi}} = \frac{d_e\sigma}{d\Omega_{\varphi}} \cdot \frac{T}{h\nu} = \frac{d_e\sigma}{d\Omega_{\varphi}} \cdot \frac{h\nu - h\nu'}{h\nu}
\]

Integrating over all scattering angles \(\varphi\) from 0 to 180°, it is obtained the K-N energy-transfer cross section \(\sigma_{tr}\). So the average fraction of the incident photon’s energy given to electron is given by
\begin{equation} \frac{T}{h \nu} = \frac{e \sigma_{tr}}{e \sigma} \end{equation}

A plot of this fraction as a function of the incident photon energy $h \nu$ is reported in fig. 1.7.

\begin{itemize}
  \item **Figure 1.7:** fraction of incident energy retained by photon and absorbed by electron vs energy of incident photon.
\end{itemize}

It can be noticed that at low energies the average fraction of $h \nu$ given to the electron approaches zero and then begins to rise. A useful form of the differential K-N cross section is $d_e \sigma / d \Omega_\theta$, the differential K-N cross section for electron scattering at angle $\theta$, per unit solid angle and per electron. It is related to the differential cross section $d_e \sigma / d \Omega_\varphi$ through the following relation:

\begin{equation}
\frac{d_e \sigma}{d \Omega_\theta} = \frac{d_e \sigma}{d \Omega_\varphi} \cdot \frac{(1 + \alpha)^2 (1 - \cos \varphi)^2}{\cos^3 \theta}
\end{equation}

where $\alpha = h \nu / m_0 c^2$ and $\varphi = 2 \tan^{-1}[(\cot \theta)/(1 + \alpha)]$ from equation (3). Fig. 1.8 displays the previous equation graphically for several values of $h \nu$, plotting $d_e \sigma / d \Omega_\varphi$ vs $\theta$.

\begin{itemize}
  \item **Figure 1.8:** differential Klein-Nishina cross section vs recoil angle of electron.
\end{itemize}

Another additional form of the differential K-N cross section is $d_e \sigma / dT$, expressed in cm$^2$ MeV$^{-1}$ e$^-$ and indicating a Compton interaction in which a photon, traversing a layer containing one e/cm$^2$, transfers to that electron a kinetic energy between $T$ and $T + dT$. In this respect, $d_e \sigma / dT$ represents the energy distribution of the electrons, averaged over all scattering angles. A graphical representation of this differential cross section is given below in fig. 1.9.
However, energy distribution are those occurring at production. In an extended medium, the spectrum of Compton-electron energies at a point is degraded because electrons loss a varying amount of their energy depending on how far they travel in the medium.

1.2.2 Photoelectric effect

At low energies, the principal mechanism of interaction is represented by photoelectric effect that is even more prominent in high-Z media [11]. The photoelectric effect occurs when a photon is absorbed by an atom and, as a consequence, one electron is emitted. The absorbed photon transfers its energy to an orbital electron, which is then dislodged and exits the atom with a kinetic energy \( k_e \) given by the relation:

\[
k_e = E_x - P
\]

where \( E_x \) is the incident photon energy and \( P \) is the electron binding energy. Photoelectric interactions are most probable when the photon energy is slightly higher than the electron binding energy, the difference being transferred to the electron as kinetic energy. This is the dominant process for X-ray absorption up to energies of about 500 KeV and for atoms of high atomic number. When \( E_x \) is higher that the electron mass (\( E_x > 511 \text{ KeV} \)), Lorentz boost causes the emitted electron to exit in a narrow cone around the incoming photon beam. The emitted photoelectron may interact with other molecules or atoms in a chain reaction until all its energy is lost; moreover it leaves a vacancy in the atom, typically in the K or L shell, which must be filled, to bring the atom back to its ground state.

Figure 1.10: a schematic representation of photoelectric effect interaction.
If the ionized atom captures a low energy electron the process is followed by the emission of an X-ray photon (see Fig. 1.11); if the X-ray photon is absorbed by an outer shell electron, the result is the emission of a secondary Auger electron (fig. 1.10). The electrons ejected from the atom pass through the surrounding matter, lose their energy and move only a relatively short distance from their original location. The photoelectric effect is also responsible for the production of characteristic transition lines in the X-ray tube.

### 1.2.3 Pair production

The range of high energies is interested principally by the pair production interaction [11]. Photons must have an energy major than 1.022 MeV. In this process the photon interaction happens mainly in the nuclear coulomb field. As a result an electron-positron couple is created:

$$\gamma \to e^- + e^+ + Q$$

The exceeding energy of the photon converts in kinetic energy of the electron-positron couple. In general positrons are very short lived since they annihilate with an atomic electron, with the emission of two photons of 511 keV energy.

### 1.3 Dosimetric quantities

Three non-stochastic quantities are useful for describing the interaction of a radiation field with matter [10]: (a) the kerma $K$, describing the first step in energy dissipation by indirectly ionizing radiation, that is, energy transfer to charged particles; (b) the absorbed dose $D$, describing the energy imparted to matter by all kinds of ionizing radiations, but delivered by the charged particles; and (c) the exposure, $X$, which describes x and $\gamma$ ray fields in terms of their ability to ionize air. Only the first two will be briefly presented. The third is not useful for the arguments treated in this work.

#### 1.3.1 Kerma

Kerma is a nonstochastic quantity relevant only for fields of indirectly ionizing radiations (photons and neutrons) and for any ionizing radiation source distributed within an absorbing medium. Kerma can be defined in terms of and radiant energy $R$ and energy transferred $\varepsilon_{tr}$ and both are stochastic quantities. $R$ is defined as the energy of particles emitted transferred, or received a given volume $V$. Using this quantity, the energy transferred $\varepsilon_{tr}$ in a volume $V$ can be expressed as:

$$\varepsilon_{tr} = (R_{in})_u - (R_{out})_{u}^{nonr} + \Sigma Q \quad (1.5)$$

where $(R_{in})_u$ is the radiant energy of uncharged particles entering $V$, $(R_{out})_{u}^{nonr}$ is the radiant energy of uncharged particles leaving $V$, except those originated from radiative losses of kinetic
energy by charged particles while in $V$, and $\Sigma Q$ is the net energy derived from rest mass in $V$ ($m \to E$ positive, $E \to m$ negative). The kerma $K$ at the point of interest $P$ in $V$ is defined as

$$K = \frac{d(e_{tr})}{dm} \equiv \frac{d\epsilon_{tr}}{dm}$$

where $(\epsilon_{tr})_e$ is the expectation value of the energy transferred in the finite volume $V$ during some time interval while $d(\epsilon_{tr})_e$ is that for the infinitesimal volume $dv$ around point $P$, and $dm$ is the mass in $dv$. Thus the kerma is the expectation value of the energy transferred to charged particles per unit mass at a point of interest, including radiative-loss energy but excluding energy passed from one charged particle to another. Kerma is expressed in Gray (Gy) and 1Gy = 1 J/Kg. For monoenergetic photons kerma at a point $P$ is related to the energy fluence through the mass energy-transfer coefficient $(\mu_{tr}/\rho)_{E,Z}$, dependent on the photon energy $E$ and the atomic number $Z$ of the matter at $P$:

$$K = \psi \cdot \left( \frac{\mu_{tr}}{\rho} \right)_{E,Z}$$

where $\mu_{tr}$ is the linear energy-transfer coefficient in units of cm$^{-1}$, $\rho$ is the density in g/cm$^3$ and $\psi$ is the energy fluence at $P$ in J/m$^2$. If there is a spectrum of photon energy fluence $\psi'(E)$ at the point of interest $P$, the kerma at $P$ is calculated as:

$$K = \int_{E=0}^{E_{max}} \psi'(E) \cdot \left( \frac{\mu_{tr}}{\rho} \right)_{E,Z} dE$$

where $\psi'(E)$ is the differential distribution of photon energy fluence in units of J m$^{-2}$ keV$^{-1}$. The kerma for X and $\gamma$ rays is the energy transferred to electrons and positrons per unit mass of medium. This kinetic energy can be spent in two ways:

1. Coulomb force interactions with atomic electrons of the medium, with ionization and excitation in or near the electron track,
2. Radiative interactions with the Coulomb force field of atomic nuclei, in which bremsstrahlung X-ray photons are emitted. Compared to electrons, X-ray photons are relatively penetrating so that they carry their quantum energy far away from the charged-particle track.

In addition photons produced by in-flight positrons annihilation can be considered a type of radiative loss. Following these considerations the kerma $K$ can be subdivided in two contributions depending on whether the energy is spent in creating excitation and ionization ($K_e$) or is carried away by photons ($K_r$):

$$K = K_e + K_r$$

Collision kerma $K_e$ can be defined in an similar way like the total kerma $K$, introducing the new stochastic quantity $\epsilon_{tr}^n$ called net energy transfer and defined as

$$\epsilon_{tr}^n = (R_{in})_u - (R_{out})_{u\text{nonr}} - R^r_u + \Sigma Q = \epsilon_{tr} - R^r_u$$
where $R_u^r$ is the radiant energy emitted as radiative losses by the charged particles originated in V. Now it is possible define $K_c$ at the point of interest $P$ as

$$K_c = \frac{d\varepsilon_{ir}}{dm}$$

Thus the collision kerma is the expectation value of the net energy transferred to charged particles per unit mass at the point of interest, excluding the radiative-loss energy and energy passed from one charged particle to another. The radiative kerma $K_r$ can be written as $K_r = dR_u^r/dm$. For monoenergetic photons $K_c$ is related to the energy fluence $\psi$ by the mass energy-absorption coefficient $(\mu_{en}/\rho)_{E,Z}$, dependent on energy and type of material:

$$K_c = \psi \left( \frac{\mu_{en}}{\rho} \right)_{E,Z}$$ (1.6)

An integral calculation similar to that shown for $K$ is applied to $K_c$ when there is an energy spectrum $\psi'(E)$.

### 1.3.2 Absorbed dose

The absorbed dose $D$ is defined in terms of the related stochastic quantity energy imparted $\varepsilon$. The energy imparted by ionizing radiation to matter of mass $m$ in a finite volume $V$ is defined as

$$\varepsilon = (R_{in})_u - (R_{out})_u + (R_{in})_c - (R_{out})_c + \Sigma Q$$

where $(R_{in})_u$ and $\Sigma Q$ are defined as for eq. (5), $(R_{out})_u$ is the radiant energy of all the uncharged radiation leaving V, $(R_{in})_c$ is the radiant energy of the charged particles entering V, and $(R_{out})_c$ is the radiant energy of the charged particles leaving V. The absorbed dose $D$ at any point $P$ in $V$ is defined as

$$D = \frac{d\varepsilon}{dm}$$

where $\varepsilon$ is the expectation value of the energy imparted in the finite volume $V$ during some time interval. Thus the absorbed dose $D$ is the expectation value of the energy imparted to matter per unit mass at a point. The dimensions and units of absorbed dose are the same as those for Kerma. It is not possible to write an equation relating the absorbed dose directly to the fluence or energy fluence of a field of indirectly ionizing radiation. The absorbed dose rate at a point $P$ and time $t$ is given by

$$D = \frac{dD}{dt} = \frac{d}{dt} \left( \frac{d\varepsilon}{dm} \right)$$

### 1.3.3 Charged particle equilibrium (CPE)

Often it is useful to relate certain basic quantities. This can happen in particular conditions like radiation equilibrium (RE) and charged-particle equilibrium (CPE) [10]. In the first case, it is
possible to equate dose $D$ to the net rest mass converted to energy per unit mass at the point of interest. In the second, dose $D$ can be equated to the collision kerma. Consider an extended volume $V$ containing a distributed radioactive source. If $P$ is the internal point of interest inside volume $V$, define a smaller internal point $v$ about $P$ so that do that the maximum distance of penetration $d$ of any emitted ray and its progeny (scattered and secondary rays) is less than the minimum separation $s$ of the boundaries of $V$ and $v$ (fig. 1.11).

Figure 1.11: scheme of the Charge Particle Equilibrium for a distributed radioactive source.

It is said that radiation equilibrium exists for the volume $v$, if the following four conditions are satisfied throughout $V$:

1. the atomic composition of the medium is homogeneous,
2. the density of the medium is homogeneous,
3. the radioactive source is uniformly distributed
4. there are no magnetic and electric fields present to perturb the charged particle paths, except the fields with the individual atoms

In these conditions the expectation value of the energy imparted to matter in the volume $v$ is equal to that emitted by the radioactive material in $v$, excluding that given to neutrinos:

$$\bar{\varepsilon} = \bar{\Sigma}Q$$

where bars signify expectation values. Because the considered case is nonstochastic, the volume $v$ can be reduced to an infinitesimal $dv$ about point $P$, so that RE may be said to exist at that point. Thus it is possible to assert that when radiation equilibrium exists at a point in a medium, the absorbed dose is equal to the expectation value of the energy released by the radioactive material per unit mass at that point. Charged particle equilibrium (CPE) exists for the volume $v$ if each charged particle of a given type and energy leaving $v$ is replaced by an identical particle of the same energy entering, in terms of expectation values. Clearly if radiation equilibrium exists, so does CPE. Two cases can be distinguished: CPE for distributed radioactive sources and indirectly ionizing radiations from external sources. The second will be treated because of the importance in dosimetry with external beams. Consider again a volume $V$ containing a smaller volume $v$. Now separation between the boundaries of $v$ and $V$ is required to be equal to the maximum distance of penetration of any secondary charged particle present.
CPE exists for the volume $v$ if the following conditions are satisfied:

1. the atomic composition of the medium is homogeneous,
2. the density of the medium is homogeneous
3. there exists a uniform field of indirectly ionizing radiation; this means that rays must be negligibly attenuated by passage through matter,
4. no inhomogeneous electric and magnetic fields are present.

In this case particles are not emitted isotropically as for point sources. They are characterized by anisotropic angular distributions of secondary and scattered radiations. Nevertheless, if the medium is uniform, CPE can be satisfied for the volume $v$ and a useful relation can be obtained for the dose $D$. It is possible to demonstrate that, in CPE conditions, for an infinitesimal point $dv$ with mass $dm$, the following relation between dose $D$ and collision Kerma $K_e$ is satisfied:

$$D = K_e$$  \hspace{1cm} (1.7)

and clearly the same equality is valid for their respective time derivatives. So considering equations 1.6 and 1.7, it is possible to write the following expression for the dose rate $\dot{D}$:

$$\dot{D} = \dot{\psi} \left( \frac{\mu_{en}}{\rho} \right)_{E,Z}$$

where $\dot{\psi}$ is the energy fluence rate which, for a non-monochromatic beam, can be written as $\dot{\phi} \overline{h\nu}$, where $\dot{\phi}$ is the particle fluence rate and $\overline{h\nu}$ is the mean energy of the X-ray beam. Then the following relation can be written for dose rate:

$$\dot{D} = \dot{\phi} \cdot \overline{h\nu} \cdot \left( \frac{\mu_{en}}{\rho} \right)_{E,Z}$$  \hspace{1cm} (1.8)

Equation 1.8 allows for an estimate of the photon fluence rate per pulse coming from the Synergy Linac. According to the beam calibration discussed in par. 1.1.1, when 100 MUs are delivered through a 6 MV $10 \times 10 \text{ cm}^2$ beam to a water phantom, a dose of 1 $Gy$ 5 cm in depth must be measured.
Figure 1.13: scheme of a typical clinical set-up for the measurement of dose in a reference point in a water phantom.

Considering an ideal small volume dosimeter, this measure can be considered under CPE conditions so that equation (1.8) is valid and can be applied to make an estimate of $\dot{\phi}$. If 100 MU/min are set, the time $\Delta t$ necessary to give 1 Gy is 1 min = 60 sec. Then it is possible to write:

$$1 \text{ Gy} = D = (\Delta t \dot{D}) = \Delta t \cdot \dot{\phi} \cdot \frac{\mu_{en}}{\rho} \cdot \frac{\lambda}{W}$$

And then inverting the previous relation, $\dot{\phi}$ is calculated as:

$$\dot{\phi} = \frac{D}{(\Delta t \cdot \lambda)} \cdot \left(\frac{\mu_{en}}{\rho}\right)_{water}$$

Inserting the values $\Delta t = 60$ sec, $\lambda = 1.4$ MeV and $(\mu_{en}/\rho)_{water} = 2.9 \times 10^{-2}$ cm$^2$/gr the estimated photon fluence is $\dot{\phi} = 2.56 \times 10^9$ ph/(cm$^2$sec). 100 MU/min corresponds to a PRF of 100 Hz, so there is a pulse every 10 ms which is not spread on this time period but is concentrated in 3 µs. Then the photon fluence per pulse can be calculated multiplying the photon fluence $\dot{\phi}$ by the factor $1/(PRF \cdot 3µs)$. As a result, it is obtained a photon fluence per pulse $\dot{\phi}_{pulse} = 8.5 \times 10^{12}$ ph/(cm$^2$sec).

This estimated value will represent the reference value for all the simulation estimates which will be presented in Chapters 3 an 4. However this pulse fluence must be referred to the calibration field used in the present case (10 x 10 cm$^2$). This means that, to obtain the real number of photons per pulse sent by a given field, $\dot{\phi}_{pulse}$ must be multiplied by the area of the corresponding irradiated field. CPE is an important condition to be verified in dose measurements. Very often, however, also the condition of charge particle equilibrium (TCPE) is accepted. It is said to exist in all points in a region in which $D$ is proportional to $k_c$, with a proportionality constant of major than 1. In CPE conditions this constant is 1 (equation 7).
1.4 Patient specific Quality Assurance

In this paragraph the procedure followed in a typical treatment session will be described in order to clarify where pre-treatment dosimetry is placed along a radiotherapy route. The sequence of steps can be listed as follows:

1. Medical consultation, diagnostic data gathering;
2. Decision making on prescription dose to PTV and tolerance doses to OARs;
3. CT scans aimed to acquire data required by TPS;
4. Delineation of contours of PTV and OARs;
5. Treatment planning calculation;
6. Pre-treatment Quality Assurance (QA);
7. On-treatment in vivo dose monitoring;
8. Post-treatment analysis;

Dosimetry plays a role at point 6 and 7. 2D dosimeters, are applied mainly for pre-treatment QA controls and represents an important step for a successful radiotherapy session. In the following paragraph, a brief description of a radiotherapy control unit is presented. Then the concept of absolute and relative dosimetry are introduced according to the guidelines described in the IAEA code of practice TRS398 [12]. At the end the important gamma index acceptance tests is presented because it is widely applied on the measures performed in this work.

1.4.1 Physical aspects of a radiotherapy treatment unit

Taking into account only radiations, a typical radiotherapy treatment unit can be conceived according to the following scheme (fig. 1.43).

![Figure 1.43: schematic layout of radiotherapy control unit.](image)

It is possible distinguish four steps and for each one there are some physical quantity which can be used to describe them:

a) The source: it can be characterized in terms of activity ($^{60}$Co sources, for example) or in terms of accelerating voltages and accelerated electron currents as happens for LINACs. Another important information regards the energy spectrum of the irradiated particles.
b) Transport: typically this occurs in air. Physical quantities usually applied are flux and fluence. For photons air separating the irradiated object and the source can have some influence on the energy spectrum and in the production of secondary particles (electrons and neutrons).

c) The interaction points: that is to say all the sites where a particle interacts for the first time. Kerma, expressed in gray, takes into account this primary interaction measuring the energy acquired by the secondary charge particles after ionization.

d) Energy released in matter: at this level the required physical quantity is dose. Energetic charged particle travel in matter and undergoes to collisions with atoms releasing their energy.

1.4.2 Absolute and reference dosimetry

Absolute dosimetry is a technique that yields information directly on absorbed dose in Gy. This dosimetric measurement is also referred as calibration and is of primary importance, because, if absolute dosimetry is incorrect, everything will be wrong. In order to ensure that doses delivered to a given target is the effective prescribed dose, MU's coming from the Linac must be calibrated against a reference dosimeter in particular conditions. This is what is called “reference dosimetry”. It is generally assumed that the dose must be accurately delivered within ± 5% of the prescribed dose to ensure that treatment aims are satisfied. A reference dosimeter is calibrated in a standard national laboratory. According to the IAEA Code of Practice [12], each dosimeter must be calibrated in terms of absorbed dose to water in a $^{60}$Co beam and this is obtained placing the dosimeter is at a reference depth $z_{ref}$ in a reference beam of quality $Q_0$. It is assumed that the absorbed dose to water $D_w$ is known at a depth of 5 g/cm$^2$ in a water phantom for $^{60}$Co gamma rays. Reference dosimeter is placed in this point and its calibration factor is obtained as:

$$N_{D,w} = \frac{D_w}{M}$$

where M the dosimeter reading corrected for the influence quantities. They are not the subject of the measurement but influence the quantity under measurement. Among these the most important ones are the geometrical arrangement (distance and depth), the field size, the material and dimensions of the irradiated phantom, and the ambient temperature, pressure and relative humidity. The effect of these influence quantities can be taken into account assuming that each acts independently from the other and applying a correction factor $k_i$ for the i-th influence quantity. At the end a product of correction factors $\prod k_i$ is applied to the original value. When the beam quality is different from the reference $Q_0$, another coefficient must be considered. According to the IAEA code of practice this is treated through the beam quality correction factor $k_{Q,Q_0}$ which is not included in the previous $k_i$ factors. So, when a dosimeter is used in a beam quality $Q$ different from the reference one $Q_0$, the absorbed dose to water is given by

$$D_{w,Q} = M_Q N_{D,w,Q_0} k_{Q,Q_0}$$
Where the factor \( k_{Q,Q_0} \) takes into account the difference between the reference beam quality and the actual user quality, while the dosimeter reading \( M_Q \) has been corrected to the reference values of influence quantities. The correction factor \( K_{Q,Q_0} \) is defined as the ratio, at the qualities \( Q \) and \( Q_0 \), of the calibration factors in terms of absorbed dose to water of the ionization chamber:

\[
k_{Q,Q_0} = \frac{N_{D,w,Q}}{N_{D,w,Q_0}} = \frac{D_{w,Q}/M_Q}{D_{w,Q_0}/M_{Q_0}}
\]

The best solution is to have a direct measure of the beam quality correction factor for each chamber at the same quality of the user beam. However most standards laboratories are not able to make all these measurements. They are limited to few cases. If there are no available experimental data, the correction factors can be calculated theoretically and often the Bragg-Gray theory is applied (par. 2.1.1). All these correction factors are then used for the reference dosimetry which can be just defined as a basic output calibration of a clinical radiation beam by virtue of a direct determination of dose or dose rate in water under specific reference conditions which are defined by a precise code of practice. For the present case of radiotherapy with photons, the code of practice for high energy photon beams is followed. It applies to photon beams generated by electrons accelerated with energies ranging from 1 to 50 MeV. The most common reference beam quality \( Q_0 \) is \( ^{60}\text{Co} \) gamma rays and, usually, Primary Standard Dosimetry Laboratories (PSDLs) provide calibration factors \( N_{D,w,Q} \) based on this beam quality for a series of standard dosimeters commonly used in radiotherapy. Some PSDLs can provide also calibration factors for other beam qualities \( Q, N_{D,w,Q} \). The ratio of \( N_{D,w,Q} \) to that of \( ^{60}\text{Co} \) provides an experimental determination of the \( k_Q \) factors. In general it is preferable to have a directly measure of the \( k_Q \) value for a particular chamber. When it is not possible the calculated values for the appropriate chamber type present in the IAEA code of practice must be used.

According to the IAEA TRS398 [12], the equipment for a reference dosimetry consists of three elements:

- **Ionization chambers:** TRS398 recommends only cylindrical ionization chambers for high energy photon beams. Plane-parallel chamber can also be used if they have been calibrated in the same user beam quality. For cylindrical chamber the point of measurement is taken on the chamber axis at the center of the cavity volume.

- **Phantoms:** Typically water is recommended as the reference medium for measurement of absorbed dose and beam quality in photons beams. In addition the phantom should extent 5 cm beyond all four sides of field measurement and should extent at least 5 g/cm\(^2\) beyond the maximum depth of measurement. This type configuration is used at “Tor Vergata” radiotherapy center. It is remind to IAEA TRS398 and other specialized texts for other possible configuration.

- **Chamber sleeves:** Theses are used for non-waterproof chambers and are made of PMMA. Preferably must be not thicker than 1.0 mm. When it is not possible to use the waterproof sleeves used at standard laboratories, other sleeves of the same material and similar thickness must be used.
For high energy photon beams produced by clinical accelerators the beam quality is defined by the tissue phantom ratio $TPR_{20,10}$. It is the ratio of absorbed doses at depths of 20 and 10 cm in a water phantom at a constant Source to Chamber Distance (SCD) of 100 cm and with a field size of 10 x 10 cm$^2$ at the plane of the chamber. Other beam quality specifications can be found in Appendix III of IAEA TRS398.

### 1.4.3 Relative Dosimetry

In relative dosimetry a dose measurement is performed at non reference condition. In general no conversion coefficients or correction factors are required in relative dosimetry since it is only the comparison of two dosimeter readings. Typically two measurements are required: one in conditions where dose must be determined and the other in reference conditions where dose is known. Then the two measures are correlated to obtain the correct measured dose values. Relative dosimetry allows to evaluate many factors. Some of these are useful to characterize radiation beam: percentage depth dose (PDD), tissue maximum ratios, profiles and similar ones. Other factors are those which affect field output: field size, applicators, filters, wedges, patient specific factors and so on. The present work present a new detector for 2D dosimetry in alternative to the usual ionization chamber matrices. 2D dosimeters can also be calibrated at Standard Laboratories and used in reference conditions, but usually they are used as relative dosimeters. In the present work 2D dosimeters are used in relative mode. Typically, ionization chamber matrices are related to the true dose values calibrating them on the central chamber so that all the other chambers can provide measures which are related to the real dose value. Typically measured 2D dose distributions are then compared to the dose distributions calculated from TPS. To make this calculation, it is necessary to provide data on the material constituting the device. The standard procedure is to make a computed axial tomography (CAT) to the measuring device similar to that performed on patients. Elaborated data from CT are sent to TPS for calculation. An example of a typical procedure followed for 2D pre-treatment quality assurance (QA) is outlined in fig. 1.15 when a gafchromic film EBT is used as 2D dosimeter (see Chapter 2).
TPS provides a DICOM file which is exported via an R&V System to the radiotherapy unit where a phantom is prepared with the 2D dosimeter, in this case gafchromics (par. 2.4.1 and 2.4.2). Dose distribution acquired by gafchromics is read by a scanner in order to obtain a file which is sent to a PC for comparison with the calculated dose distribution imported on the same PC as a DICOM file. A comparison between the calculated and measured dose maps provides a precise information on the quality of the planned treatment. A widely used method for 2D dose distribution comparison is the gamma index acceptance test.

1.4.4 Gamma Index acceptance tests

One of the most widespread methods used for the comparison of two 2D dose distribution is gamma index acceptance test [15,16]. Different techniques have been developed to provide a quantitative measure of this comparison. Some of these include dose difference and distance to agreement distribution criteria, evaluated separately. Gamma index includes both in a single criterion providing a unique numerical value to evaluate dose comparison point by point. In order to illustrate this criterion, consider a 3D reference system and assume that the dose reference point is in the plane XY and identified by the vector \( r_m \), while its reference dose value \( D_m(r_m) \) is reported on the Z axis (fig. 1.15). The two criteria to consider are the following:

- **Dose-To-Agreement (DTA) criterion**: it is defined by the disk in the XY plane with center in \( r_m \) and radius \( \Delta d_M = 3 \text{ mm} \).
• Dose-Difference (DD) criterion: it is defined by the vertical line on the Z axis centered to zero and having a length of $2\Delta D_M$, with $\Delta D_M = 3\%$.

Using these two parameters ($\Delta d_M$ and $\Delta D_M$), it is possible to define an acceptance surface whose equation is the following:

$$1 = \sqrt{\frac{r^2(r_m,r)}{\Delta d_M^2} + \frac{\delta^2(r_m,r)}{\Delta D_M^2}}$$

(9)

where

$$r(r_m, r) = |r - r_m|$$

and

$$\delta(r_m, r) = D(r) - D_m(r_m)$$

is the dose difference at the position $r_m$. Equation (9) represents an ellipsoid in the 3D coordinate system introduced previously (fig. 1.15).

![Diagram showing the gamma index acceptance test](image)

**Figure 1.15**: geometrical representation of the gamma index acceptance test.

According to this criterion, comparison with a dose value $D_c(r_c)$ to be queried at the position $r_c$ is made considering the surface defined by $D_c(r_c)$: if this surface intercept the ellipsoid defined by equation (9) test passes at $r_m$. This is a general method of comparison which can be applied to more complex cases. However, more often, equation (9) is used to evaluate, point by point, a numerical factor which provides a quantitative measure of the comparison. This is known as “gamma index” and is defined as follow:

$$\gamma(r_m) = \min\{\Gamma(r_m, r_c)\} \forall\{r_c\}$$

where

$$\Gamma(r_m, r_c) = \sqrt{\frac{r^2(r_m, r_c)}{\Delta d_M^2} + \frac{\delta^2(r_m, r_c)}{\Delta D_M^2}}$$

$$r(r_m, r_c) = |r_c - r_m|$$
and

\[ \delta(r_m, r_c) = D_c(r_c) - D_m(r_m) \]

is the difference between the dose to be queried and the reference one. The value of \( \gamma \) is calculated point by point \( (r_m) \) and the comparison criterion reduces to these assessments:

- When \( \gamma(r_m) \leq 1 \), gamma index test passes
- When \( \gamma(r_m) > 1 \), gamma index test fails

Gamma index test can be used not only in the comparison between calculated and measured doses but also between two dose distributions measured by different detectors, like in the present work. In Chapter 3 and 4, it will be used for the comparison between the 2D dose distributions measured by a particular film dosimeter to those measured by GEM detectors. In this case the reference dose distribution is represented by that measured by the film, while the doses to be queried are those coming from GEM detectors.
Chapter 2

Introduction

The present chapter is devoted to dosimeters. It begins with a presentation of the cavity theories which represents the theoretical basis of dosimetry. These arguments are useful to introduce a conceptual definition of a dosimeter. Then the general characteristic parameters of a dosimeter will be introduced. Paragraph 2.3 is devoted to the main dosimeters applied in radiotherapy with special attention to the characteristic of 2D dosimeters. The last paragraph 2.4 presents a more detailed description of film dosimeters with particular attention to the EBT gafchromics (par. 2.4.2), which are the reference dosimeters used in this work.

2.1 Cavity Theories

In the following paragraph a brief description of the main cavity theories will be presented. These represent the conceptual bases on which dosimeter can be defined. The main theory is that introduced by Bragg and Gray [17,18,19]. From this several developments followed. Among the principal ones, the theories of Spencer [20] and Burlin [21] will be briefly described, because are the most widely used.

2.1.1 Bragg-Gray Theory

Measurement of dose in a medium provides insertion of a probe into the medium itself. For example, it can be a gas-filled cavity, as usually happens with ionization chamber dosimetry. The main problem is to relate the dose measured in the probe cavity to that it could be measured in the medium if the probe was not there. Bragg-Gray (B-G) theory provides a useful relation between doses measured in the two media. Consider a region of homogeneous medium $w$ which contains a layer filled with another medium $g$ and that the following two conditions are satisfied:

- the thickness of the $g$-layer is small in comparison with the range of charged particles striking it so that its presence does not perturb the charged-particle field;
- the absorbed dose in the cavity is assumed to be deposited entirely by the charged particles crossing it;

These two assumptions are known as B-G conditions and apply whether the field of charged particles enters from outside as in the case of a beam of high-energy charged particles, or is generated in medium $w$ through interactions by indirectly ionizing particles. A region satisfying the B-G conditions is also referred as “B-G cavity”. Under these two conditions, for a fluence $\Phi$ of
identical charged particles of kinetic energy $T$, the ratio of absorbed doses in the adjacent medium $w$ to that in the cavity $g$ is given by

$$\frac{D_w}{D_g} = \left(\frac{(dT/\rho dx)_{c,w}}{(dT/\rho dx)_{c,g}}\right)$$ \tag{2.3}

where $\left[(dT/\rho dx)_{c,g}\right]_T$ and $\left[(dT/\rho dx)_{c,w}\right]_T$ are the mass collision stopping powers of the two media, evaluated at energy $T$. As can be noted, assuming that the value of $\Phi$ is continuous across the interface, this ratio is not dependent on the particle fluence. Such an assumption is valid when backscattering is ignored. For a differential energy distribution $\Phi_T$ (particles/cm$^2$MeV), the appropriate average mass collision stopping powers for the two media ($m\bar{\xi}_g$ and $m\bar{\xi}_w$) must be calculated according to the following formulas:

$$m\bar{\xi}_g = \frac{1}{\Phi} \int_0^{T_{\text{max}}} \Phi_T \left(\frac{dT}{\rho dx}\right)_{c,g} \ dT = \frac{D_g}{\Phi}$$

$$m\bar{\xi}_w = \frac{1}{\Phi} \int_0^{T_{\text{max}}} \Phi_T \left(\frac{dT}{\rho dx}\right)_{c,w} \ dT = \frac{D_w}{\Phi}$$

so that the ratio of absorbed dose in $w$ to that in $g$ is given by:

$$\frac{D_w}{D_g} = \frac{m\bar{\xi}_w}{m\bar{\xi}_g} = m\bar{\xi}_w$$ \tag{2.2}

When medium $g$ is a gas in which radiation release a charge $Q$, $D_g$ can be expressed in terms of that charge as

$$D_g = \frac{Q}{m} \left(\frac{\bar{W}}{e}\right)_g$$

where $D_g$ is given in grays if $Q$ is expressed in coulombs, $m$ (the mass of the gas) in kg and $(\bar{W}/e)_g$ (the mean energy spent per unit charge produced) in J/C. When CPE exists in the neighborhood of a point of interest in a medium, the insertion of B-G cavity in that point does not perturb the “equilibrium spectrum” of charged particles.

### 2.1.2 Spencer Cavity Theory

A detailed study based on the nature of the energy spectrum was given by Spencer [20]. He considered a small cavity filled with medium $g$, surrounded by homogeneous medium $w$ that contains a homogeneous source emitting $N$ identical charged particles per gram, each with kinetic energy $T_0$. He assumed also that CPE existed, B-G conditions were satisfied and bremsstrahlung generation was absent. Under these conditions, the absorbed dose at any point in the undisturbed medium $w$ can be equaled to kerma $K_w$ and written as

$$D_w = K_w = NT_0 \quad \text{(MeV/g)}$$ \tag{2.3}
while an equilibrium charged-particle fluence spectrum \( \Phi^e_T \) (cm\(^{-2}\) MeV\(^{-1}\)) exists at each such point so that the absorbed dose can be written also as

\[
D_w = \int_0^{T_0} \Phi^e_T \left( \frac{dT}{\rho dx} \right)_w dT \quad (2.4)
\]

where \((dT/\rho dx)_w\) is the mass collision stopping power for \(w\) in absence of bremsstrahlung. Integral equation obtained by equating equations 2.3 and 2.4 has the following result for the fluence energy spectrum \(\Phi_T^e\):

\[
\Phi_T^e = \frac{N}{(dT/\rho dx)_w}
\]

However this is not a realistic spectrum because \(\delta\)-rays effects are ignored. Now it can be assumed that, because the same equilibrium spectrum of charged particles, \(\Phi_T^e\), crosses the cavity, the absorbed dose in the cavity medium is calculated also as

\[
D_g = \int_0^{T_0} \Phi_T^e \left( \frac{dT}{\rho dx} \right)_g dT = N \int_0^{T_0} \frac{(dT/\rho dx)_g}{(dT/\rho dx)_w} dT
\]

and then the ratio of dose in the cavity to that in the solid \(w\) is given by

\[
\frac{D_g}{D_w} = \frac{1}{N} \int_0^{T_0} \frac{(dT/\rho dx)_g}{(dT/\rho dx)_w} dT = m \bar{\gamma}_w^g \quad (2.5)
\]

the same B-G relation obtained previously, but with the additional Spencer’s assumptions of mono-energetic starting energy, charged particle equilibrium and absent bremsstrahlung.

### 2.1.3 Burlin Cavity Theory

This cavity theory has been introduced because of the gap between small cavities for which B-G or Spencer Cavity could be applied and very large cavities for which wall influence is negligible. In this second case, the averaged dose in the cavity is practically all delivered by particles which start and stop within the cavity. Considering an homogeneous medium \(w\) uniformly irradiated by \(\gamma\)-rays containing a cavity of medium \(g\) having intermediate dimensions, Burlin [21] made the following assumption:

- homogeneity of media \(w\) and \(g\);
- there is an homogeneous field throughout \(w\) and \(g\) and then no \(\gamma\)-ray attenuation corrections;
- CPE exists in all point of \(w\) and \(g\) which are far away respect to the maximum electron range from the cavity boundary;
- The same equilibrium spectra of secondary electrons generated in \(w\) and \(g\);
- An exponential attenuation of electron fluence along the medium \(g\) without a changing of the its spectral distribution;
- The fluence of electrons in the cavity builds up to its equilibrium value according to the same exponential low that applies to the incoming electrons.
If all the previous assumptions are valid, Burlin cavity theory provides the following expression for the ratio between the average absorbed dose $\bar{D}_g$ in the cavity medium $g$ and the absorbed dose $D_w$ in medium $w$:

$$\frac{\bar{D}_g}{D_w} = d \cdot m^g + (1 - d) \left( \frac{\bar{\rho}_e}{\rho} \right)^g_w$$

where $d$ is a parameter which takes into account the dimension of the cavity (it approaches 1 for small cavities and 0 for large ones); $m^g$ is the mean ratio of the mass collision stopping powers for $g$ and $w$, as obtained from the B-G or Spencer cavity theories; and $(\bar{\rho}_e/\rho)^g_w$ is the mean ratio of the mass energy absorption coefficients for $g$ and $w$ [10].

### 2.2 Dosimeters

The following paragraph presents the conceptual definition of a dosimeter. It is based on the cavity theories. Real dosimeter can be very different from the ideal dosimeter presented here. However the presented concepts provide the guidelines to realize a correct dose measure. The main characteristic parameters of a dosimeter will be presented. They represent the most important points by which a dosimeter must be matched. Some of these will be considered for 2D devices in the next paragraphs.

#### 2.2.1 Conceptual definition of a dosimeter

A dosimeter can be defined as a device which can provide a reading $r$ representing a measure of the absorbed dose $D_g$ deposited in its sensitive volume by ionizing radiation [10, 22]. When dose distribution inside the sensitive volume is not homogeneous, $r$ is a measure of the mean value $\bar{D}_g$. Similar considerations apply to dose rate measuring devices. However the main problem of dosimetry is the interpretation of a dosimeter reading in terms of the desired quantity (for example exposure or tissue dose). Generally a dosimeter can be considered as a volume $V$ filled with medium $g$, surrounded by a wall of another medium $w$ having a thickness $t \geq 0$. In this manner a dosimeter can be treated in terms of cavity theory. Applying this interpretation cavity walls have several important functions:

- being a source of secondary charged particles which contribute to dose in $V$ and provide CPE or TCPE conditions,
- shielding $V$ from charged particles outside the wall,
- protecting $V$ from external contaminants,
- serving as a container for a sensitive medium $g$,
- containing radiation filters to modify the energy dependence of the dosimeter.
According to the theory developed in a previous chapter, it is possible to write the following relations that are valid for photons under CPE and TCPE conditions, respectively [10]:

\[ D = K_c = \psi \left( \frac{\mu_{en}}{\rho} \right) \]  
(2.6)

\[ D = K_c (1 + \mu' \bar{x}) \equiv K_c \beta = \psi \left( \frac{\mu_{en}}{\rho} \right) \beta \]  
(2.7)

Where \( K_c \) is the collision kerma, \( \mu' \) is the common slope of the \( D, K \) and \( K_c \) curves observed in TCPE conditions at a given depth (where those curves becomes parallel) and \( \bar{x} \) is the mean distance the secondary charged particles carry their kinetic energy in the direction of the primary rays while depositing it as dose [10]. If dosimeter wall is thick enough to exclude all charged particle produced elsewhere and thin enough respect to the range of secondary charged particles generated in it and assuming that the volume \( V \) is small enough in order to not perturb charged-particle field, CPE exists near the cavity so that the dose measured in the cavity \( D_g \) can be used to measure the dose \( D_w \) in the wall material near the cavity, according to B-G or Spencer theory (equations 2.2 and 2.5). If a new medium \( x \) substitutes the dosimeter and the irradiation conditions are identical so that CPE continues to be valid, the dose \( D_x \) to medium \( x \) is given as

\[ D_x = D_w \frac{(\mu_{en}/\rho)_x}{(\mu_{en}/\rho)_w} \]

For higher-energy radiation (\( h\nu > 1 \) MeV), CPE fails and TCPE relations take place in the previous equations. A considerable advantage is to achieve a matching between the dosimeter and the medium of interest \( x \). Generally this means a matching in term of atomic composition and density state (gaseous vs. condensed). In addition a dosimeter is composed by wall which contains the sensitive media. Usually a good matching can be realized between the medium \( w \) of the wall and the medium \( x \). So one important problem to be addressed is the matching between \( w \) and \( g \). According to the Burlin cavity theory the relation between the averaged dose \( \bar{D}_g \) in the dosimeter’s sensitive volume and the equilibrium dose \( D_w \) (i.e. under CPE conditions) in the wall medium is given by

\[ \frac{\bar{D}_g}{D_w} = d \cdot m\bar{S}_w^g + (1 - d) \left( \frac{\mu_{en}}{\rho} \right)_w^g \]  
(2.8)

where \( d \) is a parameter related to the cavity size that approaches unity for small cavities and zero for large ones. \( m\bar{S}_w^g \) is the mean ratio of the mass collision stopping power for \( g \) and \( w \) and \( (\mu_{en}/\rho)_w^g \) is the mean ratio of the mass energy-absorption coefficients for \( g \) and \( w \). It can be seen that the average dose \( \bar{D}_g \) in the dosimeter sensitive volume is equal to the equilibrium dose \( D_w \) if

\[ m\bar{S}_w^g = \left( \frac{\mu_{en}}{\rho} \right)_w^g = 1 \]

independent of the value of \( d \). However these criteria are difficult to be realized, so the following more practical and flexible matching relationship is used:

\[ m\bar{S}_w^g = \left( \frac{\mu_{en}}{\rho} \right)_w^g = n \]
where \( n \) is a constant which is not necessarily equal to unity. The next step is to relate the dosimeter to the medium of interest \( x \). Also in this case the Burlin theory provides a possible calculation of the dose \( D_x \) at the point of interest where dosimeter is located. Three cases can be distinguished:

- \( d = 1 \), small sensitive volume. Indirectly ionizing radiation interacts only with the wall which must be matched as closely as possible to the medium \( x \) in order to not perturb the energy spectrum. So, applying the Burlin cavity relation, \( D_x \) can be obtained from the following relation:

\[
\frac{\bar{D}_g}{D_x} = \frac{\bar{D}_g}{D_w} \cdot \frac{D_w}{D_x} = m_S g \left( \frac{\mu_{en}}{\rho} \right)_x^w
\]

- \( d = 0 \), large sensitive volume. Now the wall influence on the dose medium \( g \) is negligible and, applying again the Burlin relation, \( D_x \) can be gotten from:

\[
\frac{\bar{D}_g}{D_x} = \frac{\bar{D}_g}{D_w} \cdot \frac{D_w}{D_x} = \left( \frac{\mu_{en}}{\rho} \right)_w^g \left( \frac{\mu_{en}}{\rho} \right)_x^w = \left( \frac{\mu_{en}}{\rho} \right)_x^w
\]

- \( 0 < d < 1 \), intermediate-sized dosimeters. This is the general case in which the full Burlin cavity theory must be used to obtain \( D_w \) from \( \bar{D}_g \) and then calculate \( D_x \) from 11.5.

Particular attention must be paid for the attenuation of uncharged radiations in penetrating into the dosimeter and into the medium. It is clear that the uncharged particle fluence reaching the center of the dosimeter (\( \psi_{dos} \)) is different by the fluence (\( \psi_x \)) at the center of that volume if it was replaced by the same material of interest \( x \). Then, in order to have the correct dose measurement, the dosimeter reading must be multiplied by the factor \( \psi_x / \psi_{dos} \), when CPE conditions are verified. In presence of TCPE, the multiplication factor becomes \( (\beta \psi)_x / (\beta \psi)_{dos} \).

### 2.2.2 General characteristics of dosimeters

In this paragraph it will be presented the parameters which characterize a dosimeter [21, 10, 22]. Some of the following characteristic parameters must be verified by 2D GEM detectors proposed in this work and compared to conventional dosimeters.

**Absoluteness**: a dosimeter is defined as absolute if it can be used to measure the absorbed dose in its own sensitive volume without requiring calibration. Typically it is preferred to work with calibrated dosimeters in order to refer to few certificate standard dosimeters in common with other radiation users. This allows to minimize the undetected errors which can appear in an absolute dosimeter.

**Precision**: it means the reproducibility of dosimeter measurements. Not only fluctuations in instrumental characteristics, ambient conditions and so on but also the stochastic nature of radiation fields determine statistical fluctuations in dose measurements whose precision can be obtained from repeated data measurements. Typically it is the standard deviation of these distribution which
provides the precision on single measurement or on a medium value calculated from a set of data respect to the expectation value.

**Accuracy**: the accuracy of a dosimeter measurement expresses the proximity of its expectation value to the true value of the quantity to be measured. An evaluation of accuracy from data themselves as is done for precision is impossible, because it is a measure of the collective effect of the errors in all parameters that influence measurements.

**Dose sensitivity**: a dosimeter must have an optimal dose sensitivity \( dr/d\bar{D}_g \) throughout the dose range to be measured. A linear response is preferable. However even a non-linear function \( r(\bar{D}_g) \) may be acceptable. In this case calibration require multiple point measurements.

**Dose rate response**: it is preferable that the dosimeter would be independent of dose rate. This is particularly important for Linacs in which dose is delivered in pulses at a variable frequency so that an assigned dose can be released at different times.

**Lower range limit**: this can be defined as the instrumental background or zero-dose reading, that is to say the value of \( r = r_0 \) when \( \bar{D}_g = 0 \).

**Background readings**: typically, when a dosimeter is set up in measurement conditions with \( \bar{D}_g = 0 \), it provides an additional background reading \( r_b \) which sums to the lower limit \( r_0 \). This is the result, for example, of cosmic rays, natural and man-made terrestrial sources.

**Dose range upper limit**: it represents the maximum dose range which a dosimeter can measure. Depending on the dosimeter it can have inherent limits which can include exhaustion of the entities (atoms, molecules and so on) being acted upon radiation, competing reactions by radiation products and radiation damage to the dosimeter. Usually when a dose range limit is reached it appears as a decrease in dose sensitivity.

**Energy dependence**: an ideal dosimeter should be independent from different beam quality. Its energy dependence derives from the irradiated materials. Preferably the dosimeter is made of a material with an effective atomic number as close as possible to the material under investigation.

**Spatial resolution**: a dosimeter should be able to determine the absorbed dose in an infinitesimal volume, a point by point measure. Real dosimeter can measure in a discreet number of points where dose is limited by the size of the detector in that point. The lack of a high resolution for most dosimeter makes difficult the dose evaluation in regions of high-dose gradients.

**Easy of handling**: a dosimeter should be simple to use and physically sturdy. It must be suitable for the real clinical measurement conditions.
2.3 Principal dosimeters in radiotherapy

Dosimeters used in radiotherapy [24] are devoted mainly to beam calibration when working in reference conditions and doses checking when a treatment plan must be verified. In this second conditions the dose checking is executed without patient. However sometimes dosimeters are placed together with the patient and measures are performed during the treatment session. These methods belongs to the so called “in-vivo dosimetry”. In the following paragraph some of the most important dosimeters will be presented grouping them in 1D and 2D dosimeters. 1D refers to those dosimeters which can provide a dose measurement in a point, while 2D refers to those which are able to provide a dose measurement on a given area. Detectors presented in the present work belong to this second class. Film dosimetry is also a 2D technique but it will be treated in more detail in the par. 2.4, because it is the comparative technique for the new presented detector.

2.3.1 1D dosimeters

Among the most important dosimeters used to make a point measurement in radiotherapy are the ionization chambers and some solid state dosimeters based on different technologies [22,23]. Among these last ones it is possible to distinguish passive dosimeters (TLD and OSL) and active dosimeters (silicon diodes and diamonds). There are other type of dosimeters less used, but for extensive treatment is remind to specialized texts [22, 23, 24]. In the present paragraph, a brief description will be given for ionization chambers and solid state dosimeters.

IONIZATION CHAMBERS

Ionization chambers have been used in all type of dosimetry because of their simplicity in design and low operating cost. In clinical practice they became standard dosimetry tools. In radiotherapy small ionization chamber are routinely used for absolute and relative dosimetry. An ionization chamber is basically a cavity surrounded by a conductive outer wall with a central collecting electrode. The wall and the collecting electrode are separated with a high quality insulator to reduce leakage current when a polarizing effect is applied. Often an guard electrode is provided in the chamber to further reduce the leakage currents which are intercepted and flowed to ground bypassing the collecting electrode. Another important device needed for this type of chambers is the electrometers which typically is an high gain negative feedback, operational amplifier with a standard resistor and a standard capacitor in the feedback path. It is able to measure very small currents of the order of $10^{-9}$ A.
Fig. 2.1 shows the scheme of a Farmer-type thimble ionization chamber, one of the most used devices used in radiotherapy. A different scheme is represented by the parallel-plane ionization chamber. It consists of two plane walls, one serving as an entry window and a polarizing electrode and the other as the back wall and collecting electrode as well as guard-ring system. The back wall is made of a non-conducting material (plastic, for example) with a thin conducting layer of graphite forming the collecting electrode (fig. 2.2).

Figure 2.2: layout of parallel plate ionization chamber.

Usually this type of chambers requires significant fluence perturbation correction because of an inadequate guard width.

TLD and OSL

Both dosimeters belongs to the so called “luminescence dosimetry”. These are based on the property of some materials which, upon absorption of radiation, retain part of the absorbed energy in metastable states. Then this energy is released in the form of ultraviolet, visible or infrared light. This is defined as “the luminescence” coming from these materials and it can be of two type: florescence and phosphorescence. They differ from the time delay between stimulation and emission of light. In the first case, the time delay is between $10^{-10}$ and $10^{-8}$ seconds, while, in the second case, it is major than $10^{-8}$ seconds. The phosphorescence can be very slow and it can be accelerated with a suitable excitation which can be in form of heat (thermos-luminescence, TL) or light (optically stimulated luminescence, OSL). Usually they are crystalline materials like...
Magnesium and Titanium doped Lithium Fluoride (LiF: Mg, Ti) and Carbon doped Aluminium Oxide (Al2O3: C). The interaction of high energy X-rays produce secondary high energy electrons which interact in the material and release numerous low energy free electrons and holes. The free electrons and holes will either recombine or trapped in an electron or hole trap. In general there are two types of traps: storage traps where free charges are trapped before being released after a subsequent heating or optical stimulation and recombination centers in which a storage charge carrier can recombine with another trapped charge carrier of opposite sign. Thermal and optical stimulations are the bases of thermo-luminescence dosimetry (TLD) [25] and optical stimulated dosimeters (OSL) [26], respectively. TLDs are available in various forms (powders, chips, rods, ribbon, etc.). In addition they need of a reader system consisting of a planchet for placing and heating the TLD dosimeter; a photomultiplier (PMT) to detect TL light emission, covert into an electrical signal and amplify it. At the end there is an electrometer to record the PMT signal as a charge or current (fig. 2.3).

![Figure 2.3: TLD dosimeter read-out processing.](image)

Optical stimulated luminescence presents a similar scheme. Typically an OSL dosimeter consists of a small chip coupled with a long optical fiber, a laser, a beam splitter with a collimator, a PM tube, electronics and software. Laser is used to excite, through the optical fiber, the chip. Produced OSL luminescence is carried back in the same fiber, reflected at 90° through the beam splitter and measured by the PMT.

**SEMICONDUCTOR DOSIMETRY**

Generally these dosimeters are based on p-n junction diodes [28]. They are obtained by taking n-type or p-type silicon and counter-doping the surface to produce the opposite type material, obtaining n-Si or p-Si dosimeters. In radiotherapy, p-Si type dosimeter is used because it is less affected by radiation damage and has a much smaller dark current. When a radiation interacts in the body of the dosimeter, it produces electron-hole pairs. Then the minority carriers diffuse into the depletion region and are swept across the depletion region under the action of the electric field. In this way a current is generated in the diode. Diodes area particularly used in measurements of small fields like those applied in stereotactic radiosurgery and high dose gradient areas like penumbra regions. They are also used in water phantoms, also with beam scanning devices. One important application concerns in-vivo dosimetry. Usually, in this case, they are provided with a build-up encapsulation and hence must be appropriately chosen, depending on the type and quality of the clinical beams. Encapsulation serves also to protect the fragile diode from physical damage. A valid
alternative to diodes is represented by Metal-Oxide Semiconductor Field Effect Transistor (MOSFET).

The working principle can be explained with reference to fig. 2.4. When a high voltage is applied to the polysilicon, a large number of holes move from the surrounding regions into the oxide layer and the adjacent silicon substrate. If a large number of holes are gathered in that area, they form a current channel between the source and the drain SiO$_2$ regions. Then a small voltage, called the threshold voltage, can be applied to initiate the current flow. The radiation passing through the oxide region produces electron hole pairs. The holes move towards the interface of silicon and SiO$_2$, where they get trapped. This excess positive charge induces current in the channel between the source and the drain. The applied threshold voltage allows a shift proportional to the positive charge buildup which is proportional to the energy deposited by the incident radiation.

2.3.2 2D dosimeters

In general 2D dosimetry is a widely used technique in radiotherapy. It has the advantage of providing 2D map of the dose delivered by a prescribed field. 1D dosimeter can measures dose point by point along some specific directions like the IN PLANE and CROSS PLANE directions (par. 3.3.2), or along the beam direction as happens in PDD dose measurements. However this is not sufficient to obtain a correct dose distribution information, especially for those fields, which have not regular shape like IMRT fields. To obtain 2D dose map several devices can be used, but basically they can be grouped in chamber matrix arrays, radiosensitive films and, as an alternative use, Electronic Portal Imaging Devices (EPID) [28]. Film dosimetry is discussed in par. 2.4.

CHAMBERS MATRIX ARRAYS

These 2D dosimeters can be defined simply as a matrix of point chambers [27, 28]. Each of these chambers can be small ionization chamber or a diode similar to those presented in the previous paragraph [29, 30]. There are a lot of commercial available matrix arrays. The number of chamber is variable depending on field areas to measure and the required degree of resolution. Pitch between two successive chambers can also be variable. Often the central region has a small pitch respect to the neighboring side regions. In the following figures there are presented some of the widespread chambers matrix arrays usually used in clinical practice.
Figure 2.5: PTW seven29: 729 ionization chambers matrices, chamber dimension 5x5 mm, max field 27x27 cm², pitch 1 cm.

Figure 2.6: SunNuclear Mapcheck and Mapcheck 2: 1527 diodes matrix, diode dimensions 0.8x0.8 mm², pitch 7.07 mm

Figure 2.7: IBA MatriXX: 1024 ionization chambers matrix, chamber dimensions 4 mm in diameter, max field 24.4x24.4 cm², pitch 7.5 mm.

All the previous devices have a pitch of few millimeters. Doses between two consecutive detectors are not measured. Usually these systems are supplied with a software which is able to calculate the intermediate dose values by interpolating the measured points. This procedure can not represent a
problem on large area uniform fields, but become a drawback for small fields and high gradient doses. These configurations are particularly frequent for IMRT fields and an higher resolution is required.

EPIDs

During a treatment session, an important step is represented by the patient positioning before being irradiated. It is needed that reference points identified in the TPS prescriptions realized on CT data coincide with the real patient position and his internal structures. This kind of check is realized with EPIDs [31] which are silicon flat panels positioned on the opposite side of the accelerator head beyond the bed. It is able to provide a radiography of the patient using the same irradiating source applied for the treatments. The obtained images are not optimal because high energy x-rays are not a good probe to realize radiographies, but they allow to distinguish the important anatomical structures (like bones, for example) so that patient can be correctly positioned. Many clinical centers studied the possibility to use these flat panels also for 2D dosimetry [32]. Two class of EPIDs can be distinguished: direct and indirect. In the first case, EPID are made of a matrix of amorphous silicon diodes and have a thin build-up layer (usually, 0.6 mm thick copper) which works as a converter. An high energy X-ray interacts in the converter and produces secondary electrons which are detected from diodes. In the second case, EPID have a converter plus a phosphor layer. Now secondary electrons excite the phosphor layer which produces visible photons which, in turn, are detected from diodes. More advanced models provides matrices of about 800 x 1000 diodes with a pitch minor than 400 µm. It seem the best solution to be exploited, especially in terms of spatial resolution. However dose measurement with these devices is very complicated: actually there is no a unique calibration method to convert an EPID image in a 2D dose map. Usually they are calibrated using a reference ionization chamber or Monte Carlo Simulations and converting the pixel value in grays scale. These calibration methods are not sufficiently reproducible to cover all treatment techniques and irradiation configurations. In addition indirect EPIPs, which are more sensitive, suffer for X-ray and light scatterings. Direct EPIPs do not presents these drawbacks, but not sufficiently sensitive to realize a valid measure of the dose distributions. Recently also ageing problems have been reported.

2.4 Film dosimetry

A valid alternative to 2D matrix arrays is represented by film dosimetry. It is possible to distinguish radiographic and radiochromic films [33]. The first ones are applied in different fields: diagnostic radiology, radiation therapy and radiation protection. The second ones are specific for radiotherapy dosimetry. An unexposed radiographic film consists of a base of thin plastic with radiation sensitive emulsion coated uniformly on one or both sides. Emulsion consists of silver bromide AgBr grains suspended in a gelatin. After they have been exposed to radiation, AgBr ionizes and a latent image forms: it must be developed to became visible. Then using special devices called densitometers film opacity is measured in terms of optical density OD which is defined as follows:

\[ OD = \log_{10}(I_0/I) \]
where $I_0$ is the initial light intensity and $I$ is the intensity transmitted through the film. Because energy dependence is pronounced for low energy photons and their response depends on several parameters, the dose range of films is limited. The relationship between the dose and OD should be linear, but it usually is not the case. There can be nonlinear regions. Then for each film the dose vs OD curve, known as the H&D curve, must be established before a dose measurement. A typical H&D curve (from the initials of their developers, Hurter and Driffield) is shown in the fig. 2.8.

![Figure 2.8: characteristic H&D curve for a radiographic film.](image)

Four regions can be distinguished: (1) fog at low or zero exposure, (2) toe, defined as the region of under exposure, (3) linear portion which is referred as the optimum measurement conditions and (4) shoulder, defined as the of overexposure. Based on these curve, some important characteristic parameters are defined:

- the gamma of the film, which is defined by the slope of the straight line portion of the H&D curve; exposure should be chosen in order to have all parts of the radiograph lying in the linear portion of the H&D curve;
- the latitude, defined as the range of exposures over which the densities will lie in the linear region;
- the speed, defined as the exposure required to produce an optical density of 1.0 greater the OD of fog

Radiographic films are applied for qualitative and quantitative measurements in radiotherapy, including electron beam dosimetry, quality control of radiotherapy machines, verification of treatment techniques in various phantoms and portal imaging.

### 2.4.1 Radiochromic films

It is a new type of film which is used mainly in radiotherapy dosimetry [33]. Actually the technology of so-called “radiochromic films” has reached an high level of progress: it is a colorless film with a nearly tissue equivalent composition (H 9.0 %, C 60.6 %, N 11.2 % and O 19.2 %) that develops a blue color upon irradiation exposure. Coloration occurs because it contains a special dye that gets polymerized after irradiation. Radiochromic films are self-developing and does not need
neither developer nor films. In addition it is grain-less and then it has a very high resolution so that it can be used in high dose gradient regions. Respect to radiographic films it has some advantage: ease of use, requiring dark rooms, film cassettes or film processing; dose rate independence; better energy characteristics except for low energy X rays (lower than 25 kV); a relative low insensitivity to ambient conditions (even if excessive humidity must be avoided). Radiochromic film are less sensitive than radiographic films and are more suitable at higher doses, even if dose response non-linearity should be corrected at higher doses. There are different commercial available gafchromics. Some of these are listed together with its main characteristics in the table below.

<table>
<thead>
<tr>
<th>GAFchromic type</th>
<th>dimensions Nominal dose range</th>
<th>Approximate sensitivity</th>
<th>Schematic structure</th>
</tr>
</thead>
<tbody>
<tr>
<td>GAFchromic HD-810</td>
<td>25cm x 25cm x 0.1 mm 10-1000 Gy</td>
<td>3 mUA/Gy</td>
<td><img src="image" alt="HD-810" /></td>
</tr>
<tr>
<td>GAFchromic MD-55-2</td>
<td>12.5cm x 12.5cm x 0.23 mm 1-100 Gy</td>
<td>20 mUA/Gy</td>
<td><img src="image" alt="MD-55-2" /></td>
</tr>
<tr>
<td>GAFchromic XR-QA</td>
<td>36cm x 43 cm x 0.26 mm 0.001–02 Gy</td>
<td></td>
<td><img src="image" alt="XR-QA" /></td>
</tr>
<tr>
<td>GAFchromic EBT</td>
<td>20cm x 25cm x 0.23 mm 0.05–100 Gy 400-800 mAU/Gy</td>
<td></td>
<td><img src="image" alt="EBT" /></td>
</tr>
</tbody>
</table>

Table 2.1: some of the main gafchromics used in radiotherapy dosimetry with their most important characteristics.

Usually sensitivity is expressed in mill-Absorbance Units per Gy. Absorbance A is a measure of the amount of light absorbed by a material and is defined as follows:

\[ A = \log_{10}(I_i/I_r) \]

where \( I_i \) is the measured intensity of the incident beam and \( I_r \) is the measured intensity of the transmitted beam. In this work, the reference dosimeters are EBT type gafchromics which will be discussed in par. 2.4.2.
2.4.2 EBT GAFchromics

This paragraph presents a brief description of the EBT type gafchromics [13, 14], which is the reference dosimeter used in this work. In particular the model used in the present work are EBT2 and EBT3. Both are an evolution of EBT1 which has been presented in the previous table, and now is no longer available. The structure of EBT2 gafchromics is illustrated in fig.2.9.

It is made of clear polyester substrate (175 µm thick) coated with an active layer film (normally 30 µm) over which a topcoat 5 µm thick is applied (not shown in the figure). EBT1 is limited to only these three layers. EBT2, in addition, has an over-laminate polyester (50 µm) with approximately 25 µm of pressure sensitive adhesive layer. The over-laminate protect the active layer/topcoat from mechanical damage like water or other liquids. EBT3 has a more simple structure: the active layer is sandwiched between two Mette Polyester layers 120 µm thick. This symmetric configuration avoids the formation of artefacts on the scanned image, like Newton rings. In addition, its active layer contain a special dye which make it appear as a transparent yellow film. The yellow marker dye decreases UV/light sensitivity and, used in conjunction with an RGB film scanner, enables all the benefits of multichannel dosimetry. As regards to the other characteristics EBT3 is practically identical to EBT2. Chemical composition of the constituent layers of EBT2 is reported in table 2.2.

When an EBT2/3 gafchromic is exposed to X-ray, the active layer, which contains a special dye, gets polymerized. The polymer absorbs light and the transmission of light through the polymer can be measured using a special scanner. In the present work the scanner is the EPSON 10000XL model [35].
Both EBT2 and EBT3 have a characteristic absorption spectrum (fig. 2.11) in which it is possible to distinguish three peaks which define the three channels of the gafchromic: RED, GREEN, and BLUE.

Figure 2.11: characteristic absorption spectra for an EBT2 gafchromic: it is possible to distinguish the BLUE, GREEN and RED peaks. This last one is indicated by an arrow.

Each channel has a different sensitivity to doses. RED channel has an higher sensitivity respect to the others in the range from the minimum detectable dose (1 cGy for EBT2/3) to about 8 Gy. GREEN channel, instead, has an higher sensitivity in the range from 8 Gy to 40 Gy. These characteristic are clarified by the plots of fig. 2.12, in which the net optical densities and their derivatives are plotted in function of dose [32].

Figure 2.12: net optical density curves and their first derivatives [34].
In the second plot it is clear when the GREEN line overcomes the RED and BLUE lines. It can be noticed also that the BLUE curve remain always lower the others. The blue channel had a lower response gradient at any dose because the signal has weak dose dependence while it has a strong dependence on the thickness of the active layer. This makes the blue channel less useful than the other channels for dose measurements but provides a response signal for automatic film uniformity enhancement via a special marker dye in the active layer. Like all other radiochromic films, the main advantage of EBT are:

- No post irradiation processing: they became colored directly after irradiation, not like radiographic film or TLD, for example.
- High spatial resolution: spatial resolution of a fraction of a millimeter is not uncommon for typical films.
- Good spatial uniformity: best obtained values are better than 95%.

On the other hand, there are some disadvantages which must be taken into account when using these dosimeters:

- Fading: a radiochromic film shows a post irradiation fading and typically this is higher immediately after irradiation and then decreases slowly. Usually it is recommended to wait for a given time before scanning the dose. In particular, when it has been calibrated, the same delay time applied during calibration, must be ensured.
- Temperature dependence: optical density of radiochromic film is highly sensitive to temperature and this parameter must be kept constant during irradiation.
- UV sensitivity: some radiochromic are sensitive to UV light.
- Film orientation: constituent materials of radiochromic films are non-isotropic crystals and then optical density must be measure with the same orientation at which the film was exposed.

2.4.3 Estimate of the minimum dose on EBT GAFchromic

In Chapter 1, the Synergy medical LINAC has been described. It has pulsed temporal structure with an estimate of $8.5 \times 10^{12}$ photons/(cm$^2$sec). The aim of this paragraph is to set an Fluka simulation in order to obtain an estimate of the dose released in an EBT type gafchromic by a single pulse. gafchromic is defined as a simple polyester foil with a thickness equal to the EBT2 model (285 µm). The high energy X-rays source is placed at a distance of 100 cm from the GAF. It has the 6 MV spectrum and an area of 8 x 8 mm$^2$. Beyond the GAF there is a 12 thick mylar foil followed by an ArCO2 gas layer 3 mm in thickness. Beyond there is a sequence of layers which together simulate the constitutive elements of a triple-GEM detector. Its description is reminded to the following chapter 3. Now it is introduced because GAFchromic measurements have been performed placing the foil on the detector window. Then all these layers are necessary to take into account the real experimental conditions. Fig. 2.13 shows the results of these simulations.
Figure 2.13: simulated results for the interaction of an EBT gafchromic with a 6 MV 8 x 8 mm² beam; UP: side view of the photon distribution of the beam (left) and energy distribution along the detector on which the gafchromic is placed (right); DOWN: dose distribution along the detector with gafchromic (left) and the dose deposition along the detector on a central region of 100 x 100 µm².

Now if the simulated value of $D_s = 2.5 \times 10^{-6}$ GeV/(gr·pr) is taken as the dose released in the GAF film active layer (30 µm), the estimated dose $D_{GAF}$ (in Gy) can be calculated from the following formula:

$$D_{GAF} = \gamma_{\text{pulse}}D_s \times 1.6 \cdot 10^{-7}$$

where $\gamma_{\text{pulse}}$ is the number of primary photons per pulse calculated as the product of $\dot{\gamma}_{\text{pulse}} (= 8.5 \times 10^{12}$ ph/cm² sec), the area of the beam (8 x 8 mm²) and the pulse time width (3 µs). The numerical factor (1.6 x 10^{-7}) comes from the conversion of Ge/gr to J/kg (= 1 Gy). The result is that the dose released per pulse to the GAF film is

$$D_{GAF} = 6.4 \times 10^{-6} \text{ Gy/pulse}$$

This is means that to reach the sensitive value (1 cGy) of the EBT2/3 GAF a minimum of 1600 pulses must be sent. This value of dose can be easily exceed irradiating, for example, for 30 sec with a dose rate of 100 MU/min which corresponds to 100 Hz of PRF and then to 30 sec x 100 Hz = 3000 pulses. In the present case GAFs have been irradiated with a number of MUs major than 1000 with dose rate of 100 or 200 MU/min and then 6 x 10^4 pulses which corresponds to about 40 cGy. Dose rate was not important because the new EBT gafchormics are practically independent from this parameter.
Chapter 3

Introduction

Gas dosimeters like those presented in the previous chapter are detectors working in ionization mode. A GEM detector, instead, can be defined as a proportional gas chamber and can be conceived as an evolution of the Multi-Wire Proportional Chamber (MWPC) [36], which has been used not only in high energy particle physics experiments, but also in other fields like astro-particle physics and medical applications. However, the advent of new high luminosity colliders has shown some limitations for the MWPC concerning the capacity to tolerate high energy radiation fluxes. As it is known, the rate capability of wire detectors is limited by the drift of ions from the anode wire toward the cathode. High rate fluxes produce an ion charge around the wire with a consequent rate dependent gain drop and a loss of efficiency. The maximum rate capability of typical MWPC is below 1 MHz/cm$^2$ [37]. The attempt to overcome these limits triggered the research to new technologic solutions. Many new alternative detectors have been invented: the microgap chamber (MGC) [38], the Microdot [39], the “Computer à Trou” (CAT) [40], the Micromesh gas chamber (MicroMegas) [41], the Micro-Groove [42] and so on. All these detectors belong to the family of the so called micro-pattern detectors. The Gas Electron Multiplier (GEM), proposed by Sauli in 1997 [43], belongs to this family. Along these years, the GEM has represented the main elements of a new generation of detectors which found applications not only in high energy physics. In the following chapter, a new GEM detector, conceived for X-ray applications, will be described. Moreover a series of preliminary tests on therapeutic X-ray beams will be presented in order to validate its applicability as a 2D dosimeter on high energy X-rays. After the first experimental tests, the X-rays detector interactions have been simulated in order to understand first observed results and validate the next measurement sections. The last section presents experimental measurements performed simultaneously with GEM detector and gafchromic films with their relative gamma test acceptance analisys.

3.1 Gas Electron Multiplier technology

A GEM is made of a thin kapton foil (50 µm), a copper cladding (5 µm) on each side, perforated with a high surface density of holes. Each hole has a bi-conical structure with external (internal) diameter of 70 µm (50 µm) and a pitch of 140 µm. The GEM manufacturing technology is realized using conventional photolithography methods [44]. The GEM is the basic building block of a new generation of gas detectors, in particular it is used as charge multiplication stage. Applying a suitable voltage difference (~ 100 kV/cm) between the two copper sides, a high electric field is generated inside the holes. An electron which enters in a hole can acquire enough energy to develop an avalanche with a gain which can be greater than $10^3$. Gain can be increased rising the applied voltage or reducing the hole diameter. However, working at a given potential, it has been observed
that the GEM gain saturates for hole diameter below ~ 70 µm and this effect is due to the increasing losses of electrons in the avalanche to the lower GEM electrode. Hole pitch is correlated to the collection efficiency of the electrons released in the upper volume of the GEM foil. A small pitch provides a high collection efficiency but manufacturing difficulties arise. Hole diameter and hole pitch are related to the so called optical transparency $t_{GEM}$ of a GEM. It is defined as the ratio between the total area of holes and the total area of the foil: $t_{GEM} = \mu D^2 / 2\sqrt{3}p^2$, where $D$ is the external diameter and $p$ the pitch, having assumed a cylindrical hole. Another important parameter is the hole shape, because this influences the charging-up: electrons and ions from the avalanche can be collected and accumulated on the kapton surface and can produce an alteration of the electric field inside the multiplication channel. The cylindrical shape is the best geometry which minimizes this effect. The bi-conical shape is a compromise between a good production yield and a limited charging-up effect. The following paragraphs are devoted to the description of a typical GEM detector like as been conceived at Laboratori Nazionali di Frascati (LNF) in particular the triple-GEM configuration.

### 3.1.1 The GEM detector

The simplest GEM detector is made of a GEM foil sandwiched between two flat parallel electrodes which act as cathode and anode and define also the volume of the chamber. Typically the cathode represents also the detector window and the chamber is sealed so that a gas mixture can flow inside. The gas region between the cathode and the first GEM foil (drift region) represents the active volume of the detector. In order to collect ionization in this volume, an electric drift field is applied between the cathode and the first GEM electrode. A similar electric field is applied in the gap between the second GEM electrode and the anode (induction region). A particle which interacts in the drift region produces electron-ion pairs which, due to the electric field, began to drift. Electrons drift towards the GEM and are collected into the holes where they undergo charge multiplication. At the same time, the positive ions produced in the holes are mainly collected on the upper side of the GEM foil instead of drifting towards the cathode, so that the holes are liberated from positive charges in a very short time (few µs). The electron avalanche produced, instead, drifts toward the anode and induces a current pulse signal. So it is possible to reduce the effect of positive space charge and, at the same time, use the fastest electrons to induce the signal on the anode. It is clear that these two effects represents a great advantage which improves the rate capability of the detector to a value major than 100 MHz/cm$^2$. Considering the scheme of a single GEM detector, it is possible define the main parameters which describe a typical GEM detector at a defined geometry and gas mixture:

- electric fields in the drift and induction gap
- thickness of the drift and induction region
- voltage difference applied to the GEM foil

The study of electric fields in a GEM detector has been performed with MAXWELL and GARFIELD simulation tools [46]. Fields lines in the drift and induction region are distributed uniformly and are similar to that seen in a parallel capacitor. Instead lines close to the GEM increase in density near and inside the holes. The drift field has the role to collect the ionization
electrons produced by radiation into the GEM holes. As can be seen from figure 3.1, varying the drift field, a corresponding variation of the relative signal amplitudes on the PCB anode is obtained.

At low values (< 0.5 kV/cm) the curves drop due to low electron drift velocity and large diffusion. The same happens at higher values (> 3 kV/cm) of the drift field; but, now dropping is due to the defocusing effect of the field lines above the GEM, which leads the primary electrons to be directly collected on the upper electrode of the GEM. At intermediate values, a plateau is reached and represents the working point of most GEM detectors. On the other side, the induction field has the role of extracting the multiplied electrons from the GEM holes and transferring them to the anode. Also in this case the optimal working value is a middle one: at low induction fields, all secondary electrons are practically collected on the bottom GEM electrode with a resulting vanishing induced signal, while at high induction fields (> 8 kV/cm) discharge on the anode can occur due to the high electric field in proximity of the readout electrode edge. Drift and induction gaps are the other important parameters to consider. The thickness for the first one must be chosen in order to ensure an high particle detection efficiency. Typically a thickness of 3 mm is the shared choice which guarantees a full efficiency. This point will be discussed in relation to high energy X rays, subject of this thesis. The induction gap, instead, is chosen to be 1 mm thick in order to maximize the signal fraction integrated by the amplifier according to the Ramo’s theorem [48]. To further improve, a thinner gap could be used, however it is not preferable to go below 1 mm, because it would require a high mechanical tolerance in order to avoid discharges on the Printed Circuit Board (PCB), and gain non-uniformity of detector. The last parameter is the GEM voltage. When a voltage \( V_{GEM} \) is applied between the two GEM electrodes, it is possible to define an intrinsic gain \( G_{in} \) directly proportional to \( V_{GEM} \), according to the following relation:

\[
G_{in} = A e^{\langle a \rangle V_{GEM}}
\]

where \( A \) is a proportionality constant and \( \langle a \rangle \) is the first Townsend coefficient along the electron path through the hole. It depends on the gas mixture and the applied electric field. However, due to dispersive effects, the real number of electrons transferred on the anode is lower. In order to take
into account these phenomena, an effective gain $G_{\text{eff}}$ is defined as the product of the “so called” electron transparency $T$ and the intrinsic efficiency $G_{\text{in}}$:

$$G_{\text{eff}} = T \cdot G_{\text{in}} = \varepsilon_{\text{colt}} \cdot f_{\text{extr}} \cdot G_{\text{in}}$$

$T$ is the product of the collection efficiency $\varepsilon_{\text{colt}}$ and the extraction function $f_{\text{extr}}$. The first quantity is defined as

$$\varepsilon_{\text{colt}} = \frac{\text{electrons collected in the holes}}{\text{electrons produced above the holes}}$$

and represents the ratio between the number of electrons entering the multiplication channels and the number of primary electrons generated above the GEM. The second quantity is defined as:

$$f_{\text{extr}} = \frac{\text{electrons extracted from the holes}}{\text{electrons produced in the holes}}$$

and represents the ratio between the number of electrons extracted from the holes and transferred to the PCB and the number of electrons multiplied inside the amplification channels.

### 3.1.2 The triple-GEM detector

The scheme of a triple-GEM detector is shown in the following figure 3.2 [44, 45]. It consists of three GEM foils piled-up and sandwiched between two electrodes, a cathode and an anode. Shown in the fig. 3.2, there are also the names and the typical geometrical dimension of the gaps between the various elements: drift and induction gaps are the same as in the previous case and play the same roles, but now the gain is realized using three GEM foils. This configuration allows to reach higher detector gains before the appearance of discharges, without requiring too high voltage on each single GEM foil.

![Figure 3.2](image)

*Figure 3.2: cross section of a triple-GEM detector: $E_D$, $E_{T1}$, $E_{T2}$ and $E_I$ are the electric fields for drift gap, 1° and 2° transfer gap and induction gap; while $V_{GEM1}$, $V_{GEM2}$ and $V_{GEM3}$ are the voltages applied to the GEM foils.*
The transfer gap regions $g_{T1}$ and $g_{T2}$ represent transition regions in which the secondary electrons produced in the holes of the upper GEM are transported towards the holes of the next GEM. The charge transfer is realized applying a proper electric field in order to maximize at the same time the extraction fraction from the upper GEM and the collection efficiency to the lower GEM. The following figure 3.3 shows the induced current on the electrode readout as a function of the transfer fields for an Ar/C02 (80/20) gas mixture for typical values of drift and induction fields ($ED = 2$ kV/cm, $EI = 5$ kV/cm).

![Figure 3.3: Induced current on the readout electrode as a function of the transfer field ($E_{T1}=E_{T2}$) for the Ar/C02 (80/20) gas mixture.](image)

It can be observed that for low transfer fields, electron current is affected by the low extraction fraction, while at high electric fields, the electron charge is reduced due to the poor collection efficiency due to the high defocusing effect. Typically values in range from 3 to 5 kV/cm are applied.

### 3.1.3 Detector construction

The new triple-GEM detector has been assembled in the clean rooms of the LNF. The usual procedure follows some standardized construction steps:

1. Three 10 x 10 cm$^2$ GEM foils have been tested in current to verify the correct functionality. A gradually high voltage is applied to the GEM electrode and the corresponding current has been measured. Typically a GEM foil works correctly when at 500 V, the measured current is not major than 2 nA.

2. Each GEM foil is placed on a “tendi-GEM” device which serves to stretch the foils.

3. Each stretched foil is glued to an epoxy glass frame. Frames have a thickness of 1, 2 and 1 mm for GEM1, GEM2 and GEM3 respectively.
4. Stretching and gluing on a frame 3 mm thick is also executed for an aluminate mylar foil which will represent the cathode and the window of the detector. Mylar is aluminate on one side only and this faced the internal side of the detector.

5. The frame with mylar, those with GEMs and the PADized anode are glued and piled up together.

![Figure 3.4: the 128 PADs anode before assembling (left), piling of a triple-GEM detector (center) and a glued tri-GEM detector with the electrode strips to be soldered (right)](image)

6. All the electrode strips from the GEMs, the cathode and the anode are soldered on a set of copper connector pins.

7. At the end some two epoxy glass connectors are fixed on the detector in order to insert the exit and entrance gas tubes.

### 3.1.4 High Voltage power supply and current measurements

A triple GEM detector requires a High Voltage system with seven power supplies to create the electric fields inside the chamber and to power the GEM foils. Up to now the HV systems for GEM detectors have been realized using either passive dividers or seven independent HV channels referred to ground. In the first case the fields and the GEM voltages are determined once forever by the values of the resistors used and the current drawn by the divider is two order of magnitude higher than that needed for the chamber operation. In the case of seven independent channels the flexibility is higher but a particular attention has to paid when switching on/off the chamber in order to prevent an irreversible damage to the GEM foils. Actually power supply to the GEM detector is realized with a new High Voltage System, called “HVGEM” [51]. It is an Active High Voltage Divider, with seven floating power supplies. It is similar to a set of seven batteries stacked in a row. The low operating voltage and power consumption are some of the most remarkable characteristics of this device. A functional scheme of the HVGEM module is reported in fig. 3.5.
It has seven fully floating channels which are built using resonant switching sinusoidal wave model. For each channel it is possible to set and read the output voltages. Moreover the first six channels can provide a maximum current of 150 µA, while the seventh can provide a maximum of 100 µA. An important characteristic is that for each channel there is a nano-ammeter which is able to measure currents collected on the cathode and on each GEM electrode.

Previous figure 3.6 represents schematically electric field lines near and inside the holes. Electrons and positive ions are drawn in blue and in red, respectively. As can be observed some of the field lines terminate on the upper and lower GEM electrodes, so that a given fraction of the charge produced inside the holes is collected on the electrode and this is particularly important for positive ions which must quickly release the multiplication region. In the present study, measuring the current produced by ionization radiation is an important point to evaluate the correct functionality of the detector: it is not a localized measure as in the case of pads, but a global measure which could be correlated to the total counts registered on the anode by the FEE electronics.

3.1.5 Front-End-Electronics

On the other side of the anode there is the all the Front-End-Electronics (FEE) which is made of a set of CARIOCA chip cards [52] and an “intelligent” mother board based on a Field-Programmable-Gate-Array (FPGA) [53].
For each card there are two CARIOCA chips which are integrated circuits based on 0.25 µm C-MOS technology. A scheme of the CARIOCA chip is reported in the following fig. 3.8.

The input of the CARIOCA circuit is pseudo-differential and consists of two identical current-mode amplifiers. One amplifier is connected to the chamber pad through a trace on the printed circuit board (PCB). The other is a dummy amplifier and is used as a floating input to provide a DC balance to the shaper. Following the amplifier there is a shaper circuit and a differential amplifier which provide further gain together with signal and amplifier tail cancellation. The signal coming from the differential amplifier is sent to a discriminator which select all the signal over a set threshold. At the end the discriminator output is sent to an LVDS driver which provides the chip output signal. In addition a baseline restoration circuit, realized as a nonlinear feedback loop on the second differential amplifier, limits baseline fluctuations. Its sensitivity can reach 2-3 fC. Its max input detector capacitance is 120 pC and its maximum counting rate is $8 \times 10^5$ cps. FPGA mother board receives signals from LVDS driver and can manage the acquisition parameters like thresholds, integration time width and trigger setting. This board can also be managed remotely through an Ethernet connection. It is important to underline that until now CARIOCA chip can manage 8 channels, so considering that, for this type of detectors, there are 8 CARIOCA cards and that for each card there are 2 chip, the total number of channels is 128. For this reason, there is an inverse proportionality between spatial resolution and the available active area: small pads means an higher spatial resolution, but a small active area.
3.2 A GEM detector as 2D dosimeter

As described in the previous paragraphs, the active volume of the triple GEM detector is represented by the gas region between the aluminated mylar cathode and the first GEM foil: a gap 3 cm thick of ArCO2 mixture. A GEM detector works in photon counting. The first important question to be addressed is if GEM reading can be used as a dose measurement. GEM detector, indeed, is a proportional counter: charged released in the drift region is amplified through the three GEM stages so that the signal produced in the induction region can exceed a given threshold level. At this point, induced signal is registered as one count. In principle, each time a photon interacts within the gas drift region, a count must be registered. Proceeding in this manner, a measurement of the particle fluence could be obtained and, in particular conditions, it can be correlated to the dose. Unfortunately this argument is quite simplistic and somewhat unrealistic. Three are the energy spectra coming from the Synergy LINAC which can work at three accelerating voltages: 6, 10 and 18 MV (Fig. 3.9).

![Figure 3.9: X-ray spectra coming from the Elekta Synergy LINAC when the accelerator works at the three possible accelerating potentials: 6, 10 and 18 MV. Areas under the curve are normalized to 1.](image)

The corresponding photon energies are of the order of few MeV. A slab of gas 3 mm thick is practically transparent to this kind of photon beams. First measurements have been performed with the detector window located at 1 m from the source without any thickness before it. 1 m is the distance separating the source from the linac isocenter and is the standard distance typically used for 2D dosimeters. It was hypothesized that the main photon interactions take place in the in the aluminate mylar foil which has a thickness of 12 µm. At these energies the main interaction mechanism is Compton scattering. So it is the scattered electron produced in mylar which reaches and ionizes the gas. In addition the amount of ionization produced by this secondary electron would be much lower than that which could be produced by a soft X-ray in the same gas layer. For 2D dose measurements, incident beam is perpendicular to the 2D dosimeter, so the main contribution to ionization in the gas layer is due to the photons coming from one side. However there can be another small contribution coming from backscattering radiation due to the back material behind the cavity. For GEM detector, backscattering radiation is considered negligible because the Compton scattering, which represents the main interaction mechanism, produce charged particles (electrons)
in a forward direction with an angle which spans from 0 to 90 degrees. This is what it is expected from simulations.

3.3 Preliminary measurements on the X-ray beam

The X-ray beam coming from Synergy Linac has two specific characteristic which make complicated the use of a GEM detector as a dose monitor:

- the beam is pulsed with a very high particle fluence per pulse of $8.5 \times 10^{10}$ ph/(mm$^2$sec),
- photons have energy spectra in the range of gamma photons.

While the first one cause pile-up of particles, the second means a very low probability of interaction. At first sight, there was no clear idea on how perform a measure in these conditions, because typical operating parameter of a GEM detector (in particular, drift fields and high voltage gain) could be not correct in this case. The first idea was to use an already prepared standard GEM detector and then optimize its characteristics for the present case.

3.3.1 The first GEM detector measurements

In order to verify the applicability of a triple-GEM detector for MeV energy photons, a GEM detector having an anode with 16 x 8 pads each one with an area of 3 x 6 mm$^2$ has been used. Clearly pads large like these cannot satisfy high resolution requirements for 2D dose measurements as was underlined for conventional ionization chamber arrays in chapter 2. These tests permitted to understand the optimal working parameters for these type of particle beams. Typically for ArCO$_2$, gain voltage can range between 900 and 1400 V, while the field between the gaps are set to 3.0, 3.0, 3.0 and 5.0 kV/cm beginning from the drift region and ending with the induction region. Like in the other application of this type of GEM detector, gas mixture inside the detector set at atmospheric pressure and the gas flux is constantly monitored by means of a flowmeter. The value applied oscillates between 60 and 65 cc/min, a value which has maintained also for the other detector configuration developed in this work. It must be highlighted that, in all cases, no observable effects on measures have been observed due to temperature and humidity variations of the gas mixture in the ordinary working ambient conditions. A first measurement was performed on a 6 MV X-ray beam having an area of 2.4 x 2.4 cm$^2$ and a PRF of 50 Hz setting a GEM voltage gain of 1120 V and the gap fields at the standard values. 2D counts distribution is shown in fig. 3.10. In the present work, X-ray spectra coming from the Linac are identified by the corresponding accelerating potential of electrons.
As a result, it can be observed that central region which corresponds to the square beam area underestimates the counts. It would be the area where counts would assume maximum values. It can be observed that the standard working parameters of a triple-GEM detector are not the optimal choice to obtain the correct measure on these high energy photon beams. Reducing the gain voltage until 940 V, 2D counts distribution becomes more realistic in the sense that it reproduces the fluence distribution of the beam, but counts are few and subject to high statistical fluctuations. So the next step was to rise the gain voltage to 960 and reduce progressively the applied drift field (fig. 3.11).

On the left of fig. 3.11, there is the 2D counts distribution obtained with a drift field of 0.5 kV/cm, while on the right the applied drift field is 0.2 kV/cm. It is clear now that counts distribution is more realistic in order to represent photon beam fluence. At the end, a good result was obtained rising the gain voltage in order to get more counts (and then few statistical fluctuations) and setting the drift field to zero. But what are the physical phenomena which justify these observations? It is necessary to analyze the processes which determine signal formation inside the detector.

### 3.3.2 Signal analysis

As described previously photon interactions take place principally in the mylar foil and, to a lesser extent, in the gas drift layer. As a result, a field of secondary electrons is produced. According to Compton scattering, it is expected that each electron is emitted with a certain angle and the probability to be scattered at a small angles is higher than that at large angles. At the same time,
forward scattered electrons are more energetic. Then it is important to consider the pulsed structure of the beam. At each pulse the estimated value of the particle fluence is about $8.5 \times 10^{12}$ photons/(cm$^2$ sec) while the pulse lasts 3 µs, then a very high fluence concentrate in a small time interval. Consider the 6 MV spectrum which has a maximum at 1.50 MeV. A mono-energetic photon beam pulse which interacts with a mylar foil 12 µm in thickness would produce about 20 electrons/mm in 3 µs and if the interaction with the drift gas layer is considered, this estimated number must a little more. This means about one electron every 100 ns. On the other hand, the induced signal on the pad depends on the charge reaching the pad and the drift velocity in the induction region. According to Ramo’s theorem, it is expected that each electron emerging from the last GEM induces a rectangular current signal in the nearest pad with a width dependent on the time spent $t$ by the electron to cross the induction gap:

$$i = -\frac{q}{t} = -\frac{q v_d}{t}$$

where $q$ is the moving charge and $v_d$ the electron velocity in the gap. However, in this case, the total current pulse induced on a pad is the sum of the direct and the cross induced currents [52]. The cross induced component is mainly related to electrons that will be collected along the drift region, while the direct component is due to the time that the electron avalanche leaving the third GEM takes to reach the anode. If the trajectory of a secondary electron is contained entirely inside the volume projected over a pad (fig 3.12a), there will be a pulse signal which will be non zero until the electrons originated from the primary ones closest to the cathode are collected.

![Figure 3.12](image.png)

*Figure 3.12: schema of the electron trajectories respect to the gas columns identified by PADs (left) and drift velocity vs electric field for ArCO$_2$ (70/30) gas mixture (right).*

The current pulse will have a time width which is the sum of drift time in the drift region $t_d$ and the induction region $t_i$ (fig. 3.13). For ArCO$_2$ gas mixture drift velocity in the drift and induction regions which usually works with drift fields of 3.0 and 5.0 kV/cm is about 7.0 and 7.5 cm/µs, respectively (fig 3.12b).
Figure 3.13: simulated pulse time width on a read-out PAD in a triple-GEM detector: direct ($t_d$) and cross ($t_i$) induced currents [54].

This means that the sum $t_d + t_i$ is about 60 ns, which is the total duration of the induced current pulse. Taking into account that a photon is expected every 100 ns, there is a higher probability to have 0 counts or 1 count, according to Poisson distribution. In order to have a more realistic description of X-ray interaction a Monte Carlo simulation has been performed using the FLUKA code. Simulation takes into account the real experimental conditions, in particular the X-ray spectrum, the angular aperture of the source, its distance of 1 m from the detector and the presence of the air between detector and source. Fig. 3.14 shows a 2D plot of the X-ray beam as result from simulation. The beam comes from a point source and widens in a pyramidal way in order to obtain the square shape of $2.4 \times 2.4\, \text{cm}^2$. This takes into account the effect of multileaf which cut the beam to the required shape.

Figure 3.14: side view of the simulated photon beam irradiating the GEM detector with an area of $2.4 \times 2.4\, \text{cm}^2$.

The GEM detector has been reconstructed like a series of layers: mylar, drift gas, GEM1, T1 gas, GEM2, T2 gas, GEM3, induction gas and PCB. GEM foils where simply reconstructed like a kapton foil with two layer of copper on each side. No hole were considered: their effect is to reduce the real interaction region which can be taken into account considering the optical transparency of the GEM. Their role would be more important in the simulation of the electric fields, but this calculation is not possible with FLUKA and will be considered later. Fig. 3.15 shows the photon and electron distribution per $\text{cm}^3$ and per primary photon in the region where the detector structures are present.
Plot in Fig. 3.15a shows clearly that photon beam attenuation when it crosses the detector is considerably reduced and this means that x-ray absorption from detector is greatly reduced at these energies. According to simulations, the estimated photon absorption from mylar window and gas drift layer is about \((1.0 \pm 0.8) \times 10^{-4}\) absorbed photons per primary. Nonetheless it registers a significant signal, like first experimental results have demonstrated. The detector is working at very low efficiency. Then his very high sensitivity is confirmed also to these energies. A clear result can be evidenced from a 1D plot in which the electron and energy distribution are plotted along a column having a square base of 1 cm\(^2\) and centered on the beam axis.

Each bin of the previous plots is 100 µm. Then the number of steps along the gas column in the drift region is 30 and, as can be read from the first plot, the number of electrons/cm\(^3\)/pr at each step is about \(5 \times 10^5\). As a result, considering the number of primary photons \((1.5 \times 10^8\) ph) calculated for a 2.4 x 2.4 cm\(^2\) beam in a similar way as seen in par. 2.4.3, the number of electrons entering in the gas is about 7500 e/cm\(^2\), which means about 1300 e\(^-\) per PAD (3 x 6 mm\(^2\)) and a time between two particles of the same pulse equal to about 2-3 ns. This value is lower than that estimated previously. Then when a pulse like this is induced on the pad, there is an higher probable that more than one electron crosses the gas column on the PAD. Ionization charge produced by these other electrons overlaps with that of the first one and the resulting pulse is greater in amplitude but is registered as one. A look to fig. 3.12a permits to explain how current pulses are generated for the simple case of
five pads. Suppose that the incident beam had a fluence concentrated on the central pad (the darkest one). Electrons are emitted at different angles and not all trajectories are localized inside the volume identified by a pad. Most of them intersect more than one pad and the ionization charge is released only by a segment of the whole trajectory. Tracks crossing the drift volume corresponding to the central pad are more than those crossing the nearest ones. So the central would provide a count equal or proportional to the number of track passing on it, but this is not the case: all ionization produced by the electron tracks drifts toward the pad inducing a large current pulse which lasts for about 60 ns as minimum, accounting for the electron drift velocity in ArCO₂ gas mixture. This pulse is registered as one count. On the nearest pads there are few tracks and there is a smaller probability that the relative ionization charges overlaps so that registered counts are more close to the real number of tracks. When the drift field is lowered not all the ionization produced along the drift regions is collected so that it is much probable that two consecutive ionization tracks can be seen as separate events. The plot in Fig. 3.16b shows the energy released along the same region specified for electrons. It is clear that energy is practically uniform in the drift region. This means that also the corresponding produced ionization is uniformly distributed. When the drift field is reduced, only a fraction of the total charge released along the drift region is collected. When the limit of zero drift is reached, the electric field is localized only in close proximity to GEM holes. Fig. 3.17 shows a simulation of the electric field near the GEM holes obtained with GARFIELD.

Field lines are closed on the holes and passes through a limited volume inside the drift region (Fig. 3.9a). The variation of the electric field E with z along the line perpendicular to the GEM and passing through the central point of the hole is reported in fig. 3.9b and the same is shown magnified in fig. 3.9c. As can be noted the electric field extends in the drift region for about 100 µm and it is this layer which now represents the active volume. In fact all the charges released in the volume which is not interested by the field recombine. The segments of electron trajectories near the holes contribute to charge collected in the GEM hole and then to the signals. This reduces considerably the cross induced current contribution (see fig. 3.13) so that pulse time width is determined mainly by the direct current and the probability of pile-up is considerably reduced. Each photon interaction produces a scattered secondary electron that can reach e GEM hole and produce a signal. Considering that now the real active volume is reduced to 1/30 of the original one, the number of estimated electrons is about 40 and then a time between two consecutive particles equal to 70 ns, higher than the time needed for induce current pulse time width. This explains the more realistic counts distribution but does not means that pile-up does not appear. Ionization charge produced by a single electron passing in the active volume can be different depending on its

---

**Figure 3.17**: a lateral section of a GEM hole with the field lines as calculated from Garfield (left), electric field profile from the drift to the first transition region (middle) and a magnification of the same profile from Garfield simulations.
trajectory and energy. A simulation in terms of energy will be presented in par. 3.4.2. Clearly the
electron trajectory can also end on an area of the GEM without holes and does not contribute to the
signal. In other words, the detector works at a very low efficiency and this represents an advantage
because, as described in Chapter 1, in radiotherapy, typical beam fluencies are of the order of $10^9 –
10^{12}$ photons/ cm$^2$sec, which is behind the functionality limits of GEM detector with these FEE
electronics and the usual electric field configurations.

3.3.2 First prototype measurements

Previous tests provided some important guidelines to set up a triple-GEM detector for dose
measurements on therapeutic photon beams. Taking into account the first experimental results
outlined before and the conclusions reached in the previous paragraph, a new detector was realized.
It has the same geometry of the first one, but now the anode has a row of 128 aligned PADs with an
area of 500 x 500 µm$^2$ and a pitch of 700 µm. This layout meets two basic requests:

- an higher spatial resolution,
- and a smaller PAD area which further reduces the pile-up phenomenon.

However, in this configuration, it would be possible to obtain a beam profile along only a given
coordinate. According to the nomenclature used in radiotherapy, the intersection between the beam
axis (Z axis) and the axis around which the accelerator rotates (Y axis) is called “isocenter”. Y axes
is called “in plane” direction, while the axis perpendicular to it and passing though the isocenter is
called “cross plane” direction (fig. 5.7).

GEM detector was positioned so that its Source-Surface-Distance (SSD) was again 100 cm (the
standard value) and the PADs raw was located along the in plane direction. In this case the detector
mylar window was exposed without any thickness of tissue equivalent material in contrast to what
happens for conventional 2D dosimeters (matrix arrays and films). Detector was irradiated with a 6
MV X-ray beam having an area of 2.4 x 2.4 cm$^2$ in order to be fully covered by the detector window
(10 x 10 cm$^2$). In this configuration detector provided as a measure a count distribution which could
be related to the dose profile along the in plane direction. Performed measures demonstrated that it
is possible to obtain a dose distribution in a slice 500 µm thick with a pitch of 700 µm, a spatial
resolution at least an order of magnitude higher respect to the commercial available devices. Fig. 3.19 shows the profile obtained on the pads row with an integration time of 500 ms, applying a zero drift field, a gain voltage of 1100 V and a threshold of 1600 mV.

![Figure 3.19: the PADs row central profile fitted with a double-Boltzmann curve.](image)

Red curve is the double-Boltzmann (d-B) function fit. In the following, it will be often used for the slices of 2D beam profiles, in order to assign some quantitative parameters related to the measured profiles. The d-B function is defined by the following expression:

\[
y = \frac{A_1 - A_2}{1 + e^{(x-x_{01})/dx_1}} + \frac{A_2 - A_1}{1 + e^{(x-x_{02})/dx_2}} + A_2
\]

where \(A_1\) and \(A_2\) are constants related to the function amplitude as shown in the following plot of a d-B function.

![Figure 3.20: characteristic parameters of the double-Boltzmann function.](image)

\(dx_1\) and \(dx_2\) are related to the curve slope in the points \(X_{01}\) and \(X_{02}\), according to the following relations:

\[
y'(X_{01}) = \frac{A_1 - A_2}{4dx_1} \quad \text{and} \quad y'(X_{02}) = \frac{A_2 - A_1}{4dx_2}
\]

The difference \(X_{02} - X_{01}\) can be considered as a measure of the beam profile width. All these parameters will be used to match profiles measured in different conditions, at least for these first measures. A 2D beam profile is obtained mounting the detector on a linear stage, in order to make a scan along the cross direction. Linear stage is a commercial translational motor (TM): model VT-80 from PImicos® (fig. 3.21).
Figure 3.21: Plmicos linear stage model VT-80 (left) and the same linear stage with the mounted GEM detector (right).

It can move on a range of 10 cm with a resolution less than 1 µm and a velocity range which goes from 0.001 to 20 mm/sec. A LabVIEW© program manages detector scan acquisitions integrating FPGA and TM controls.

Figure 3.22: Labview Control panel for the scan GEM detector. It is possible to distinguish the Threshold Control Panel, the Time Parameters settings, the Linear Stage Control Panel and the Monitor Panel on which the counts distribution is displayed after the scan.

Once integration time has been set, the values of velocity of TM is calculated as the ratio of the difference between the first and the last scan positions and integration time multiplied by 128. The TM moves until all the scan distance has been covered. Fig 3.21 show the counts map distribution obtained making a scan of 128 x 0.7 mm = 89.6 mm (i.e. 128 slices, each one having a width of 0.7 mm) and an integration time of 200 ms for two 6 MV beam (2.4 x 2.4 cm² and 4.0 x 4.0 cm²) delivered at a PRF of 200 Hz (corresponding to 200 MU/min).
Figure 3.23: a scanning acquisition with triple-GEM detector for a 2.4 x 2.4 cm$^2$ and 4.0 x 4.0 cm$^2$ beams at 6 MV with a voltage gain of 1100 V, a threshold of 1400 mV and a time width of 500 ms.

This type of distribution shows that working with zero drift field is an optimal choice, while the use of small pads not only increase spatial resolution but also reduce the phenomenon of vertical signal pile-up increasing the number of counts per integral time. One important point to underline is that not all pad channels have the same sensitivity: at a fixed X-ray fluence and at a given integration time, some register more counts than others. This was verified irradiating the detector with a uniform field having an area greater than 10 x 10 cm$^2$. A series of scans were performed at different threshold values. A result of this scan is shown in fig. 3.24. On the left there are the measured counts profiles along the PADs row obtained as the medium value on all the scanned positions.

Figure 3.24: counts distribution obtained after a scanning on a 10 x 10 cm$^2$ beam at 6 MV with a voltage gain of 1100 and a time width of 200 ms (left); the PADs row profiles obtained as the media on all the positions at different thresholds set on the CARIOCA chips (right).

It can be observed that profiles at different thresholds change in the number of counts but preserve their shape. This represents clearly a systematic error which must be corrected at each working threshold in order to realize a count equalization along the 128 PADs. This was realized performing an horizontal line fit for each threshold and obtaining some correction factors as the ratio between the measured count and the corresponding fitted value. A result of this equalization process is shown in fig. 3.25 for a 4.0 x 4.0 cm$^2$ fields at a threshold of 1400 mV.
Figure 3.25: 6 MV 4.0 x 4.0 cm$^2$ beams before and after the PADs equalization according to the results obtained by a uniform irradiation of all the available scanning detector area.

Applied corrections are intrinsic to the detector used because depends on its pads and all FEE which manage the 128 channels. To reconstruct the real physical signal it is necessary to apply the right correction factors to the measured fluence map and this means in the same working parameters of high voltage gain and thresholds.

3.3.3 Currents measurements

During the scan acquisitions, it was observed a significant rise of the currents measured on the HVGEM nano-ammeter channels (fig. 3.25).

Figure 3.26: Labview HVGEM control panel. The free monitors reports the applied fields, GEM voltages and currents measured by the nono-ammeters on the seven detector electrodes: cathode and GEM caper layers.

This represents an important analogical information to evaluate because it is correlated the corresponding digital counts measurements. As described in the previous chapter actual 2D dosimetry is realized using matrices of small ionization chambers and the dose is measured as the ionization current released in these chambers, hence an analogical signal. The nano-ammeters of the HVGEM module provides a digital measure and, if the detector is working correctly, currents and counts must be correlated. It must be considered that the amount of currents in a GEM detector may
change according to two effects: the triple-GEM gain and the fluence of the incident particles. X-ray beam coming from the LINAC is a pulsed beam with bunches of 3 µs. The estimated number of photons per bunch is $8.5 \times 10^{12}$ ph/(mm$^2$ sec) and this cannot be changed. It is through PRF that dose rate can be set in order to have different number of incident photons per unit time. Counts and currents must follow these variations. The Synergy LINAC has the possibility to set only 7 different values of the PRF: 6, 12, 25, 50, 100, 200 and 400 Hz. Plot represented in fig. 3.27 reports the mean GEM counts collected by a single PAD versus PRFs for two different HV gains (1000 and 1040 V) when the detector is irradiated by a beam greater than 10 x 10 cm$^2$ in area.

![Figure 3.27](image)

**Figure 3.27**: GEM counts on a PAD vs Pulse Repetition Rate at two working gain voltages.

Straight lines used to fit data shows an optimal linear correlation between counts and photon fluence rate. A similar correlation is observed for the corresponding HVGEM currents, in particular the current collected on the upper electrode of the third GEM (G3up), as it was expected. Currents can be varied changing the gain voltage and it was observed that also in this case the number of registered counts follows the voltage applied on the GEMs, as reported in table 1. This means that the gain voltage has a magnification effect not only on the currents but also on the corresponding registered digital counts.

<table>
<thead>
<tr>
<th>High Voltage (V)</th>
<th>Mean [2Hz]</th>
<th>err Mean</th>
<th>Sigma</th>
<th>err Sigma</th>
</tr>
</thead>
<tbody>
<tr>
<td>1060</td>
<td>417</td>
<td>4</td>
<td>22</td>
<td>5</td>
</tr>
<tr>
<td>1070</td>
<td>570</td>
<td>3</td>
<td>24</td>
<td>2</td>
</tr>
<tr>
<td>1080</td>
<td>764</td>
<td>2</td>
<td>20</td>
<td>2</td>
</tr>
<tr>
<td>1090</td>
<td>857</td>
<td>2</td>
<td>20</td>
<td>3</td>
</tr>
<tr>
<td>1100</td>
<td>1070</td>
<td>3</td>
<td>24</td>
<td>2</td>
</tr>
</tbody>
</table>

*Table 1*: GEM voltages scan on a 2.4 x 2.4 cm$^2$ beam at 6 MV with the corresponding registered mean counts and sigmas calculated on the counts collected by PAds.

Plots in graph 3.28 shows the measured currents on the cathode, G1 up and G3 up when the HV gains of table 1 were applied. As expected, current on cathode remains at the low values presented when the applied HV is off, while different current values are observed on G1 up and G3 up. GEM electrodes (in particular G3 up) collect a lower current when GEM gain decreases.
Figure 3.28: HVGEM currents measure on the Cathode, the GEM1 up and GEM3 up electrodes.

The corresponding count distributions are reported on the following histogram plot (fig. 3.29) where a Gaussian fit was applied. The central value of each distribution is reported in table 1.

Figure 3.29: counts distributions when the applied gain voltage is varied between 1060 and 1100 V at step of 10 V.

The plot of fig. 3.29 shows clearly a linear correlation between the analogical current and the digital registered counts. The previous measurements emphasizes that a GEM detector has a great dynamic range, even with high energy X-rays like those coming from a medical linac: if the fluence of the photons irradiating the detector is high enough to trigger signal pile-up and cause a resulting decrease in the integrated number of counts, it is possible to act on the HV gain to reduce the amount of current produced and then obtain a correct count distribution which reflects real 2D
fluence distribution coming from the source. However there is only a partial advantage in this direction because, in the present case, photon fluence is not continuous but pulsed and pile-up is due mainly to pulse time width and to the photon fluence per pulse, rather than to the medium fluence per second, as explained in 3.3.2. Working at low gains, it is possible to reduce the number of current pulses detected at each beam pulse, but it is not possible to distinguish two or more particles falling in the time width of an induced current pulse.

### 3.3.4 Threshold scan measurements

As explained previously, the amplitude of the current pulse induced on a pad is proportional to the amount of charge released in the gas, which, in turn, is proportional to the energy released in the gas. A scan in threshold from low to high values will allow to cut all the signals at low energy. A measure of this type has been performed on a 2.4 x 2.4 cm$^2$ X-ray beam with an HV gain of 1100 V and an integration time of 1 sec (fig. 3.30).

![Figure 3.30: 2D counts distribution for a 6 MV 2.4 x 2.4 cm$^2$ field acquired at 1100 V of gain voltage and 1200 mV of threshold (left) and counts profiles obtained on the 64° PAD at different thresholds (right).](image)

Then a 1D beam profile was taken at the ordinate value $Y = 32$ cm for all the 2D profiles at different threshold values (fig. 3.30). Again a double Boltzmann fit was applied on the measured profiles and their fit parameters are summarized in table 2.

<table>
<thead>
<tr>
<th>THR (mV)</th>
<th>$A_1 \pm \text{err (Hz)}$</th>
<th>$A_2 \pm \text{err (Hz)}$</th>
<th>$X_{01} \pm \text{err (mm)}$</th>
<th>$dx_1 \pm \text{err (mm)}$</th>
<th>$X_{02} \pm \text{err (mm)}$</th>
<th>$dx_2 \pm \text{err (mm)}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>2000</td>
<td>-365</td>
<td>763</td>
<td>2</td>
<td>24.5</td>
<td>0.2</td>
<td>3.07</td>
</tr>
<tr>
<td>1600</td>
<td>-440</td>
<td>874</td>
<td>3</td>
<td>24.5</td>
<td>0.2</td>
<td>3.12</td>
</tr>
<tr>
<td>1400</td>
<td>-673</td>
<td>1010</td>
<td>3</td>
<td>24.3</td>
<td>0.1</td>
<td>3.18</td>
</tr>
<tr>
<td>1200</td>
<td>-1336</td>
<td>16244</td>
<td>6</td>
<td>24.0</td>
<td>0.1</td>
<td>3.09</td>
</tr>
</tbody>
</table>

*Table 2: double-Boltzmann parameters obtained by the fit on the profiles of fig. 3.30.*

Results shows that parameters $X_{01}$ and $X_{02}$ are consistent for differ thresholds and the same happens for $dx_1$ and $dx_2$. This means that beam profiles preserve their shape, the only changes concerning their amplitude. According to these results, it can be stated that the energy distribution released per unit incident fluence is uniform on all the sensitive area, otherwise regions corresponding to high...
energy and then higher induced pulses, would be emphasized with a resulting profile distortion. It will be demonstrated that a FEE able to perform charge measurements can improve considerably the triple-GEM measures for these particular application (chapter 4). A correlation plot between the profile counts at 2000 mV threshold and all the others allows to point out a further information related to the fluence intensity (fig. 3.28). At high thresholds, indeed, the few counts are registered and no pile-up phenomena are expected. When the threshold is decreased (or, alternatively, the gain is increased), counts must increase both on the central area and on the side areas where fluence drops to zero. Then a linear correlation is expected between lower threshold and all the others. This is the reason of the plots reported in fig. 3.31.

![Figure 3.31: correlations between counts obtained at the threshold of 2000 mV with the counts of all the other lower thresholds.](image)

Each plot was linear fitted and an optimal $\chi^2$ was obtained in all cases. However plots at 1400 and 1200 mV threshold show slight tilt to low values in last range of high counts. This is due probably to a beginning of signal pile-up with a consequent reduction on counts for low thresholds. To preserve linearity it is necessary to work at high threshold or, in alternative, to reduce the HV gain. Once again GEM detector show an optimal flexibility regarding its dynamic working range.

### 3.4 GEM detector and gafchromic detectors

Results presented in the previous paragraph have shown clearly that a GEM detector can provide some counts distributions also with high energy X-rays and that these distributions are closely related to the charge released in the detector. To make this type of measure it has been clarified that it is necessary to work with zero drift field and with small pads. This second condition satisfies also the request of an higher spatial resolution, one of the main objective of the new detector. However FEE electronic has a limited number of channels and the most immediate solution is to make a scan system. Next step is to verify that the measured counts can be correlated to the dose distribution. Typically quality controls in radiotherapy require that the dose distribution maps planned with TPS and the actually dose irradiated areas overlap each other, in terms of relative intensities and shapes. This means also that a 2D device must first be calibrated respect to a reference point dose provided
by reference dosimeter. In a typical 2D control check, 2D matrices are placed at a standard distance and covered up and down with slabs of tissue equivalent material. They are inserted not only to simulate the presence of water in standard calibration conditions but also to have a sufficient build-up necessary to increase the amount of ionization produced in the cameras. At the same time, TPS takes data coming from a CT of the measuring system (2D dosimeter + slabs) in the same conditions applied during dose irradiation. CT data are used from TPS to make a calculation of 2D dose distribution on the plane where the matrix of ionization chambers measure. Afterwards calculated and measured dose distributions are compared through the gamma index test which must be verified on more than 90% of activated chambers. Dose measured by a 2D matrix is limited to the chamber pitch which, in the best case, is few mm. Often intermediate values are interpolated. As an alternative to the 2D matrices, gafchromic films can be used. This time spatial resolution is increased to few hundred microns, but now it need of a post-processing analysis before comparison with calculated doses. A similar argument can be applied to the GEM detector. However it will be demonstrated that, for a GEM detector, this procedure cannot be applied and the best way is to realize a relative dose checking between the GEM detector and a gafchromic film. The standard procedure described previously represents the next step to demonstrate that a GEM detector can be used as a 2D dosimeter in typical quality controls in radiotherapy.

3.4.1 GEM detector and tissue equivalent material

As shown in the previous chapter, GEM detector is very sensitive and can work without converters. However, the possibility to have a tissue equivalent material must be considered in order to have irradiating conditions similar to those used for ionization chamber matrices in 2D dosimetry and then to apply a similar procedure to compare it to TPS calculations, like explained previously. In the present case, preliminary results have shown that detector can undergo to counts saturation when particle fluence is very high and this means high secondary electron fluence producing very long signal currents. It has been evaluated the possibility to make a TPS calculation on a GEM detector without tissue equivalent material. For this purpose the detector has been CT scanned and data sent to TPS for calculation. As a result, however, calculated dose was wrong. TPS is not able to make correct estimates in the sensitive volume of the GEM detector. TPS, indeed, is not correctly calibrated for the calculation of released doses in air. It is necessary to have at least a tissue equivalent material before the detector window. However working with this conditions can increase the number of secondary electrons to values which cannot be managed correctly by FEE giving rise to the already described phenomenon of counts pile-up. In the present case it was observed that the GEM detector with a 6 mm thick slab of PMMA before can be correctly processed by TPS. Then some tests have been performed in this configuration. The irradiating beam is the usual 2.4 x 2.4 cm$^2$ in area delivered at 200 MU/min at an accelerating potential of 6 MV. As a result, detector begins to saturate. Despite the current measured by HVGEM module is not greater than 30 µA, detector underestimates counts corresponding to the center of the beam. Clearly PMMA works like a converter before the detector producing an amount of secondary electrons which piles up and saturate the detector channels. In the following paragraph a brief fluka simulation, for a 6 mm PMMA slab, is presented in order to make an approximate evaluation of the amount of ionization produced in these conditions.
3.4.2 Fluka simulations

One of the goals of the following Fluka simulation is to make an estimate of charged produced in the active volume with and without a converter material. Among the most used tissue-equivalent materials there is an adapted form of Polymethylmethacrylate (PMMA). In the present case conventional PMMA has been used because it has dosimetric characteristics very similar to water. A first simulation was obtained considering a point source at 100 cm from the detector and a pyramidal beam with a given angular aperture. The projected area on the detector window is $2.4 \times 2.4 \text{ cm}^2$. It is equal to one of the typical irradiating fields used in the first measurements. Photons have the spectrum distribution of a 6 MV accelerating potential and are considered perpendicularly emitted respect to the source plane. Detector was simulated with a series of material slabs: mylar foil, drift gas volume, first GEM foil and so on until to the PCB anode 2 mm thick. These layers are in the XY plane. Binning along the X and Y direction has been set to 0.1 mm. Using these settings, profiles along the Y direction passing though the center of the window has been calculated. In particular the calculated profiles are relative to the electron and energy released in the drift volume. In the following figure there are the calculated 2D map for electrons and energy distributions inside the 100 µm thick gas layer on the first GEM foil when the detector window is exposed directly to the beam and when a 6 mm thick slab of PMMA is placed in front of the detector window.

![Figure 3.32: simulated electron (up) and energy (down) distribution in the gas layer (100µm) on the first GEM generated by 2.4 x 2.4 cm² beam at 6MV when the detector is exposed free in air and when there is a 6 mm PMMA slab.](image)
Color scales express the number of electrons and the released energy (GeV), respectively, per cm$^3$ and per incident primary. Now let consider the thin gas layer (100 µm) over the first GEM. Restricting the Y coordinate to the single bin between 0 and 0.7 cm (the pitch between two consecutive PADs), it is possible obtain a an electron and an energy profile along the X coordinate. Fig. 3.31 shows the results regarding the detector with and without 6 mm thick PMMA.

![Figure 3.33: simulated electron distribution profile obtained with an irradiating X-ray beam 2.4 x 2.4 cm$^2$ without (left) and with 6 mm PMMA slab. Electron distribution is calculated along a strip of gas 700 µm wide and 100 µm thick on the first GEM foil.](image)

Fig. 3.33: simulated electron distribution profile obtained with an irradiating X-ray beam 2.4 x 2.4 cm$^2$ without (left) and with 6 mm PMMA slab. Electron distribution is calculated along a strip of gas 700 µm wide and 100 µm thick on the first GEM foil.

![Figure 3.34: simulated energy distribution profile obtained with an irradiating X-ray beam 2.4 x 2.4 cm$^2$ without (left) and with 6 mm PMMA slab. Both plots represent the energy released in a strip of gas 700 µm wide and 100 µm thick on the first GEM foil.](image)

Fig. 3.34: simulated energy distribution profile obtained with an irradiating X-ray beam 2.4 x 2.4 cm$^2$ without (left) and with 6 mm PMMA slab. Both plots represent the energy released in a strip of gas 700 µm wide and 100 µm thick on the first GEM foil.

According to the these results, the energy released in a volume of gas 700 x 700 x 100 µm$^3$ can be estimated. Considering the values coming from the case with no PMMA, it can be observed that the typical values on the central beam area, even if with great uncertainties, varies in a range between 8x10$^{-11}$ and 1.5x10$^{-10}$ GeV/cm$^3$pr. The estimated fluence rate for a single pulse is 8.5 x 10$^{12}$ ph/cm$^2$s, so considering that each pulse last for 3 µs, the number of photons per pulse for a beam of area 2.4 x 2.4 cm$^2$ is 1.5 x 10$^8$ ph. So considering the mean deposited energy equal to about 1.1 x 10$^{-10}$ GeV/cm$^3$pr, the total energy released by a single pulse in a volume with pad base area and a an heigh of 100 µm is about 81 keV with oscillations which can range from 59 to 110 keV, according to the extreme values considered above. This means that fluctuations can range with a sigma of 20 % around the central value. For ArCO2 gas mixture, the minimum energy needed to create a electron-ion couple is about 32 eV so that the mean number of ioniztion electrons is 81000/32 ~ 2500 e'. If the mean time for a induced current pulse is 60 ns and the beam pulse last 3 µs, the
mean charge collected for each pulse can be estimated as $2500 \ e^{-} / (3 \ \mu s / 60 \ ns) = 50 \ e^{-}$. This limit corresponds to countings proportional to photon fluence, as observed on the measured fields. When PMMA is before the detector both the number of electrons and the released energy increase of one order of magnitude. Then mean time between two consecutive electrons reduce to $1/10$ respect to the case with no PMMA (70 ns, par. 3.3.2) and charge is increased of 10. In these conditions pile-up becomes much more probable. So it was preferred to perform measures with no tissue equivalent materials before the detector window.

3.4.3 Comparison with GAFchromic films

It is clear from the previous results that a GEM detector cannot be used with converter materials like those applied for the conventional devices (matrix arrays and gafchromic). However, it could be calibrated with respect to an absolute dose value and its dose measures compared with those calculated from TPS. At the moment, the simple way is to compare the GEM measures with those given from another 2D reference dosimeter used in the same conditions with no tissue-equivalent material like the GEM detector. Matrices are limited for two main reasons: each chamber is not enough sensitive to measure the small doses without converter and its spatial resolution is limited respect to GEM detector to make a valid comparison. The ideal comparison dosimeter turns out to be the gafchromic film. It has certainly an higher resolution respect to GEM detector and can also be used without converter. To obtain a sufficient blackening, however, it must be irradiated for a longer time respect to the usual applications. In this case the measure is no longer absolute, but relative. This is satisfactory because a film can always be calibrated in absolute dose in these conditions. Now it is important that in relative measurements the present detector can perform measures comparable with a standard detector routinely used in clinical practice. In this paragraph measures with GEM detector on a 6 MV X-ray beam are presented. Then measures compared with those obtained with gafchromic films performed in the same experimental conditions of GEM detector.
3.4.4 Measurement set-up and experimental results

GEM detector was mounted on a liner stage which is fixed to an aluminium structure. This structure has been positioned on the bed of the Linac treatment unit. Detector window is positioned to the isocenter and is orthogonal to the Z irradiation axis (fig. 3.33).

![Figure 3.35](image)

Figure 3.35: the main elements of an Elekta Synergy radiotherapy unit with the ISOcenter definition and the location of the GEM detector on the bed ready to be irradiated.

Source-to-Detector-Distance (SDD) is 100 cm, the standard value. Detector has been irradiated with square fields of 2.4 x 2.4 cm$^2$ and 16.0 x 21.0 cm$^2$ at 200 MU/min. The applied GEM high voltage gain has been set to 1100 V while acquisitions have been made with an integration time of 500 ms and different thresholds. As shown in par. 3.3.4 detector counts are sensitive to threshold and this can vary the number of particle registered in a significant manner. The large irradiation field, instead, has been performed in order to equalize the systematic errors of each channel, because their response is not uniform (par. 3.3.2). Fig. 3.36 shows the original measures obtained by the GEM detector for two different values of the thresholds (1200 and 1600 mV) together with the corresponding gafchromic measure.

![Figure 3.36](image)

Figure 3.36: GEM detector scans for a 6 MV 2.4 x 2.4 cm$^2$ irradiation field acquired at thresholds of 1200 and 1600 mV ($V_{GEM} = 1100$ V) and the corresponding measure of gafchromic obtained after an irradiation of 1000 MU.

Fig. 3.36 shows the 2.4 x 2.4 cm$^2$ fields before and after equalization for the three available energies.
These equalized maps are then compared with the dose distribution acquired with GAFchromic film. For each field the film has been placed on the detector window in the same measurement conditions of GEM detector and irradiated with 1000 MU, a value which assured a significant blackening without signal saturation. Dose rate was 100 MU/min and then 10 min of irradiation. Once irradiated films have been scanned with a Scanner EPSON 10000XL which typically is used at the Tor Vergata Radiotherapy Center. It has been tested in order to provide a uniform response on a precise area so that scanned films are not affected by non-uniform regions of the scanner. Data provided from the scanner are TIFF files from which a numerical matrix has been obtained. Each value is proportional to the GAF opacity so that, in the regions of more blackening, numerical values are lower than those on the sides which were not on the beam. If $G_{\text{back}}$ is the medium value on the side areas away from the central beam and $G_m$ is the measured value of the blackened regions, the considered values for comparison with GEM detector are $G_g = G_{\text{back}} - G_m$. Then normalized values $G_N$ for the GAF are given as:
\[ g_N = \frac{(g_g - g_{gmin})}{(g_{gmax} - g_{gmin})} \cdot 100 \]

where \( g_{gmin} \) and \( g_{gmax} \) are the minimum and the maximum value of the GAF distributions \( g_g \). A similar normalization has been performed on the GEM data. For the GEM detector the normalized values \( C_N \) are calculated as follows

\[ C_N = \frac{(C_m - C_{min})}{(C_{max} - C_{min})} \cdot 100 \]

where \( C_m \) are the measured counts while \( C_{min} \) and \( C_{max} \) are the minimum and maximum values of the counts distribution. More often distribution presents some long tails out of the beam area not interesting from a dosimetric viewpoint. Many analysis programs apply a low threshold to cut these tails. Typically the criterion is that it must be 7% of the maximum value. Others apply also a percentage of 10% or more. The following figures (3.40, 3.41 and 3.42) show the results of gamma index comparison between 2.4 x 2.4 cm\(^2\) beams measured with GEM detector at three different threshold values (1600, 1400 and 1200 mV) and gafchromic which, according to gamma index test, represents the reference dose distribution. For both doses distributions (GEM and gafchromic), a cut of 10% is applied before calculating gamma test map. Again dose distribution equalizations and gamma index calculations have been performed by an algorithm developed through ROOT[55]. For the gamma index maps, values minor than 0.1, which correspond to the white area, are not plotted.

Figure 3.40: UP: normalized GEM detector (zero electric drift field) and EBT GAF dose distributions measured on a 2.4 x 2.4 cm\(^2\) field at 1200 mV of threshold with the gamma index map; DOWN: X and Y central profiles of gafchromic (red line) and GEM.
Previous results show a good matching in the central area where photon fluence is uniform. However gamma test shows that a significant dose mismatch is observed especially along the beam.
sides where there is a significant dose gradient. As a result, gamma index acceptance test does not reach the required 90% of validity on the detected area. It can be observed that when the threshold is increased from 1200 mV to 1400 and 1600 mV gamma index test provides percentage 10% higher. This effect is reflected on the high dose gradient tails around the central beam area and this is observed especially along the X coordinate which identifies the moving direction of leaves. The beam cut along the side orthogonal to leaves is so clear like the other direction. At low thresholds, GEM detector is more sensitive to X-rays coming from the side orthogonal to leaves and this partially explains the improvement in gamma index along the X coordinate. In addition, there is also the effect of GEM saturation that becomes evident at 1200 mV of threshold (fig. 3.31). However gamma mismatch does not disappear completely and persists along the Y coordinate also at higher thresholds. The general overestimate on the tails can be ascribed to the higher sensitivity of the GEM detector respect to gafchromic. According to simulations presented in 2.4.3, the minimum estimated dose delivered per pulse to the gafchromic active layer is about $6.5 \times 10^{-4}$ cGy, then, as the minimum sensitive value is 1 cGy, about 1600 pulses are necessary to reach this minimum value. When gafchromic is irradiated with 1000 MU the maximum estimated dose is 40 cGy. This means that the measured dose distribution presents a cut on the tails which corresponds at least to 2.5% (= $1/40 \times 100$) of the maximum value. GEM detector, on the contrary, is sensitive to the single pulse and at minimum it can provide also a single pulse which respect to the maximum value registered is few percentages. Taking into account that the maximum registered value at 1200 mV of threshold is about 1800, this percentage lower than 0.1%. However, previous results underlines clearly that a triple-GEM with this read-out electronic allows to make measurements in a counting mode with an higher spatial resolution respect to ionization chamber arrays. In addition the time needed to make a scan, in this case, is 500 ms x 128 = 64 sec, against the 10 min needed for the gafchromic. These large differences in acquisition time must be considered in relation to the observed statistical fluctuations. GEM detector works in a counting mode and then counts acquired in low integration times are affected by large statistical fluctuations. The larger integration time applied to the measures performed with scan GEM detector is 1 sec. Then, if gafchromic values are rescaled to 1 sec of integration, estimated fluctuation is 10%, against the 6.0% measured for the GEM detector acquisitions at 1600 mV of threshold, which are most the pronounced ones.
Chapter 4

Introduction

In the previous chapter, preliminary results with GEM detector have been presented. They provided some important conclusions about a triple-GEM detector as a 2D dosimeter on radiotherapy control checks: an higher spatial resolution respect to matrix arrays and an higher sensitivity which allows to obtain 2D dose distributions in few tens of seconds against minutes of gafchromics. However gamma index test indicates that there is not yet an optimal dose matching on the area with high dose gradients. This means that a digital counting electronics like that presented in the previous chapter cannot be the optimal choice for high dose gradients. A valid alternative is represented by a readout electronics able to perform a measure of the charge released in the active gas layer. Carioca chips are not able to provide a measure of the charge, a quantity related more closely to the released dose. In this chapter a new detector is proposed. It is an hybrid between a triple-GEM and a Medipix [56, 57,59]: it has been named “GEMpix” [58]. Now the FEE is that of quad-medipix [60], four medipix chips hold together without the silicon converter and this will be the new read-out electronic for the new prototype of triple-GEM detector. Active material of medipix chips usually is silicon layer; in the present case it is substituted by a gas detector with a triple-GEM. This configuration allows to exploit the high capability of medipix chips including also the possibility to realize charge measurements. In addition spatial resolution is considerably increased, because a quad-medipix has 512 x 512 pixels having an area of 55 x 55 µm². On the contrary, the active area is reduced to only 2.8 x 2.8 cm². So presented measures have been performed on small fields which can be applied for small localized tumors. Usually in radiotherapy larger irradiating fields are used. However these results shows that this hybrid detector can be used for 2D dosimetry on small fields. Relying on these results, future developments are desirable. The first part of this chapter begins describing the medipix chip version Timepix on which is based the new FEE. Then a description of the GEMpix construction follows. In the second part there are a series of preliminary measures aimed to characterize the detector for this particular application. After dosimetric measures are presented in comparison with those obtained with gafchromic films. A discussion on the results accompanies the presented results.

4.1 Timepix

Timepix [62] is one of the last developments of Medipix collaboration. It is an hybrid pixel detector. As it is known, a pixel detector is a 2-dimensional matrix of microscopic (<<1mm) sensitive elements connected to its own pulse processing electronics. Typically one possibility of designing pixel detectors is that the sensor material and the readout electronics are processed on the same substrate. This is known as monolithic pixel detector. The hybrid approach, instead, consists of two consecutive steps: processing of the sensor and the readout chips on different substrates and
connecting them to obtain the imaging system. Figure 4.1 shows the architecture of the hybrid pixel Timepix detector used in the present work. In this case, the connection between the sensor semiconductor material and the readout electronics is realized with flip-chip technology (also called “bump-bonding”).

![Figure 4.1: design of a medipix pixel: the semiconductor active layer (300 µm) and the ASIC electronic hold together through a tin bump-bonding welding.](image)

This approach is applied to all the family of Medipix detectors. Timepix represents an evolution of the previous Medipix2 chip. This last chip is based on a similar principle of FPGA electronics described in Chapter 3. Charges produced by interacting particles in the active material induce a current pulse which then is amplified and discriminated by a threshold level. If the pulse amplitude is higher than threshold, it is registered as one by a digital system. It has also the possibility to set two threshold (high and low) in order to select a definite range of pulse amplitudes. Typically, charge is produced in a semiconductor layer 300 µm thick which is the sensitive volume bump-bonded to the read-out electronics. However, depending on the particular application, this layer material can be substituted with other compounds. These characteristics have shown great potential in different applications requiring single photon counting approach and the chip has been used successfully in several imaging applications: X-ray diffraction, neutron imaging, electron microscopy, Micro Patterned Gas Detectors, Mammography etc.

However, the Medipix2 chip did not provide information on the arrival time of the electron in the sensitive gas volume nor on the quantity of charge deposited. Timepix, instead, allows for measurements in arrival time, “time-over-threshold” (ToT) and/or event counting modes independently in each pixel. Like Medipix2 chip, also Timepix chip is made of 256 x 256 pixels, each one having an area of 55 x 55 µm². The front-end electronic is realized in CMOS technology: each cell contains about 550 transistors and the static power consumption is ~ 13.5 µW. Some of the main nominal characteristic parameters of a Timepix chip are summarized in Table 4.1.

<table>
<thead>
<tr>
<th>Physical dimensions</th>
<th>14.08 x 14.08 mm²</th>
</tr>
</thead>
<tbody>
<tr>
<td>PADs</td>
<td>55 x 55 µm²</td>
</tr>
<tr>
<td>Charge collection</td>
<td>e⁻, h⁺</td>
</tr>
<tr>
<td>Pixel functionality</td>
<td>Counting, ToT, Timepix</td>
</tr>
<tr>
<td>Amplifier gain</td>
<td>~ 18 mV/ke⁻</td>
</tr>
<tr>
<td>Noise</td>
<td>~ 75 e⁻</td>
</tr>
<tr>
<td>Linearity</td>
<td>Up to 50 ke⁻</td>
</tr>
<tr>
<td>Minimum threshold</td>
<td>~ 500 e⁻ (expected)</td>
</tr>
<tr>
<td>Counter depth/overflow</td>
<td>14-bit/yes</td>
</tr>
<tr>
<td>Max Analog Power</td>
<td>6.5 µW/pix; 190 mA/Chip</td>
</tr>
<tr>
<td>Static digital power</td>
<td>200 mA @ 100 MHz</td>
</tr>
<tr>
<td>Read out</td>
<td>Serial/Parallel</td>
</tr>
</tbody>
</table>

*Table 4.1: characteristic parameters of the TimePix Chip [61].*
A block diagram of the Timepix pixel cell is shown in the following figure.

![Block Diagram of Timepix Pixel Cell](image)

**Figure 4.2:** block diagram of a Timepix pixel after the bump-bonded welding: it can be distinguished an analog and a digital part.

Each pixel is divided in two large blocks: the analog side and the digital side. The main elements of the first one are:

- Preamplifier
- Discriminator (with polarity control pin)
- 4-bit threshold adjustment

while for the second block, the principal constituents are:

- Timepix Synchronization Logic (TSL)
- the 14-bit shift register
- the overflow control logic
- the Ref_Clk pixel buffer
- 8-bit Pixel Configuration Register (PCR)

The pixel has two working states, depending on the Shutter signal which can be in an high or a low state. When this signal is high, an external clock is used to shift the data from pixel to pixel, the 8-bit configuration register (PCR) is programmed and the 14-bit shift register is readout. If it is low, the 14-bit shift register works as a linear feedback shift register counter with a dynamic range of 11810 counts. In this state, the pixel counter is incremented by the Ref_Clk depending on the settings of the pixel operation mode bits (P0 and P1):

- Event counting mode (P0 = 0, P1 = 0): each event above threshold increments the counter by 1,
- ToT mode (P0 = 1, P1 = 0): the counter is incremented continuously until the input charge is over threshold,
• Arrival time mode (P0 = 1, P1 = 1): the counter is incremented from the moment the discriminator is activated until the Shutter signal is set high.

Any collected charge from the pixel is integrated and compared to a global threshold by the discriminator. This component generates a pulse whose width is proportional to the length of time the preamplifier output voltage remains over threshold. At this level, there are three DAC parameters which can be used to control the preamplifier signal:

• **Preamp**: it tunes the peaking time which can go from 90 to 180 ns,
• **Ikrum**: this regulates the switch-off time (the return to zero) of the preamplifier signal pulse and it can be varied from 500 to 2500 ns, when the input charge is 10 ke⁻. High values of Ikrum corresponds to small times and vice versa.
• **Vfbk** voltage: it controls the DC output level of the preamplifier. In default DAC settings, the amplifier gain is 18 mV/ke⁻ with a linear voltage dynamic range up to ~ 50 ke⁻, according to the expected nominal values. Some recent publications give a measured value until 20 ke⁻ [...].

After 20 ke⁻, the preamplifier output voltage could begin to saturate. In the present case linearity will be verified because the original front end electronics has been slightly modified. The loss of linearity in the preamplifier voltage represent a limit on the amount of charge that can be correctly processed from the readout electronics. The real linear working range will be verified, because it is an important information which involves the correct operation of the detector.

### 4.2 The GEMpix detector

Typically the previous detector has a 300 µm thick layer of silicon as active material. Particles interacting in this volume release electron-hole pairs. Few eV are needed to produce an electron-hole pair in a semiconductor. It is possible to collect either positive or negative charge. Efficiency can be increased applying a bias voltage to the silicon. In these conditions, the effective active volume (the semiconductor depletion layer) is increased and a larger amount of charge is collected. However, this semiconductor devices suffer when exposed to high energetic radiation like heavy ions or high intensity radiation like radiotherapy X-ray beams. In the last years, new approaches have been proposed. These are based on the substitution of the semiconductor layer with a gas mixture converter which has a density 3 order of magnitude less than a solid. On the order hand, however, it is necessary to amplify charge released in the gas because of its smaller density and the higher ionization potential compared to a semiconductor. In a gas energy one order of magnitude higher respect to semiconductors is needed to produce an electron positive ion couple. Among the various techniques, some researchers suggest the use of micro-pattern structures like MICROMEGAS and GEMs. In the following paragraph, a new GEM detector equipped with the Timepix electronic is presented. In order to have a larger active area respect to Timepix, a quad-medipix layout has been used, a set of 4 chip hold together. The substitution of semiconductor with a triple GEM has also the main advantage to increase the dynamic range of about 3-4 order of
magnitudes. It is possible to act on the GEM gain so that when an high fluence of particles is expected, the gain is set low and the detector continues to work in its linear region. On the contrary if the fluence is low, GEM gain can be set high in order to observe also the single particle.

4.2.1 The new GEM layout

The active area of the new detector is defined by a quad of 2.8 x 2.8 cm$^2$, four times the Timepix area (1.4 x 1.4 cm$^2$). The idea of the new detector is to use quad electronics on a triple-GEM gas detector, in a similar way to the 10 x 10 cm$^2$ GEM detector presented in the previous chapter. Now the area is smaller so the geometry of this new detector must be properly adjusted. A set of new GEM foils 3 x 3 cm$^2$ in area were required to the CERN production center (fig. 4.3) [..].

![Figure 4.3](image)

Figure 4.3: a 3 x 3 cm$^2$ GEM foil with high voltage strip electrodes.

As can be seen from the figure, each GEM foil has two electrodes to which it will be possible to apply a voltage difference. GEM foils are glued on new epoxy glass frames 10 x 10 cm$^2$ in area.

![Figure 4.4](image)

Figure 4.4: the of epoxy glass frames used in the assembling of the GEMpix detector.

Five frames have been used, each with a different thickness: 3, 1, 2, 1 and 2 mm. The first thickness defines the drift region and active volume of the detector like the first prototype. The subsequent 2 thicknesses defines the two gaps between the three GEM foils 1 and 2 mm, the usually transition regions. Assembling procedure for GEMpix followed the same steps described in par. 3.1.3, except that now PCB with the quad-medipix chip is screwed on the last epoxy glass frame. This is simply a convenient choice which allow to substitute the FEE electronic if would be necessary. For the induction region, instead, the last two thicknesses are made in order to have an induction region between the last GEM and the Timepix electronics of 2 mm (fig.4.5)
As can be noted from the figure, GEM electrodes exit from the glued frames and are soldered on a PCB board. On the other side of the chamber, the PCB board has a 3 x 3 cm$^2$ mylar window which is alluminated on the internal side and acts as the chamber cathode like for the first prototype (fig.4.6).

The chamber has only two hole to which a couple of plastic gas tubes is attached (blue tubes in fig.4.6). In the previous figure, it can be also notice the high voltage connector (the gray box) and the quad control unit, the so called “Blue Box”. Like the first prototype, the high voltages are provided by the HVGEM module. Blue Box, instead, is simply USB connected to a PC, like for the original Timepix. But now the quad needs of a power supply of 3 V to work correctly.

### 4.2.2 Medipix and ToT modes

In this paragraph, a detailed description of the Medipix and ToT mode of operation will be presented. As set of test pulse have been performed in order to make a first characterization of the quad electronics performances. The following figure reproduces the layout of a Timepix pixel electronics. The area marked by an octagonal corresponds to the area (20 µm long) where the tungsten bump-bonded sphere is welded. In the present case this area is free and represents the place where charge is collected. It is assumed that all the charge reaching the pixel area (55 x 55 µm$^2$) is collected by the octagonal.
As already mentioned, a charge collected from a pixel is integrated by the preamplifier which generates a voltage pulse whose amplitude is proportional to the integrated charge reaching the pixel.

A set of measured preamplifier output pulses is shown in fig. (4.8) when an input charge of 18.7 ke\textsuperscript{-} is sent on the pixel [61]. All the pulses have the same amplitude, but differ for the *Ikrum* value which manages the switch-off time of pulses. In medipix mode this parameter would have no effect because the pulse is registered as a single count when its amplitude is over a set threshold. However there is a dependence. Its trend has been studied in test pulse modality. Each channel has a test capacitor which can be enabled to send definite pulses to the preamplifier. In this case the Timepix electronics works in “Test Pulse Mode”. Each pulse is defined by its height and period. Pulse height is given by the voltage applied to charge the capacitor and can range from 0 to 1.4 V. Pulse period defines the pulse time width and can range from 0 to 236 µs. The following plots (fig. 4.9) show the mean counts as a function of the *Ikrum* value when 200 test pulses of three different amplitudes and with a time width of 3 µs are sent on the pixel matrix. In this case noise threshold is set up to 40 units over the maximum electronic noise levels for all the four chips. At this value no noise counts are registered.
Previous plots show clearly that for Ikrum values major than 10, the mean number of counts per pulse is 1. High values of Ikrum corresponds to short pulses and then as they become shorter, the registered counts are equal to those sent on the detector. Experimentally some measurements on the single X-ray pulse have been performed (fig. 4.10) with an Ikrum value equal to 5.

The result shows the GEMpix in Medipix mode is not able to counts single particles in a pulse. Rather it gives a count for each pulse and rarely two counts. Setting higher value of the Ikrum parameter does not give a different result. The important think, here, is that if used in Medipix mode GEMpix is not able to provide a quantity proportional to the number of particles coming from the irradiating beam but to the number of pulses coming from the accelerator and this information cannot be correlated to dose. However it is possible to work in ToT mode and this represents the great potential of Timepix electronics. The following schemes illustrates the Medipix and ToT modes of operation.
In both cases there is a shutter which opens in the time width set on the control unit panel. It is started and stopped by an internal timer. During shutter opening, counts from an internal clock are registered every time the discriminator output generate a squared pulse which has a time width equal to the time a preamplifier analog pulse is over threshold. But while in Medipix mode there is a single clock pulse for each discriminator pulse, in ToT mode there is a set of consecutive clock pulses until the discriminator pulses are switched off. So ToT mode allows to correlate the pulse amplitude to the number of registered clock pulses and then to the charge released in the gas. This time Ikrum parameter can significantly affect the registered counts for each pulse and its value must be properly set.

4.2.4 ToT mode performances

According to the nominal values given in table 4.1, preamplifier output is linear until an input charge of 50 ke⁻. Above this limit, the correct processing of the signal is not guaranteed. Also for the ToT mode, it was useful to perform a pulse test. It was useful to verify the linearity between pulse amplitude and ToT counts (fig. 4.12).

Test pulse counts are the sum on the number of pulses sent on the pixels. In this case 100 test pulses are sent on the pixel matrix. Each pulse has a period of 3 µs while its height is varied from 0.01 to 0.30 V with step of 0.01 V, while the Ikrum value is set to 5. It is expected direct proportionality
between pulse amplitudes and ToT counts. However this proportionality is observed in the range between 0.001 and 0.1 V.

![Figure 4.13: charge linear working range of the FEE GEMpix.](image)

For higher values, counts begin to decrease slowly and then reach a plateau. The amount of charge $Q_{inj}$ for each pulse of a given height $V_{TP}$ is calculated from the following relation:

$$Q_{inj} [e^-] = \frac{\alpha_{BUF} \cdot V_{TP} \cdot C_{Test}}{e}$$

where $\alpha_{BUF}$ is the gain in the injection test pulse buffers (~ 0.825), $C_{Test}$ is the test capacitance which is equal to 8 fF and $e$ is the electron charge ($1.6 \times 10^{-19}$ C). According to this relation, it can be said that ToT counts provide a reliable measure when the injected charge varies from 40 $e^-$ to 5 ke$^-$, according to the values estimated by test pulse analysis. The lower limit, however, is compatible with electronic noise (~75 $e^-$), so it is possible to declare the correct working range is between 0 and 5 ke$^-$. It must be noted that the corresponding ToT time would be next to zero for very low values, but this is not observed. It has been observed that ToT test pulses depends on the on the $Ikrum$ parameter and on the applied threshold. Changing these parameters lower value can be obtained but the linear trend is always limited to this range.

### 4.3 Fluka simulations for GEMpix

From the previous paragraph, it is clear that injected charge on the quad pixels must not exceed an upper limit of 5 ke$^-$. Simulation presented in the present paragraph are aimed mainly to evaluate the number of secondary electrons reaching the active volume and the relative energy released in the gas mixture, in order to estimate the amount of produced ionization charge. It will be demonstrated that charge released in the gas drift region can exceed the linear range and that also in this case it is an optimal choice to work with drift field off. This also minimize the effect of secondary electron scattering along the drift and contributes to refine the shaping of the fields.
4.3.1 Simulation set-up

Like for the previous GEM detector, geometry of the detector was realized using a series of planes in order to define the mylar window, the three GEM foils, the PCB with read-out electronics and the gas mixture layers between all these elements. The incident X-ray beam has a squared area of $8 \times 8 \text{ mm}^2$ and is located 100 cm from the detector window. It has an angular divergence necessary to have an $8 \times 8 \text{ mm}^2$ field on the detector window and an X-ray spectrum which corresponds to the accelerating potential of 6 MV.

![Figure 4.15: side view of the simulated photon beam with an $8 \times 8 \text{ mm}^2$ area on the detector window (left) and the simulated electron distribution along the detector elements (right).](image)

The total estimated number of electrons on upper side of the first GEM foil is $(1.93 \pm 0.06) \cdot 10^{-3}$ electrons/primary. Simulator takes into account all the entering and backscattered electrons. The energy spectra distributions for photons and electrons on the first GEM foil is shown in the following plots.

![Figure 4.16: photon spectra before and after the mylar foil and the drift gas layer (left) and the electron energy spectrum at the entrance of the first GEM foil.](image)

Right plots shows the spectrum distribution of the photons entering the mylar window. Practically the same spectrum is calculated for the photons entering on the first GEM foil. Their equality highlights that mylar foil has no influence on beam fluence and X-ray spectrum. Calculated photon transparency in mylar and drift gas layer is $0.99455 \pm 0.00002$ ph/pr. The number photons converting on mylar is very small. With no mylar, indeed, the estimated number of electrons reaching the upper first GEM foil is $(1.20 \pm 0.09) \cdot 10^{-3}$ e/pr which is more than 80% of the total number of electrons produced in mylar ($(1.50 \pm 0.09) \cdot 10^{-3}$ e/pr). Most of the electrons comes from X-ray interactions in air. Electron energy distribution is peaked at low energies and decreases until...
about 5 MeV. So the main contribution to ionization comes from low energy electrons which, according to Compton scattering, corresponds to those electrons released at greater angles. In addition, at low energies, also photoelectric gives its contribution.

### 4.3.2 Energy deposition

Both acquisition modes (counting and ToT) are related mainly to the number of secondary electrons reaching the first GEM surface and the energy they release along their path. In the following plots, there are the Fluka simulation results obtained for the electron fluence and the deposited energy in drift gas layer over the first GEM foil. At moment it is considered the possibility to work with drift on in the ToT mode.

![Figure 4.17: simulated side view of the energy deposited along the detector (left) and energy deposition in the 100 µm gas layer on the first GEM foil for an 8 x 8 mm² beam at 6 MV.](image)

The released energy (GeV/cm³/pr) allows an estimation of the ionization charge released in the drift region. However electrons collected in the drift region depends on the applied electric field. For standard drift field of 3.0 kV/cm typically used in head-on GEM detectors, all the charge released in 3 mm is collected towards the first GEM foil. As described in chapter 3, digital read-out leads to a counts saturation. A similar discussion is valid for GEMpix when it is set on Medipix mode. ToT mode would work correctly, however, it is not helpful to collect all the charge released in the drift region because it spreads in the volume and the resulting detected area of the energy distribution can spread over the limits defined by the real beam limits. Some test measure with drift on and ToT mode have been executed on 8 x 8 mm² beams. One screenshot of these test is shown in fig. 4.18, from which it is clear that the detected beam is widespread on an area larger than that where beam fluence is concentrated.
Moreover, if all the charge is collected, the preamplifier can saturate and the registered ToT counts are no more linearly correlated to the charge. So, also for GEMpix, it is advantageous to work with zero drift field collecting only the charge released in the last 100 µm before the first GEM. To clarify these mechanisms a simulation is performed in which the energy released inside a gas column 3 mm high and 1 x 1 mm$^2$ in base area is calculated with a pitch of 50 µm, when the detector is irradiated with 6 MV beam with an area $A_b = 8 \times 8 \text{ mm}^2$.

\textbf{Figure 4.19}: the gas column 3 cm high with a base of 1 mm$^2$ along which the deposited energy is calculated with a step of 100 µm.

Plot on the right outlines the trend of energy released along the gas mixture column. As can be noted uncertainties are large so that the histogram can match with a constant line along the column. The mean value is between 5.0 and $9.0 \cdot 10^{-10}$ GeV/(cm$^3$pr). Considering $E_{ArCO_2} = 7.0 \cdot 10^{-10}$ GeV/(cm$^3$pr) as the mean energy released per unit volume and taking into account that the number of photons per pulse $\gamma_{\text{pulse}}$ is given by

$$\gamma_{\text{pulse}} = \dot{\phi}_{\text{pulse}} \cdot A_b \cdot \Delta t_{\text{pulse}} = 4.1 \cdot 10^6 \text{ photons}$$

where $\dot{\phi}_{\text{pulse}} (= 8.5 \times 10^{12}$ ph/cm$^2$sec) is the photon fluence per pulse and $\Delta t_{\text{pulse}} (= 3$ µsec) is the pulse length, the calculated mean energy released in the gas column per mm$^2$ is calculated as

$$\gamma_{\text{pulse}} \cdot E_{ArCO_2} \cdot 3 \text{ mm} \cong 8600 \text{ eV/mm}^2$$

The ionization energy required to produce a ion-electron couple in Ar/CO$_2$ mixture is about 32 eV, whereby an estimate of the total number of electrons produced in the column can be calculated as the ratio between the total released energy and the ionization energy. The resulting electron charge
is 270 e⁻/mm² and then a very low charge on a 55x55 µm² pixel respect to the expected preamplifier input (0.04 – 5 ke⁻). However, ionization charge must be amplified by the triple-GEM. This is the reason why GEMs are used: generally ionization charge produced in gas is very low respect to a semiconductor layer. On the other hand, the configuration with a triple-GEM allows a broad dynamic range. The sum of voltages applied to the GEM foils $V_{GEM}$ is correlated to the gain $G$ through an exponential relation like this:

$$ G = A e^{BV_{GEM}} $$

where A and B are parameters which depends on the gas mixture. In the present case, A and B for Ar/CO₂ (70/30) have been measured and are equal to $1.2 \cdot 10^{-4}$ and $1.68 \cdot 10^{-2}$, respectively. The following graph reports gains curves for different gas mixtures on a logarithmic scale in the range of applied voltages between 900 and 1400 V.

![Figure 4.20: gain vs applied potential curves for four gas mixtures: ArCO₂ (70/30), ArCO₂CF₄ (60/20/20), ArCO₂CF₄ (45/15/40) and ArCF₄C₂H₁₀ (65/28/7).](image)

The black curves are the exponential fit from which A and B parameters are calculated. So if the input charge is 270 e⁻/mm², it is possible to apply the appropriate voltage to obtain charge values in the range required by the preamplifier. A gain of about 190 is obtained with 850 V, so that the input charge per mm² is $5.1 \times 10^4$ e⁻ and then the medium input charge per pixel (55 x 55 µm²) is about 150 e⁻ which falls in the linear range (until 5 ke⁻). At a voltage of 950 V the gain is about 1000 and the input charge per pixel becomes 840 e⁻ and again it falls in the linear range. It is important to underline that the real number of ionization electrons is lower than the calculated one using the energy released in the gas. Indeed not all the energy is used to ionize the gas. However, this percentage is considered of small amount in gas. When the detector work with zero electric drift field, the volume of gas on which charge is collected that is 0.1 x 1.0 x 1.0 mm³. A beam profile has also been calculated considering a row of these small volumes.
Observing this profile distribution, in can be said that energy is concentrated on the central area where is the beam, but there are also some value outside. Deposited energy for each volume undergoes large fluctuations meaning that released charge can vary a lot between two adjacent volumes. Now the real active volume is 1/30 of the original column. At each photon pulse a distribution of random blobs would be observed rather than a uniform distribution which decreases out of the beam area. Some experimental results are reported in the following fig. [4.22] in which three successive pulses have been acquired for a 8 x 8 mm$^2$ beam at a PRF of 200 Hz with a time width of 5 ms. Each image has been obtained with a re-binning of 18 x 18 original pixels so that measured distribution along the central Y coordinate has the same pitch of the simulated ones.
For each pulse there is a different distribution which is concentrated in the center and drops on the sides, but profiles have some large statistical oscillations, as expected by simulations. They reduce when a series of pulses are successively integrated. These measures demonstrate that the GEMpix detector is sensitive also to the single beam pulse and, for each pulse, few particles are registered. They appear as blobs because, in this case, charge released in the gas mixture is measured.

4.4 Preliminary measures

Same preliminary measure have been performed in order to evaluate the optimal working parameter of the new detector. High voltage parameters of GEMpix can be extrapolated from the information obtained by the GEM detector presented in Chapter 3. Indication coming from the first prototype is that drift field must be zero and this parameter will be retained also in this case. The triple-GEM scheme is the same even if it must be correlated to a new FEE. The present paragraph explain some of the working acquisition parameters used for the HVGEM module and for the acquisition system based on the Blue Box control unit. Some results obtained from an HV scan which gives important indications on the detector response to charged produced by the beam are first presented. For this purpose, it has been decided to irradiate with $8.0 \times 8.0 \text{ mm}^2$ 6 MV beam. This energy value is the lower one and it is expected an amount of interaction higher than the others two energy spectra (10 and 18 MV). PRF (MU/min) has been set to 200 Hz, a typical value used in radiotherapy.

4.4.1 High voltages scans

Before a real dosimetric measure, it was necessary to evaluate the detector response to the amount of charge produced when the triple-GEM high voltage gain is varied. As explained in par. 4.2.4, the preamplifier of the FEE electronic puts a limit on the detector linear response. When the triple GEM voltage $V_{\text{gem}} = V_{g1} + V_{g2} + V_{g3}$ is increased, the amount of multiplied charge per pulse increases and this can reach the region of non-linearity of the preamplifier. According to the
simulations presented in the previous paragraph and taking into account that the drift field is off, the amount of charge could not overcome the upper limit of the charge range estimated (0 - 5 ke\textsuperscript{-}), even for GEM voltages higher than 1000 V. However an high voltage scan is necessary to evaluate the detector response. HV scan has been performed applying the values of table 4.1 to the GEMs and integrating on 100 frames, each one having a time width of 5 ms.

<table>
<thead>
<tr>
<th>Vgem</th>
<th>Vg1</th>
<th>Vg2</th>
<th>Vg3</th>
</tr>
</thead>
<tbody>
<tr>
<td>865</td>
<td>295</td>
<td>285</td>
<td>285</td>
</tr>
<tr>
<td>886</td>
<td>306</td>
<td>295</td>
<td>285</td>
</tr>
<tr>
<td>906</td>
<td>306</td>
<td>305</td>
<td>295</td>
</tr>
<tr>
<td>926</td>
<td>316</td>
<td>305</td>
<td>305</td>
</tr>
<tr>
<td>946</td>
<td>326</td>
<td>315</td>
<td>305</td>
</tr>
<tr>
<td>966</td>
<td>326</td>
<td>325</td>
<td>315</td>
</tr>
</tbody>
</table>

*Table 4.1: the total voltage applied to the triple-GEM with the single voltages applied to the single GEMs: Vg1, Vg2 and Vg3. Drift field is set to zero.*

To study the response of the detector, a central area of 64 x 64 pixels has been selected for each counts distribution and the mean count value has been evaluated. The result is presented in fig. 4.23.

![central count vs HV](image)

As can be observed counts begin to rise and reach a linearly between 890 and 930 V. From about 930 – 940 volts the curve begins to bend and then rise again. It has been observed that counts distributions having a voltage higher than 930 V begins to register an higher contribution on the
tails. This means that detector, at high voltage gains, become more sensitive on to the charge released by the more angulated scattered electrons distributed on all the sensitive area of the detector. This explains the second rise of the curve. Away from the central beam, their contribution dose not correspond to a significant dose value which can be of clinical interest in radiotherapy. For these reasons, it has been decided that the best values to work are the voltages minor than 930-940 V. However these cannot be too small, otherwise the signal is to weak. The value used for all measures is 930 V, so that the detector has a low sensitivity to penumbra regions around the beam.

4.2.3 ToT acquisition parameters

All dosimetric measures have been executed in ToT mode. Fig. 4.24 shows the control panel of the Pixelman software which controls FEE of GEMpix. This software is provided by the Institute of experimental and applied physics of the Czech Technical University in Prague [ref].

![Figure 4.24: Caption of the Pixelman control panel. It is possible to distinguish a Device Control panel in which it is possible to set the acquisition type (integral), the number of acquired frames (200) and the acquisition time (0.005 sec). In the bottom, there is the DAC control panel in which several acquisition parameters can be set: Ikrum, THL, etc.](image)

All parameters have been left in their default values, except Ikrum, THL and internal clock frequency for ToT acquisitions. As explained previously, Ikrum is related to the current pulse duration. In this case it has been set at 5, which corresponds to current pulses with large switch off times. With this configuration, larger current pulses can be registered so that much more charge could be integrated. THL is the parameter which represents the thresholds and can range from 0 to 1024. There are 4 values, one for each chip. The working thresholds have been set to the minimum values in which no electronic noise was observed. As an example, one of the applied set is: 372, 364, 290 and 419. The internal clock frequency has been set to 10 MHz which corresponds to minimum observed time of 100 ns. Frequencies until 50 Mhz are possible, but this has only the effect of refine the time resolution. It these case long pulse are expected and 10 MHz is an optimal choice. Another important parameter is the time width on which counts are integrated. It has a
minimum value of 1 µs. In this case a value of 5 ms has been applied. It corresponds to a frequency of 200 Hz and this means that it can detect the single pulse coming from accelerator for all the working PRFs, except for the last value (400 Hz) for which two consecutive pulse are registered. A measure of the total charge released is performed setting acquisition on the integral mode. In this configuration each frame acquired at 5 ms, is summed to the next and so on until irradiation is over.

### 4.5 Analysis on the measured field

To quantify the performance of a 2D dosimeter, a set of dosimetry tests has been executed. The most important ones for 2D dosimeters include the following:

- **Dose and dose rate dependence.** The dosimeter must be able to provide a measure directly proportional to dose and dose rate. In particular it is preferable a linear response on a large dose and dose rate range. In the present case, only dose rate has been tested using the seven values available from the Linac (6, 12, 25, 50, 100, 200 and 400 MU/min). GEM detectors presented in this work, indeed, acquire frame by frame and the total dose released can be calculated integrating all the frames registered during the irradiation time. This represents a remarkable advantage because measurable dose range can be very wide, without losing linearity.

- **Gamma index test on dose distributions.** This test has been introduced in Chapter 1. It is applied for comparisons between a dose distribution to be queried and a reference dose distribution. Generally reference doses are those measured by a calibrated dosimeter, while the compared one can be that calculated by a TPS or measured by another dosimeter. The same test will be also applied for comparison between dose distribution acquired by GEMpix and gafchromic films. In the present case, the reference is represented by gafchromic.

In this paragraph results related to the last two parameters will be presented. For gamma index test results relative to the 6 MV spectrum will be treated in more detail. Then the other two energy spectra will be considered and its results interpreted also with the contribution of new Fluka simulations.

#### 4.5.1 Dose and dose rate dependence

GEMpix FEE is able to acquire a sequence of frames both in counting and in ToT mode. Time width can be sufficiently reduced considering X-ray beam coming from the Synergy Linac. Even if the minimum time width is 1µs and the beam pulse last 3 µs, it is not possible to obtain an at least rough time evolution of a pulse. Limitation indeed comes from the gas drift time of ionization charges along the detector, in particular in the drift and induction region. However it is not important to obtain this kind of information. However, time width so small allows to make a charge distribution measure of the single pulse in all the range of frequencies of the Linac. At its maximum
value (400 PRF) it is enough to work with a time width of 2 ms to observe the single pulse. This means that a measure of the amount of dose released by a beam can be evaluated by simply integrating single frames acquired in a time width small enough to observe the single pulses. Then there is no upper limit over which the dose cannot be measured. Chamber matrix detectors are not sensitive to the single beam pulse, integrate the total ionization charge collected during beam irradiation and have a saturation limit over which the output signal does not respond to a further increase of collected charge. The same happens for gafchromic films. As discussed in Chapter 2, it has a linear working range from 1cGy to 10 Gy (for EBT2). On the other hand, the lower limit for GEMpix is very small because it can measure the single pulse. No films are able to detect a single pulse. Usually film are used with a tissue-equivalent material to increase the amount of charge reaching the detector. In the present case, gafchromic have been located on the detector window and irradiated in air with 1000 MU. Results obtained from GEMpix are resumed in the plot of fig. 4.25.

Figure 4.25: 3D counts distributions obtained with GEMpix for a 8 x 8 mm² beam at the three different energies of the accelerator: 6, 10 and 18 MV.

Figure 4.26: GEMpix pulse scan repetition rate at the three acceleration energies.

In the plot there are three series of data, each one corresponding to the three energies. On the abscissa there are six PRF values: 12, 24, 50, 100, 200 and 400. Counts have been estimated as the mean value on an area of 64 x 64 pixels (3.52 x 3.52 mm²) at the center of an 8.0 x 8.0 mm² beam counts distribution. For all three energies, linear fit shows an optimal result, as expected. When the energy increases, the number of counts decreases because X-ray interaction reduces and the detector is more transparent to incident radiation. This is further highlighted considering the central beam profiles at the three energies along the X or Y coordinates.
This reduction of signal is also observed for the GAFs. Higher energies corresponds to reduced level of blackening. High energy X-rays are more penetrating and release energy to deeper layers. Profiles along Y direction shows a trend similar to those measured by GEMpix (fig. 4.26).

Ratios between the values at the different energies for GAF and GEMpix are equal within statistical fluctuations. It must be observed that GEMpix profiles tends to broaden at high energies, a phenomenon which will appear more clear when gamma index test will be performed. Probably this phenomenon is caused by electron scattering not only on mylar window but also in air and gas mixture drift layer.
4.5.2 Measures and Gamma index tests at 6 MV

The present paragraph will present the measurements performed on 6 MV beams with GEMpix and gafchromics. In both cases, SSD is the usually standard of 100 cm.

Gafchromic films are located on the detector window to be irradiated (fig. 4.30). However measures on GEMpix and gafchromic are acquired separately. GEMpix has an active area of $3 \times 3 \text{ cm}^2$. The smaller standard fields which can be set from the Linac is $8 \times 8 \text{ mm}^2$, the next is $24 \times 24 \text{ mm}^2$. In the present case it is preferable to use the $8 \times 8 \text{ mm}^2$ beam, because the second one can be affected by electric field side effects due to the presence of conductive materials around the quad medipix chips anode. These becomes clear when the GEMpix is irradiated with a beam wider than its active area. Fig. 4.28 shows the counts distribution obtained with a $56 \times 56 \text{ mm}^2$ beam.

Figure 4.29: UP: the Elekta Synergy Linac at the Tor Vergata Radiotherapy Center with GEMpix located on the bed at a distance of 1 m from the source; DOWN: EBT gafchromics placed on the detector window after irradiation.

Figure 4.30: GEMpix counts distribution with a $56 \times 56$ and an $8 \times 8 \text{ mm}^2$ beam at 6 MV: in the last case there can be observed some non-uniformities due to the field lines along mylar window sides.
It is clear that the effective active area is lower than the 4 chips area (fig. 4.30 left). Red points delimit the effective area and indicates regions in which higher counts have been registered. This effect is probably due to detector assembling: gluing and welding of the mylar cathode caused an major concentration of field lines on the sides. This represents an important information for the future development of these detectors. Image on the right of fig. 4.30 shows a measure performed on a 8 x 8 mm$^2$ beam with a PRF of 200 Hz. Acquisition is made by integrating 100 frames having a time width of 5 ms. It is clear that each chip has a different response. This is particular evident for the chip in the upper right position. It is necessary to correct for this systematic effect. To this aim, the following procedure is applied:

- the image has been re-binned to larger pixel obtained by holding together 32 x 32 original pixels,
- the count assigned to the great pixel is calculated as the medium value of the original pixel values,
- for each chip, the value of the great pixel at the center (where all four chips meet) is used to divide the original values of the chip itself,
- then all the original value are multiplied by an intermediate value between those of the four great pixels at the center.

Fig. 4.30 shows a sequence of images corresponding to the various stages going from the original distribution until that equalized for the lack of chips response uniformity.

![Figure 4.31](image)

**Figure 4.31**: illustration of the equalization method applied on a 8 x 8 mm$^2$ beam at 6 MV: the central figure show the distribution obtained after re-binning with 32 x 32 pixels.

Gamma test is performed between GEMpix distributions and GAFchromic films. For measures performed on square fields, EBT3 gafchromics have been used, while for IMRT segments the EBT2 model. There is no particular reason, the choice depended on the availability of one type or the other at the moment of measure. In the following discussion, both EBT models are generally referred to EBT or gafchromic. The spatial resolution of both EBTs is 339 µm and this is sufficient in radiotherapy. More precisely, this value depends on the $dpi$ parameter set on the scanner. It defines the dots per inch ($dpi$) of the corresponding digitized image. The calibrated $dpi$ value set on the EPSON scanner used at Tor Vergata radiotherapy center is 75, which corresponds to a spatial resolution of 339 µm. Pixel size of GEMpix is 55 µm which is not comparable to the gafchromic resolution. In addition a similar resolution is not required in radiotherapy. For this reason, GEMpix distribution is re-binned to great pixels of 4 x 4 original pixels in order to have a spatial resolution of 220 µm. The value assigned to the great pixel is the medium value of the 16 original pixels. Re-
binning is advantage also because it make more smooth counts distributions and then less affected by statistical fluctuations. Fig. 4.31 shows the equalized re-binned field obtained from GEMpix and the relative measure obtained from an EBT.

**Figure 4.31:** original dose distributions obtained on a 8 x 8 mm$^2$ beam at 6V with GEMpix (zero electric drift field) and EBT gafchromic.

Both measures have been normalized to 100 following the same procedure described in par. 3.4.2 with a tails cut of 10%. The resulting normalized distributions and the resulting gamma index map are shown in fig. 4.33. Gamma index map is reconstructed on the points of the gafchromic film which represent the reference dosimeter and again has been calculated using the ROOT analysis program.

**Figure 4.32**: original dose distributions obtained on a 8 x 8 mm$^2$ beam at 6V with GEMpix (zero electric drift field) and EBT gafchromic.

**Figure 4.33**: UP: normalized GEMpix (zero electric drift field) and EBT GAF dose distributions measured on a 8 x 8 mm$^2$ field at 6 MV with the gamma index map; DOWN: X and Y central profiles of gafchromic (red line) and GEMpix.

Results shows that comparison between the two distributions passes the gamma acceptance test ($\gamma < 1$) on a number of measured points major than 90 %, and dose matching between the two distribution is excellent according to the commonly accepted criteria of DTA and DD equal to 3 mm and 3% (par. 1.4.4). In order to evaluate the ability of the detector to measure particular dose
distribution which are typically applied in IMRT treatments, an irregular IMRT segment has been prepared from TPS as if an IMRT field would be irradiated at 6 MV. Fig. 4.33, on the left, shows the set configuration of tungsten multi-leaf lamellae and the corresponding measure obtained by GEMpix and gafchromic.

![Image](image1.png)

**Figure 4.33:** The Synergy multileaf set-up for an IMRT segment field (left) with the corresponding dose distributions measured with GEMpix (zero electric drift field) and EBT gafchromic.

Gamma index calculation provides the results shown in fig. 4.35. Even for IMRT segments, the described procedure for equalization and normalization has been applied. Now, however, the values for normalization have been obtained not at the center but on the borders between chips on which the counts would be uniform.

![Image](image2.png)

**Figure 4.35:** normalized GEMpix (zero electric drift field) and EBT GAF dose distributions measured on a IMRT segment field at 6 MV with the gamma index map.

Acceptance criterion for gamma index is satisfied only in 61.09 % of the measuring points. It is clear that GEMpix counts distribution has a shape which cannot match the EBT distribution. In addition distribution is not smooth, but some blob structures can be observed even after re-binning. It is possible that measure is affected by the holes structure of GEMs. As explained in par. 3.1.1, electric fields lines are concentrated inside the GEM holes and ionization is collected towards the holes. Results concerning the other two energies are presented in the next section.
4.5.3 Measures and Gamma index tests at 10 and 18 MV

Also in this case, detector is irradiated with an 8 x 8 mm² beam with a PRF equal to 200 Hz. Again the normalization procedure explained in the previous paragraph is applied. Results for gamma index test are reported in the following fig. 4.36 and fig. 4.37.

**Figure 4.36**: UP: normalized GEMpix (zero electric drift field) and EBT GAF dose distributions measured on a 8 x 8 mm² field at 10 MV with the gamma index map; DOWN: X and Y central profiles of gafchromic (red line) and GEMpix.

\[ \gamma < 1, 69.16 \% \]

**Figure 4.37**: UP: normalized GEMpix (zero electric drift field) and EBT GAF dose distributions measured on a 8 x 8 mm² field at 18 MV with the gamma index map; DOWN: X and Y central profiles of gafchromic (red line) and GEMpix.

\[ \gamma < 1, 43.04 \% \]
Gamma index acceptance test is satisfied for a lower set of points respect to the 6 MV spectrum. In particular, they not pass the required percentage of 90%. At higher energies the contribution coming from secondary scattered electron becomes significant and around the principal beam large halos are observed, as underlined particularly by the 1D central plot profiles. To confirm this hypothesis, some Fluka simulations have been performed to evaluate 2D energy distribution, like those presented in paragraph 4.3.2. Results are relative to the spectra of 6 and 18 MV.

As can be observed from the plots of fig. 4.38, at higher energy spectra, the simulated energy deposition in the gas layer is spread on a wider area with a corresponding increase of the halos around the main beam area, in agreement with the experimental observations.

### 4.5.4 Measures and Gamma index tests with GEM1 off

In order to obtain a correct measure for higher energies it has been decided to work with GEM1 at zero potential. In this configuration, the first GEM foil can be used not only to block scattered electrons coming from the drift region but also as a converter. Now collected charge comes from the first transition region, while only the other two GEMs are used as multiplication stages. Results are resumed in in fig. 4.39, 4.40 and 4.41 for all the three energy spectra.
Figure 4.39: UP: normalized GEMpix (zero electric drift field and GEM1 off) EBT dose distributions measured on a 8 x 8 mm$^2$ field at 6 MV with the gamma index map; DOWN: X and Y central profiles of EBT (red line) and GEMpix.

Figure 4.40: UP: normalized GEMpix (zero electric drift field and GEM1 off) EBT dose distributions measured on a 8 x 8 mm$^2$ field at 10 MV with the gamma index map; DOWN: X and Y central profiles of EBT (red line) and GEMpix.
Figure 4.41: UP: normalized GEMpix (zero electric drift field and GEM1 off) EBT dose distributions measured on a 8 x 8 mm$^2$ field at 18 MV with the gamma index map; DOWN: X and Y central profiles of EBT (red line) and GEMpix.

Previous results show a significant improvement in the measure of dose distributions, especially at high energies. In this new configuration GEMpix counts are reduced and the tails around the main beam are greatly reduced respect to GEMpix measures with GEM1 on, even for higher energy beams. The following table summarizes the gamma index values obtained for all the square fields measured before and after the GEM1 off condition.

<table>
<thead>
<tr>
<th>ENERGY SPECTRUM</th>
<th>γ GEM1 on V$_{gem}$ = 930 V</th>
<th>γ GEM1 off V$_{gem}$ = 665 V</th>
</tr>
</thead>
<tbody>
<tr>
<td>6 MV</td>
<td>97.57 %</td>
<td>93.70 %</td>
</tr>
<tr>
<td>10 MV</td>
<td>69.16 %</td>
<td>97.28 %</td>
</tr>
<tr>
<td>18 MV</td>
<td>43.04 %</td>
<td>90.14 %</td>
</tr>
</tbody>
</table>

Table 2: gamma index percentages on the number of points with γ < 1 before and after for GEM1 off on a 8 x 8 mm$^2$ at the three different energies.
4.5.5 GEMpix Fluka simulations with GEM1 off

In order to understand the physical phenomena involved in the new configuration with GEM1 off, Fluka simulations have been performed. Geometry is the same as the previous one, but the interesting region is represented by the first transition region between GEM1 and GEM2. Fig. 4.42 shows the results obtained for the intermediate energy of 10 MV. The aim of simulations is to understand the physical mechanisms involved in this new configuration. The other two energies present a similar trend, so only the central one has been considered. First plot is an estimate of the energy released along the detector in a column having a base of $2 \times 2 \text{ mm}^2$ with a step of 100 µm.

![Figure 4.42: deposited energy in a volume 90 mm long with a base of $2 \times 2 \text{ mm}^2$ along the detector. It is possible to distinguish step changing of the energy at each constitutive element of the detector.](image)

It is clear that energy in the first transition region is higher than that estimated in the drift region. A higher amount of electron-ion couples will be produced. However these will undergo only two multiplication stages with the result that less counts are registered.

![Figure 4.44: simulated energy distribution on the gas layer (100 µm) over the first GEM (left) and it the gas first transition region (1 mm) for a 8 x 8 mm$^2$ beam at 10 MV.](image)

2D energy distributions shows a contrast enhancement on passing from the gas layer on GEM1 to the first transition region and this is further highlighted in the plots of the X profiles at zero Y coordinate (fig. 4.45). In addition, when only two GEMs are considered as multiplication stages, the contribution due to electron scattering around the central beam is significantly reduced.
4.5.6 IMRT measures and Gamma index tests with GEM1 off

In order to verify the potentiality of the configuration with GEM1 off, two IMRT fields have been prepared at 6 MV with a typical dose prescription usually applied in radiotherapy. Detector has been sequentially irradiated with the two fields. For each one gamma index test has been performed. Again chip counts have been equalized following a procedure similar to that explained in paragraph 4.5.2 and again GEMpix pixels have been re-binned in 4x4 pixels. The results obtained from measurements after equalization and re-binning are shown in the following fig. 4.46.

Figure 4.45: simulated central X profiles obtained from the 2D distribution of fig. 4.44 taking the energy deposited in a 100 µm gas strip for the gas layer on the first GEM and the first transition region.

Figure 4.46: The Synergy multileaf set-up for two IMRT segment fields at 6 MV (left) with the corresponding dose distributions measured with GEMpix (zero electric drift field and GEM1 off) and EBT gafchromic.
Acceptance gamma test have been executed on both. In particular the first segment is always that presented in 4.34 and allows to verify if this new configuration produces a best result, like for the square fields of paragraph 4.5.5. Gamma index test results are shown in fig. 4.46.

![Gamma Index Results](image1)

**Figure 4.47:** normalized GEMpix (zero electric drift field and GEM1 off) and EBT GAF dose distributions measured on two IMRT segment fields at 6 MV with the gamma index maps.

For the first segment, gamma index is minor than 1 in the 74.15 % of the reference points of the corresponding GAFchromic measure. Result is improved respect to the measure obtained with GEM1 on, even if the percentage does not reach yet the usually required 90 %. The halo around the dose distribution of GEMpix contributes to the red areas of non-acceptance. Similar results are obtained for the second segment and again percentage does not reach 90 %, due especially to the halos around the beam shapes. To refine the measures it is possible act on the threshold parameter in order to reduce the halos. Experimental measures are shown in fig. 4.48 where thresholds of 60 and 90 units are applied to the two segments respectively.
Figure 4.48: The Synergy multileaf set-up for two IMRT segment fields at 6 MV (left) with the corresponding dose distributions measured with GEMpix (zero electric drift field and GEM1 off) and EBT gafchromic. Now thresholds of 60 and 90 units over the original ones are applied for the two segments respectively.

Figure 4.49: normalized GEMpix (zero electric drift field and GEM1 off) and EBT GAF dose distributions measured on two IMRT segment field at 6 MV with the gamma index maps (89.98 %). Now threshold is set, respectively, to 60 and 90 units over the original ones.

It is clear that a visible significant improvement has been obtained (90.50 % vs 74.15 % and 77.88 % vs 87.04 %). This demonstrate also that the double-GEM configuration is sensitive enough to thresholds so that long tails which have no dosimetric interest around the principal beams can be cut. In the present case, threshold parameter can be properly calibrated to obtain the correct dose distribution as compared to gafchoromics. These last results demonstrates that the configuration with only two GEMs as gain stages is an optimal choice for 2D dose measurements. Each pixel provides a very stable measure of the ionization charge released in the first transition region. These results are very encouraging considering that a pixel energy calibration has not yet been carried out.
for GEMpix detector. Until now, there are some energy calibrations realized on medipix detector [ref.], but these cannot be transferred to GEMpix photon because of the two different converter materials (silicon and gas). Test pulses can represent a method of energy calibration, because it generates electronically pulses of known charge. However some preliminary tests have not produced satisfying results, because test pulses suffer for the fluctuations of test capacitance which can change from pixel to pixel. Probably the optimal calibration method will be based on a set monochromatic sources which cover a large range of energies. As a result, the method would provide an energy calibration curve for each pixel which would provide the result directly in energy. At the end of this process, the non-uniform responses among all the pixels will disappear providing more refined dose distributions.
Conclusions

This thesis presented the main steps followed in the development of a triple-GEM detector as a real time 2D dosimeter. Condition in which it must work in radiotherapy are unusual respect to the its typical working conditions. The source is a medical Linac which produces pulsed high-energy X-rays beams. In this case an Elekta Synergy Linac has been used: it has 3 spectra corresponding to 3 accelerating voltages (6, 10 and 18 MV) while the beam has a pulsed temporal structure with a pulse width of 3 µsec and a photon fluence per pulse equal to $8.5 \times 10^{12}$ ph/cm$^2$sec. In general a triple-GEM detector can be conceived as a proportional gas detector which can be equipped with different read-out electronics in order to collect different information from the electric signal released in the gas. Considering the triple-GEM chamber itself, it has been demonstrated that its working conditions can be optimized depending on the corresponding read-out electronics. Typically it has some standard set value of electric fields in the gaps between the detector gas layers between the detector mylar window and the first GEM, between the three GEMs and between the last GEM and the anode. The first presented prototype (Chapter 3) has a front-end-electronics based on CARIOCA chip cards and an FPGA mother board. It works in a counting mode and has 128 channels. An high spatial resolution has been obtained realizing a row of 128 PADs having an area or $500 \times 500$ µm$^2$ and a pitch od 700 µm. In this configuration measures have been performed making a scan orthogonal to the PADs row. The PADs row allows to span areas 9 cm wide with a length depending on the scanning range which in the present case is limited to 10 cm. In addition, it has been demonstrated that, working with the standard drift field the detector is affected by a pile-up phenomenon, which can be significantly reduced by switching off the drift field and reducing pad dimensions. This last point has been demonstrated from the first measures realized on an old GEM detector which has $3 \times 6$ mm$^2$ PADs.

Some irradiations have been performed with square fields ($2.4 \times 2.4$ cm$^2$). The obtained results have been compared with EBT gafchromics irradiated in the same conditions, using the gamma index (indicated with $\gamma$) acceptance test. This test combines two criteria: dose difference and distance to agreement. It is commonly accepted that the first must be 3% and the second 3 mm (par. 1.4.4). When $\gamma$ is minor than 1, both criteria are satisfied and, according to the general tolerances accepted in literature, this condition must be valid on the 90% of the dose reference dose points to which a measured dose is compared. For the first GEM prototype, gamma index maps shows that the number of points in which test passes is major than 50% for the energy spectrum of 6 MV and a low applied threshold and major than 60% for the same spectrum and higher thresholds. These percentage are low respect to the commonly accepted minimum value of 90 %. Test fails in the areas of high dose gradients and probably this is affected by the higher sensitivity of GEM detector respect to gafchromic. These results, however, were useful to understand the optimal working parameter of the new GEMpix detector.

These second prototype preserves the same structure of the first GEM detector, but now the area is reduced and the front-end-electronics is substituted by a quad-medipix chip. It is constituted by 512 x 512 pixels, each one having an area of $55 \times 55$ µm$^2$. Then a quad-medipix electronics allows a very high resolution but on a small active area of $2.8 \times 2.8$ cm$^2$. However its electronics can work also in Time-over-Threshold (ToT) mode which allows to make a measure of the charged released in the gas. In this way, the measured charge is correlated more precisely to the dose released in the
gas. Measures realized with zero electric drift field on square beams (8 x 8 mm$^2$) and compared again with the gamma index acceptance test provides best results respect to first GEM detector especially for the lower energy spectrum ($\gamma < 1$ in 96.6% of points). For higher energy spectra, the percentage of the gamma index decrease (69.2% and 40.0% for 10 and 18 MV spectra, respectively). An analogous result is obtained for an IMRT field. High energy fields are affected by large halos which have not a counterpart on the gafchromic. A significant optimization of the results has been obtained working in a bi-GEM configuration with a zero potential applied to the first GEM foil. Now only two GEM are used as charge amplifier. Gamma index percentage now is over 90% and also some tested IMRT fields have an optimal agreement, especially when a threshold cut is applied. Both detectors presents some common advantages respect to the traditional dosimeters:

- They can work on areas of uniform doses where a gamma index $< 1$ is obtained.
- The GEM detector is much more sensitive so that it does not need a build-up material to increase the charge reaching the sensitive volume. The lower limit is represented by the single accelerating pulse. Estimated dose coming from a single pulse to the gafchromic is 4 order of magnitudes lower than that need to have a observable blackening on the film. It can be used in air.
- They have an optimal linearity with dose rate and these results are confirmed also by the currents measured on GEM electrodes;
- Measurable dose range is practically unlimited, it can go from the dose released by a single accelerating pulse until the storage bit limit of the read-out register.
- In general compared to gafchromic dosimeters, it does not need a scanning process to read the measured doses. This means also no fading problems.
- Respect to gafchromics, it does not suffer for UV sensitivity and orientation.
- Spatial resolution realized with GEM detector is at least an order of magnitude higher than that obtained from typical matrix arrays;
- Spatial resolutions of GEMpix is even more higher (55 $\mu$m) and overcomes also the set EBT gafchromic spatial resolution (340 $\mu$m)
- With this spatial resolutions, no interpolation software is need to calculate the intermediate dose values like happens for arrays;
- For GEMpix, the gray levels on the dose distributions obtained can be set acting on some acquisition parameters (like thresholds and Ikrum, for example). This can represent an great advantage because it can be calibrated on known fields acting only on acquisition parameters and without a special software like for EPIDs.
- GEMpix is able to perform fast real time IMRT checking. It has been demonstrated that GEM pix is able to measure also the single pulse. Then IMRT field both in step an shoot and in continuous mode can be reconstructed in time at a very high resolution. Actually there are no device which reach these performances.

All these preliminary results shows the great potentiality of the GEM detectors as a real-time 2D dosimeters, in particular with the GEMpix configuration. The drawback is represented by the active area which is limited to few cm$^2$ while, in radiotherapy, large fields are also applied (for example, 16 x 21 cm$^2$ is maximum area for the Elekta Synergy Linac). However it must be highlighted that spatial resolution like GEMpix (55 $\mu$m) are not necessary for radiotherapy applications. It can be
enough take a spatial resolution similar to that observed for radiochromic films. This means that applying a similar read-out electronics with greater pixels it will be possible to have greater available areas. Actually a new chip (GEMINI) is under design to substitute the old CARIOCA. It will be able to manage the 64 channels, 8 times respect to CARIOCA. In addition it will be able also to make charge measurements which is the optimal choice for this kind of application, like the GEMpix performance have demonstrated in this work. Another important future target which will be considered is the possibility of using air as active material instead of gas mixture. This could be an optimal solution to further simplify the present device. High photon fluences, like those used in radiotherapy, produce an significant ionization in air and GEMpix is high sensitive to detect this ionization. In the future this technology can be implemented to realized transmission dosimetry. Usually routinely control checks are performed before patient treatments. Actually there are some preliminary attempts to realize devices for on line transit dosimetry: from a measure of the dose transmitted through the patient during a treatment session, a software would reconstruct the dose released inside the patient. Until now conventional devices, like matrix array and EPID, have not guaranteed satisfactory and universally accepted results.
44. A. Gandi, “Laboratory of Photomechanical Technique and Printed Circuits”, EST-SMCI Section, CERN, Geneva, Switzerland.
60. GEMpix Detector, https://web2.infn.it/GEMINI/index.php/gempix-detector