Portal Imaging Technology: Past, Present, and Future

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Many different electronic portal imaging devices (EPIDs) have been developed to improve geometric accuracy in radiation therapy. This article describes the two types of EPIDs that have become available commercially—the television camera–based EPID and the matrix ion chamber EPID—as well as describing the amorphous silicon array, a device that may become available in the future for portal imaging. In addition, the various image registration techniques that identify geometric errors from the portal images are described. These include interactive techniques, landmark-based techniques, contrast-based techniques, and hybrid techniques. Although great improvements in portal imaging technology have been made, more development needs to be directed towards making portal imaging convenient and reliable. Image quality must be improved further, to improve the robustness of image registration techniques and more thought must be given to integrating and automating the various steps in the image registration process. Otherwise, too much time will have to be devoted to these tasks. Finally, and most importantly, users will have to decide what is the best way of using EPIDs clinically. Much development is required before the full potential of this exciting technology can be realized.

The goal of radiation therapy is to deliver a prescribed radiation dose to the target volume accurately while sparing the surrounding normal and critical tissues. Experimental and clinical evidence shows that small changes in dose of 7% to 15% can reduce local tumor control significantly,[1-3] or increase the rate of normal tissue complications.[4] As a result, recommendations by the International Commission on Radiation Units (ICRU) suggest that the accuracy in dose delivery be ±5%.[2,3] Such accuracy can be achieved only if field placement is precise during the entire course of radiation treatment, so that the treatment beams irradiate only the prescribed regions.

Unfortunately, the geometric accuracy of radiation treatments can be compromised resulting in the irradiation of regions other than those prescribed. A number of studies[5-12] have shown that discrepancies in field placement occur frequently, especially for complicated treatment setups. Furthermore, these geometric discrepancies can also influence the outcome of treatment.[8,13-15] Fortunately, frequent monitoring of patient positioning can reduce the frequency of discrepancies in field placement.[7-9,16-19] As a result, several studies have suggested that patient positioning be checked daily.[19,20]

Trends in radiation treatments are increasing the need for accurate patient positioning. With the integration of computed tomography (CT) and, on occasion, magnetic resonance imaging (MRI) data into the treatment planning process, with the development of affordable three-dimensional treatment planning work stations, and with the advent of programmable multi-leaf collimators for field shaping, radiation treatment portals have become more highly tailored with smaller margins around the target volumes. In addition, new treatment techniques such as dynamic beam modulation are increasing the need for routine monitoring of patient positioning even further. Because of these trends in radiation therapy, much effort has been devoted to developing more convenient methods to image the patient during radiation treatment, the process known as portal imaging. This article examines the history of portal imaging developments, describes some of the devices (known as electronic portal imaging devices or EPIDs) that are currently available commercially, discusses one of the imaging devices that may become available in the future, and examines some of the image processing and image registration techniques that are essential for EPIDs to be useful in the clinic.

History

There is a long history of imaging with high-energy radiation beams. As early as 1904, radiographs of human hands, mice, purses, and other objects had been made using radium as the radiation source.[21] However, because of the high energy of the radium gamma rays, the images suffered from low contrast,

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which made them unsuited for the diagnostic applications that were of interest at the time. One of the first references to imaging in radiation therapy was by Nielsen and Jensen[22] who, in 1942, described a rotation therapy technique for treating cancer of the esophagus. In the treatment, the patient sat upright in a rotating chair while a stationary x-ray beam was directed horizontally towards the patient. The therapeutic radiation (180 kV(p) x-ray beam) exiting the patient hit a fluorescent screen that was viewed by an observer who was looking through a lead glass window. Not only did the observer view the treatment in real time, but corrections to the position of the beam were made remotely during the treatment. Thus, this treatment may have been one of the earliest examples of dynamic conformal radiation therapy! By 1951, a similar treatment for cancer of the esophagus had been developed by Hare et al[23] using a 2-MeV Van de Graaff generator. They described the use of film radiography to ensure accurate positioning of the patient before commencement of the rotation therapy. They showed a great deal of sophistication in their portal film activities, investigating the utility of introducing air into the bladder and rectum as a contrast agent as well as using double-exposure radiographs to visualize anatomy outside of the treatment field. Surprisingly, during the 1950s most of the scientific interest in imaging with megavoltage beams dealt with diagnostic applications. Several articles by Tuddenham et al[24,25] promoted the use of megavoltage x-ray beams for chest radiography to optimize exposure in both the mediastinum and the lung regions and to minimize the visibility of overlying "osseous" structures. In 1960, Perryman et al[26] described cobalt 60 radiography. The technique used Kodak type AA industrial film (Eastman Kodak, Rochester, NY) placed in a cassette where the standard intensifying screens had been replaced by two 0.01-inch lead sheets. The only drawback was that film development took approximately 30 minutes to complete. Not only were the radiographs used for treatment localization but also to give diagnostic information "regarding the extent and location of soft tissue lesions." Finally, in 1962, Springer et al[27] suggested an improvement to cobalt 60 radiography where two fluorescent screens were placed between the lead sheets and the film. This modification reduced the exposure time and, it was claimed, increased the contrast of the final radiographs.

About this time, non-film imaging methods started to be introduced into radiation therapy clinics. In 1958, several articles[28,29] described the use of a television-roentgen (TVR) system for monitoring the position of patients during pendulum therapy. The patients lay on the treatment couch while the gantry-mounted TVR system (consisting of a x-ray image intensifier and a TV camera) rotated around the patient during the treatment with a 200-kV(p) x-ray beam. The video signal was sent to a monitor located in the control room of the treatment machine. The major limitation was that this TVR system had a field of view of only 5.0 inches (2.5 inches at the patient). Independently, a somewhat similar system was being developed by Andrews et al[30] using a TV camera–based device called a "Johns Hopkins screen intensifier"[31] as the image receptor. It consisted of a fluorescent screen viewed by an image orthicon TV camera using a complicated series of relay mirrors and lenses known as a Schmidt optical system. These efforts resulted in some of the earliest megavoltage images, using a 2-MeV Van de Graaff generator as the x-ray source. Not surprisingly, the authors found that normal tissue contrast was insufficient and they had to use air or mercury contrast materials to visual anatomic structures. In 1962, another article[32] described an imaging system that used an imaging device similar to the Johns Hopkins screen intensifier. The field of view at the detector was 25 cm in diameter and the system was attached to a 30-MV betatron. Again, only high-contrast objects such as tungsten, gold, and lead markers produced sufficient contrast to be visualized by the system. The investigators showed great ingenuity in introducing high-contrast, high-density objects into various body orifices such as the esophagus. However, the limited image quality and the need for contrast agents reduced the utility of all of these imaging systems and they never came into wide spread clinical use.

One of the most important developments for portal imaging was first described by Swain and Steckel in 1966[33] and was refined further by Marks and Haus and their colleagues.[34-36] The method used a slow, wide-latitude film that was placed in cardboard film holders and was exposed for the entire duration of the treatment. Unlike portal film techniques common at the time, the patient was not moved between the exposure of the film and the treatment and the cardboard film-holders were much more comfortable to lie on than the film-screen cassettes that were typically modified for therapy verification. Not only was much less effort required to treat the patient but, in addition, a record of the

entire treatment was available from the film. Because of the efforts of Marks and Haus, a film (eventually known as Kodak XV-2 film), which was compatible with the 90-second film processors[37] that had been introduced into hospitals in 1965, became available for therapy verification. This was a major development for portal imaging. Not only did Marks and Haus and their colleagues develop a convenient method of verifying patient positioning during radiation treatment but they also showed in a series of studies[7-11] the importance of routine verification in reducing errors in patient positioning and in improving local control. As a result, use of portal films became routine and eventually generated the demand for even more convenient methods of therapy verification that has since lead to the development of EPIDs.

Commercially Available Imaging Devices

Many different devices have been examined since the early 1980s as alternatives to film. These devices can be divided into two categories: scanning systems, where the radiation detector subtends only a small fraction of the radiation beam and must be scanned underneath the patient to form the image, and area systems, where the detector subtends the entire radiation beam. Examples of these devices include scanning diode arrays,[38-40] scanning scintillator arrays,[41] storage phosphors,[42-44] coded aperture arrays,[45] matrix ion chamber systems,[46,47] and TV camera–based systems.[48-54] Insufficient space is available to discuss all of these systems and readers are referred to two comprehensive reviews of portal imaging devices for further details.[55,56] The following discussion concentrates on the matrix ion chamber and the TV camera–based EPIDs, which are both available commercially, as well as the amorphous silicon array,[57-72] which is an imaging device that may become available commercially in the future.

Matrix Ion Chamber

A schematic of the matrix ion chamber device, which was originally developed by Meertens and van Herk and their colleagues[46,47] is shown in Fig 1. The device consists of two sets of electrodes that are oriented perpendicularly to each other separated by a 0.8-mm gap, which is filled with a fluid (2,2,4-trimethylpentane) that is ionized when the device is irradiated.

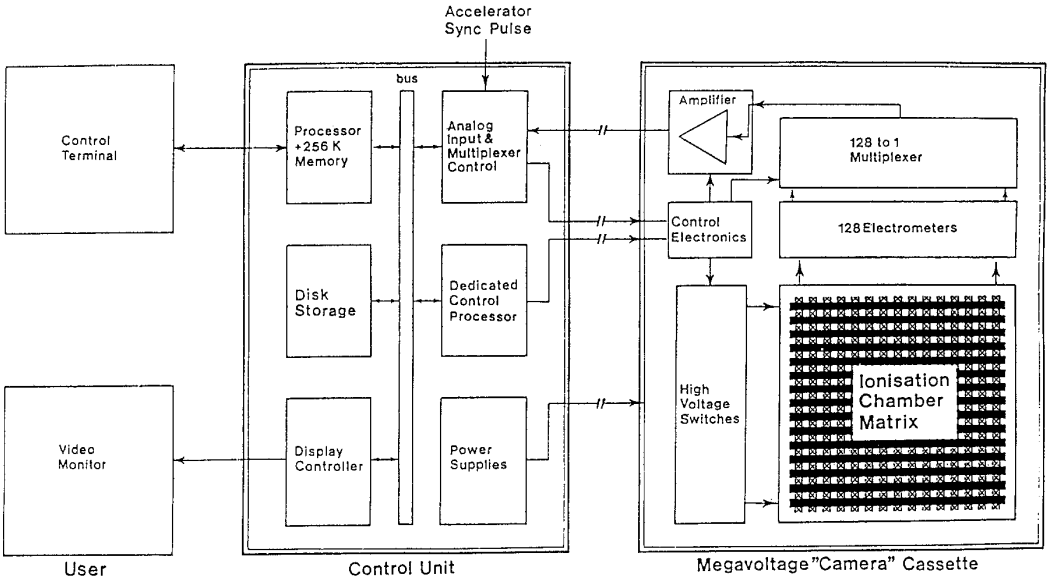

Figure 1. Schematic diagram of the matrix ion chamber. The device consists of a set of 256 signal electrodes and a set of 256 high-voltage electrodes oriented perpendicularly to each other and separated by a 0.8-mm gap filled with a fluid called 2,2,4 trimethylpentane. When irradiated, the fluid is ionized and generates signals in the signal electrodes. The active area of the device is 32.5 × 32.5 cm and its overall dimensions are approximately 60 × 60 × 5 cm including the readout electronics, which are immediately adjacent to the ion chambers. The device is controlled by a computer located in the control area of the linear accelerator. (Reprinted with permission.[46])

The electrode spacing is 1.27 mm and, since each set of electrodes consists of 256 electrodes, the active area of the matrix ion chamber array is 32.5 cm on a side. One set of electrodes is connected to 256 electrometers and the other set of electrodes is connected to a high-voltage supply that can apply a 300-V potential to each electrode individually. The matrix ion chamber array is read out by applying a high voltage to each of the high-voltage electrodes in succession (for approximately 20 milliseconds) and measuring the signal generated in each of the 256 signal electrodes. This procedure takes 5.5 seconds to read out an image. In addition, a fast (lower resolution) scanning mode is available that scans the array in 1.5 seconds by applying the high voltage for a 10-millisecond period to two high-voltage electrodes at a time. The fast acquisition mode is useful for acquiring double-exposure images.

The most obvious advantage of the matrix ion chamber is its compact size, which makes the device a convenient replacement for film cassettes. Another advantage is geometric reliability—images acquired with this EPID have no geometric distortions. Furthermore, unlike other scanning EPIDs, the matrix ion chamber has no moving parts, reducing the likelihood of mechanical problems.

The major limitation of most EPIDs that use a scanning radiation detector, such as the matrix ion chamber, is quantum utilization. Ideally, an image receptor should use all of the available radiation efficiently (even for megavoltage imaging) because this will improve image quality. Clearly, this is not the case for the matrix ion chamber, where only one high-voltage electrode (out of 256) is active at any one time. However, the physics of signal generation in the 2,2,4 trimethylpentane improves the quantum utilization of the matrix ion chamber considerably.

The signal measured by the matrix ion chamber depends on the rate of formation and the rate of recombination of the ion pairs that are generated in the ionizing fluid. Even when no high voltage is applied to the electrodes, the rate of recombination of the ion pairs generated in the 2,2,4 trimethylpentane is relatively slow. Therefore, the concentration of ion pairs can increase over a period of time until an equilibrium is reached between ion-pair formation, which depends on the dose rate at the matrix ion chamber, and ion-pair recombination, which depends on the probability of ions encountering each other and is proportional to the square of the ion-pair concentration. The signal measured by any electrode of the matrix ion chamber does not depend greatly on the dose rate during the 20-millisecond period when the high voltage is applied but on the previous irradiation history of the electrode. Calculations have shown that after 0.5 second, a latent image has been formed over the entire irradiated region of the matrix ion chamber and that irradiating for a longer time will not increase the size of the signal, ie, will not improve image quality. These observations have both positive and negative implications. The measured signal is six to seven times greater than would be expected if no charge integration occurred in the 2,2,4 trimethylpentane. However, the effective period of the charge integration ($\sim$0.5 second) is still short compared with the total image acquisition time of 5.5 seconds. Therefore, a large fraction of the radiation that interacts with the matrix ion chamber does not generate any measurable signal. For this reason, the matrix ion chamber requires higher doses to generate images than other portal imaging devices.

An example of an image acquired with the matrix ion chamber EPID is shown in Fig 2. One of the most noticeable characteristics of this image is how extensively the raw signals have to be processed before yielding a usable image. (The artifacts in Fig 2 represent an extreme case because these images were acquired using a prototype device.) Because spurious signals are generated in the electrometers and ion chambers, and because the sensitivities of each ion chamber can vary, the device must be calibrated routinely. In addition, because the matrix ion chamber is a scanning EPID, it is more susceptible to artifacts if the dose rate of the accelerator changes during image acquisition. Thus, the radiation beam has to stabilize for some period (typically 1.0 second) after startup before image acquisition can begin.

TV Camera–Based EPIDs

The design of TV camera–based EPIDs, which is shown in Fig 3, uses technologies that have long been used for other applications. The x-ray detector consists of a metal plate to which is attached a gadolinium oxysulphide (Gd_2O_2S) screen similar to the screens used in diagnostic radiology. When irradiated, high-energy electrons generated in the metal plate and the Gd_2O_2S screen are converted into light. The light that diffuses through the screen and exits the rear surface of the x-ray detector is viewed by a TV camera using a 45° mirror. The video signal from the camera is digitized and the digitized image can

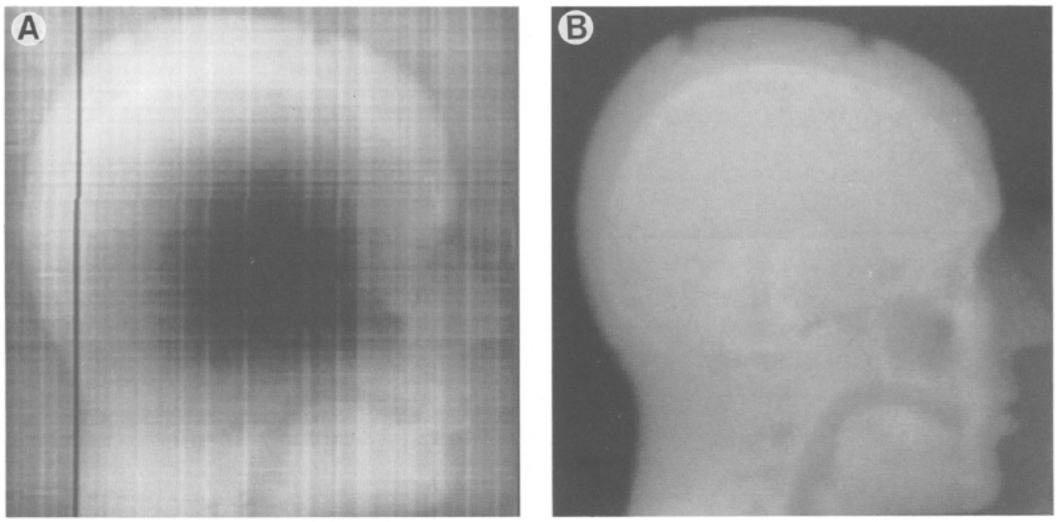

Figure 2. Image of an Alderson Rando phantom acquired using the matrix ion chamber EPID and an 8-MV x-ray beam: (A) before and (B) after corrections for variations in electrometer offset and sensitivity as well as variations in electrode shape and fluid thickness. (Reprinted with permission.[46])

be viewed on a monitor located in the control area of the accelerator.

This design has one major advantage. The x-ray detector subtends the entire area of the radiation beam so all of the radiation that exits from the patient has the potential to generate a signal in the EPID. However, the design suffers from one major limitation and this is the light collection efficiency of the optical chain. Figure 4 shows the problem schematically. Because the light is highly scattered within the phosphor screen, the light is emitted from the rear of the screen in all directions with equal probability. Only those light photons that are emitted within a small cone subtended by the lens of the camera can generate a signal in the TV camera; typically only 0.1% to 0.01% of the light emitted by the phosphor screen reaches the TV camera. This poor light collection efficiency can reduce image quality in two ways. Firstly, if an x-ray photon interacts in the x-ray detector and none of the light generated by this interaction reaches the TV camera, then no measurable signal is produced. Secondly, if only a small signal is produced in the TV camera then noise generated by the preamplifier and other electronics

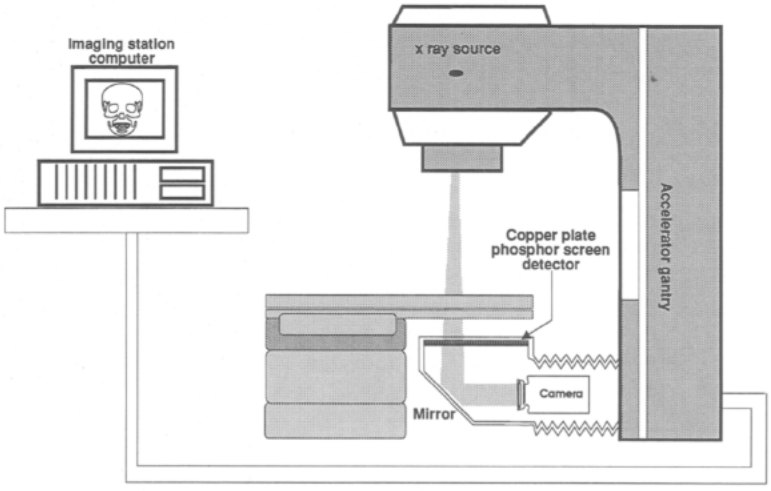

Figure 3. Schematic diagram of the TV camera–based EPIDs. These devices use a metal plate/phosphor screen as the x-ray detector. Light emitted by the phosphor screen is viewed by a TV camera using a 45° mirror.

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