

[54] RADIOGRAPHIC IMAGING

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[51] Int. Cl. G01t 39/18

[58] Field of Search 250/71.5 R, 213 R

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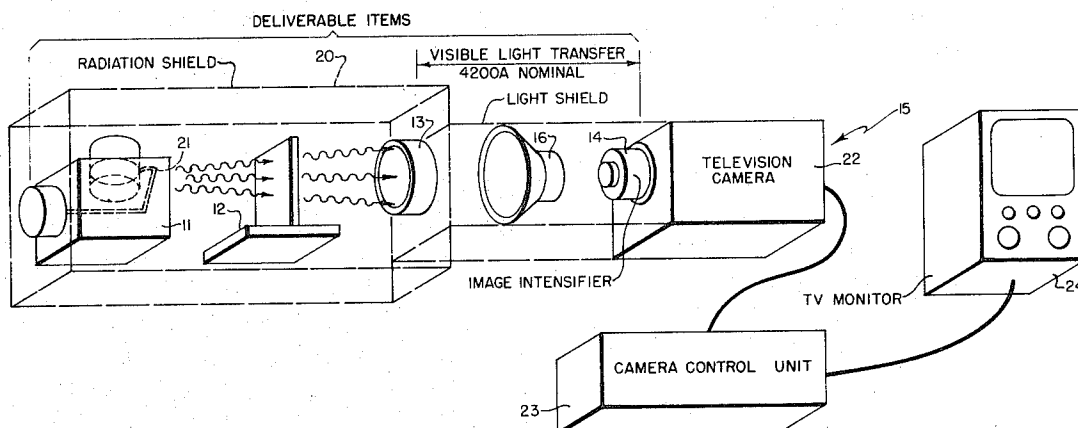
Attorney, Agent, or Firm—Charles Hieken; Jerry Cohen

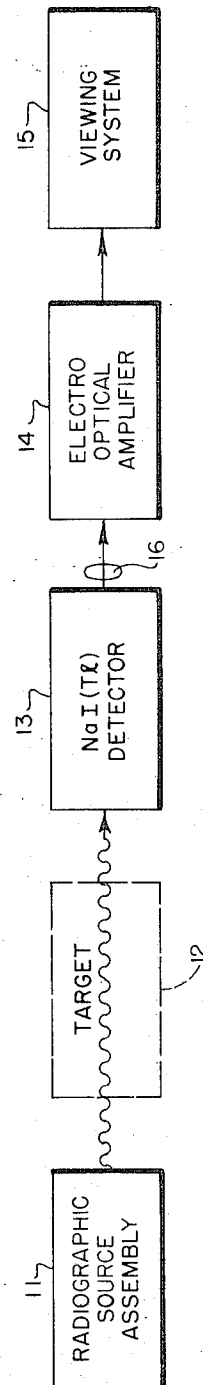
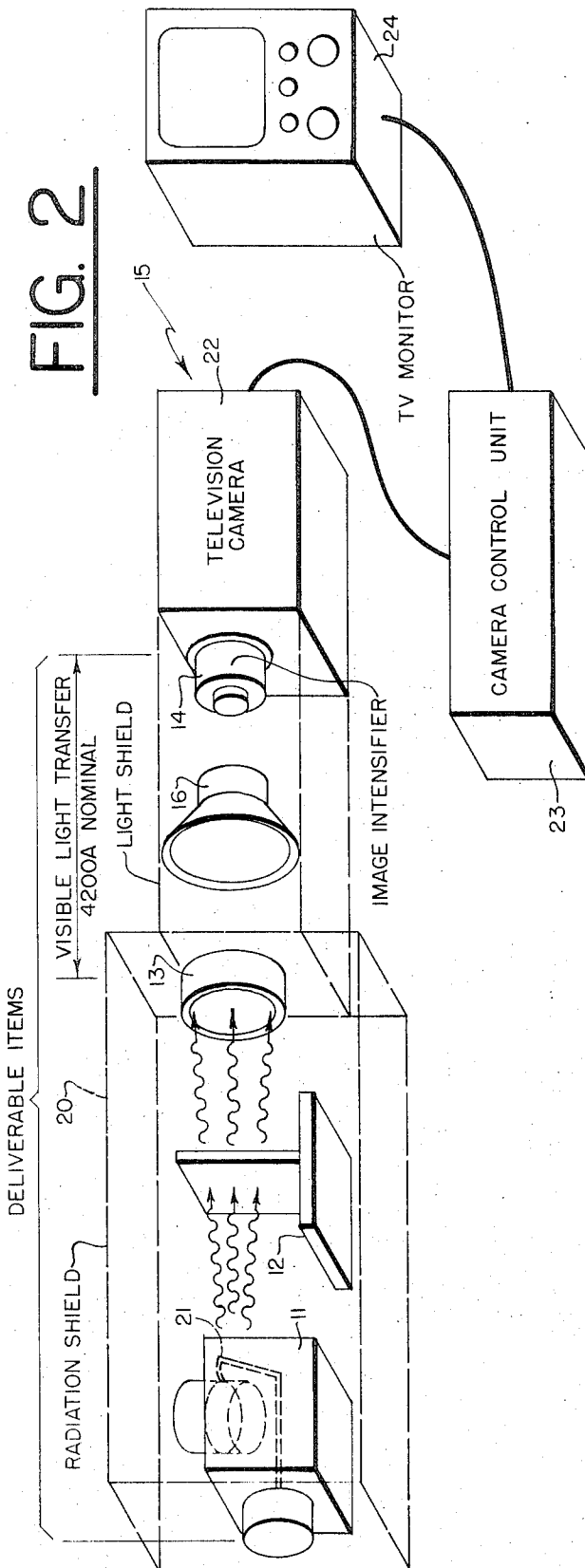
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ABSTRACT

An X-ray imaging system includes a thin sodium iodide single crystal or matrix of single crystals for converting an X-ray image into a visible image. Preferably an electro-optical amplifier amplifies the detected image and delivers the amplified image to a closed circuit television viewing system.

10 Claims, 13 Drawing Figures





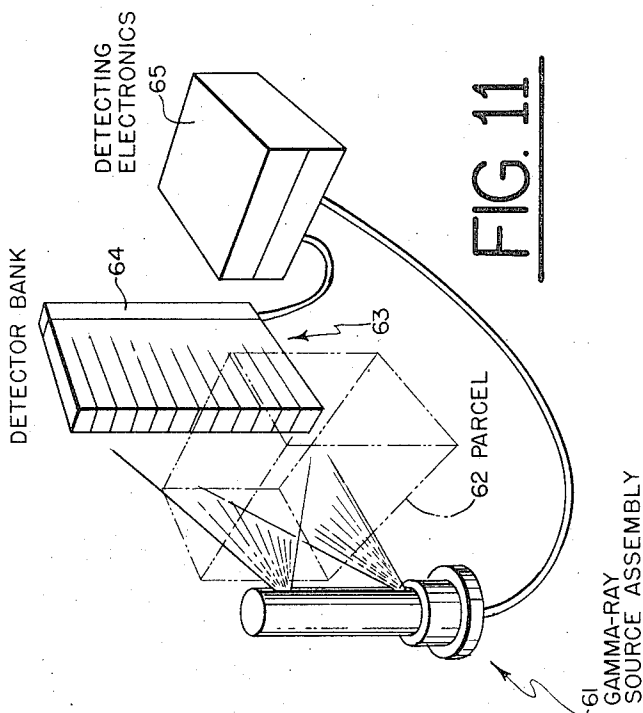
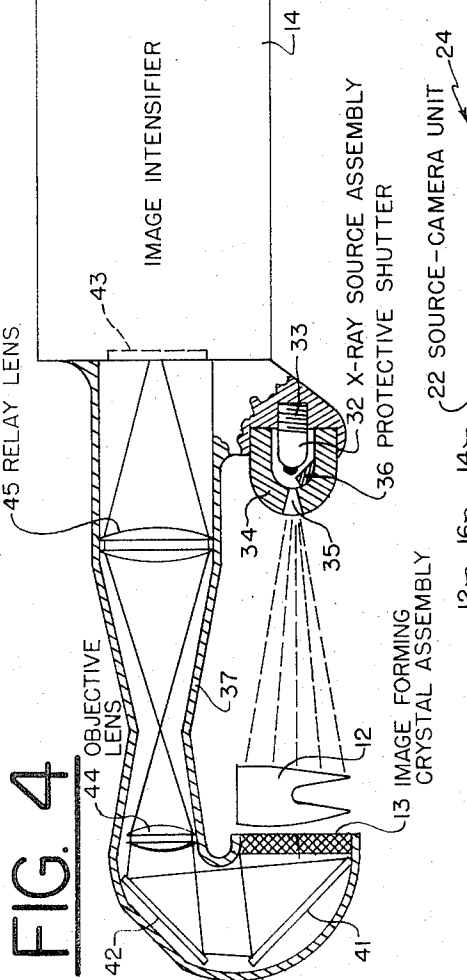
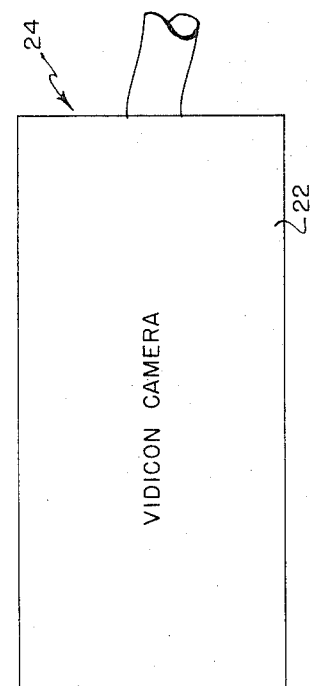


FIG. 4

FIG. 3

FIG. 11

FIG. 5

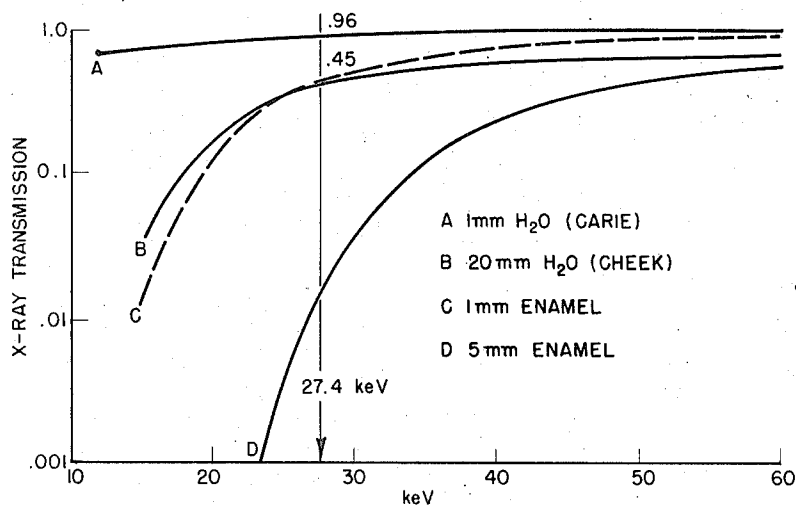
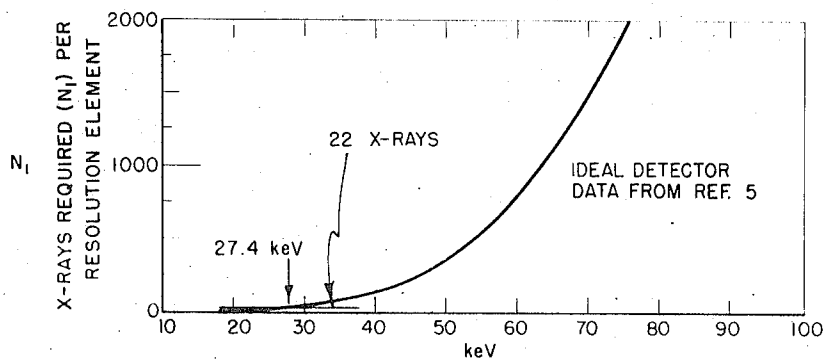


FIG. 6

FIG. 7

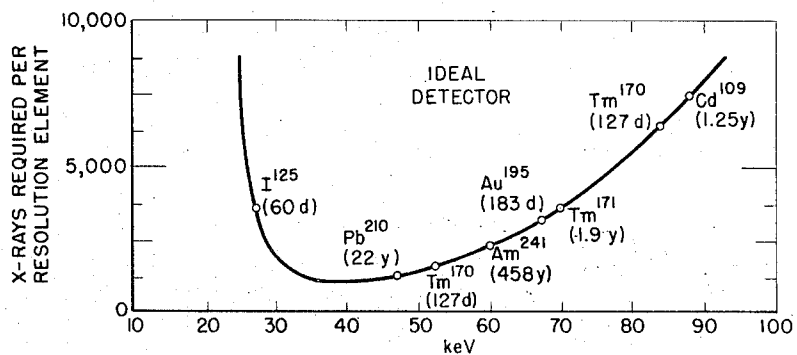


FIG. 8

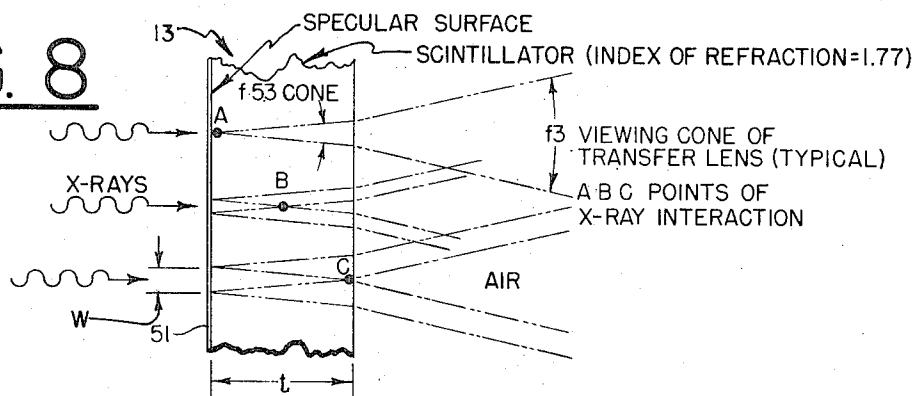


FIG. 9

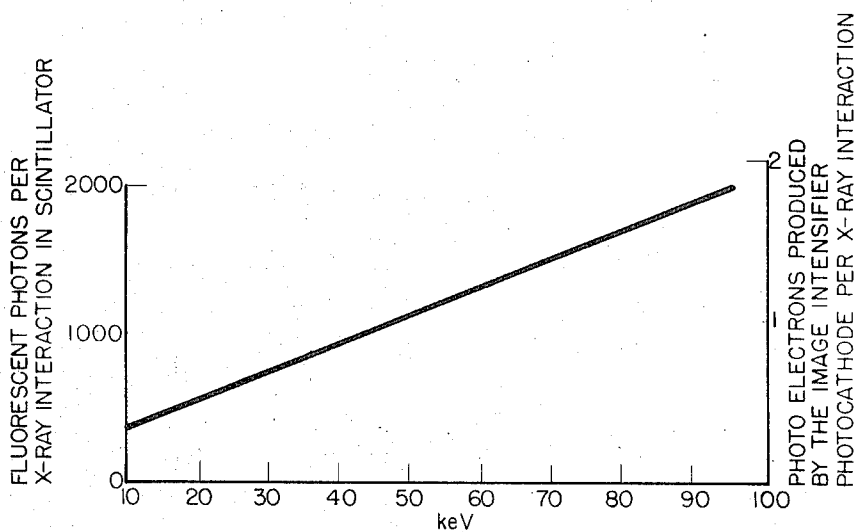
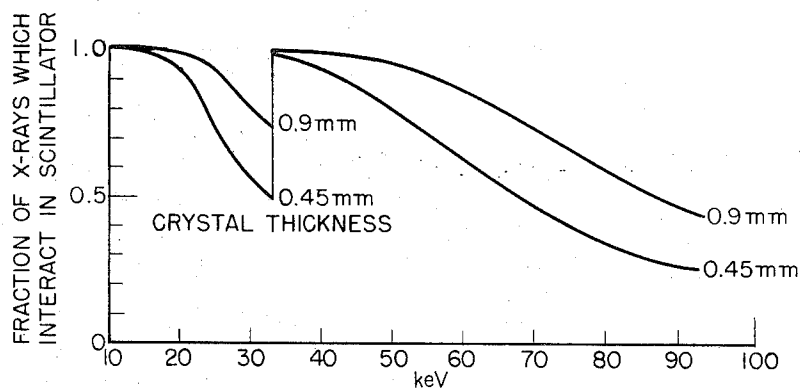


FIG. 10

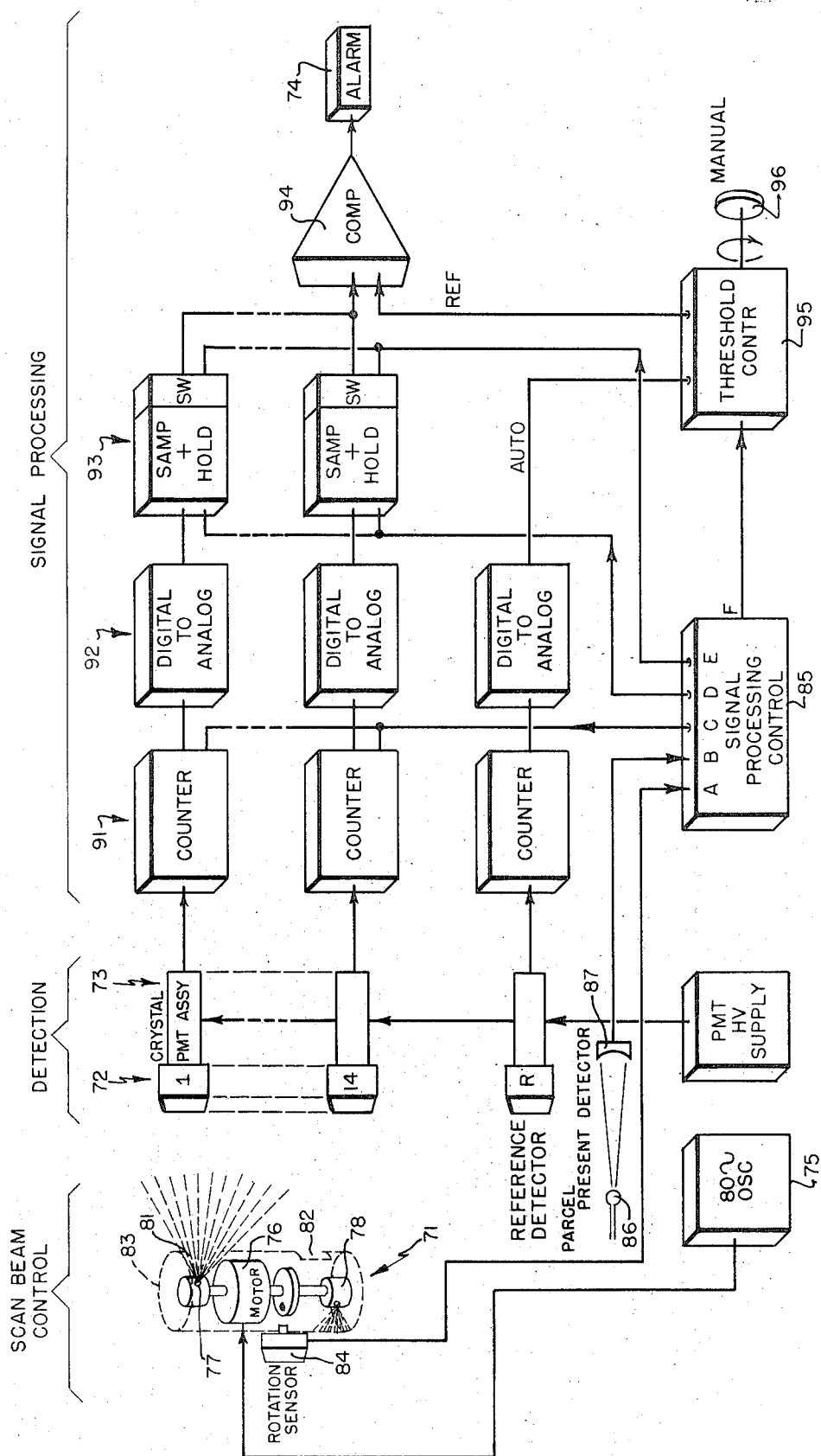


FIG. 12

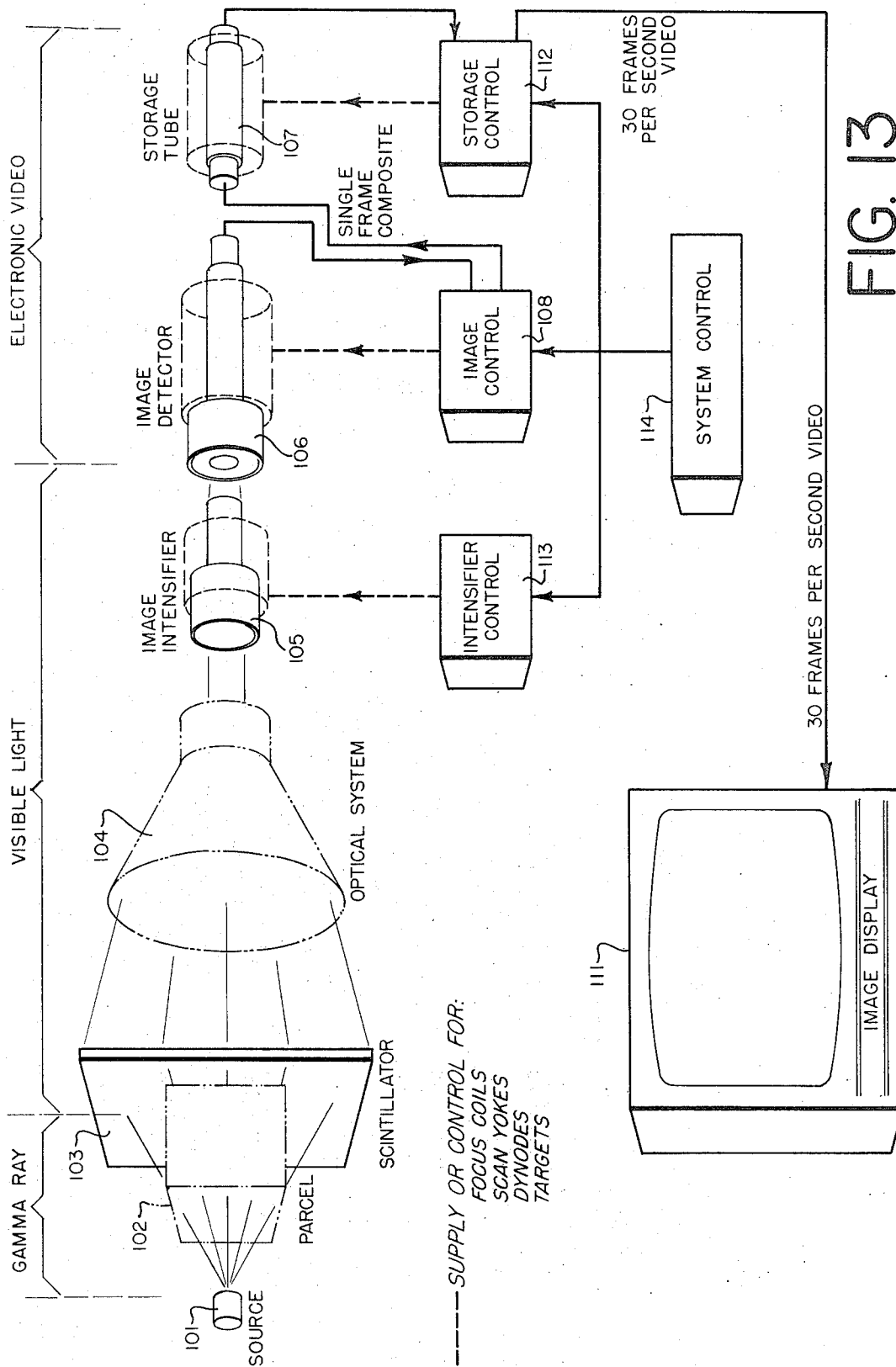


FIG. 13

RADIOGRAPHIC IMAGING

BACKGROUND OF THE INVENTION

The present invention relates in general to radiographic imaging and more particularly concerns novel apparatus and techniques for radiographic imaging characterized by low radiation levels, compact apparatus and direct real-time observation of the radiographic imagery.

Innovations in general medical radiology have led to typical decreases in the dosage required for direct exposure of X-ray film and/or phosphor screens of the order of several hundred. Photographic film emulsions are effective photon detectors, but they require X-ray exposure levels much above that required for information transfer because a sufficient number of emulsion grains must be activated to produce the picture density required for direct observation.

Although electro-optical image intensification is known, the application of this technique to dental radiology has been limited by the more confined geometry of a dental examination. According to one prior art approach a coherent fiberoptic bundle transmits the visual image produced at the tooth in a phosphor layer to a remote image-intensifier/vidicon camera. A phosphor is located in the image plane, and a pulsed electron beam machine produces X-rays.

Still another prior art approach uses physically small yet highly radioactive sources of radioactive Iodine-125 in conjunction with X-ray film. A shortcoming of this film-based system is the relatively long time required to expose X-ray film to an easily readable optical density. Present source technology requires times of the order of minutes for exposure while values of one second or less are considered desirable so that subject motion does not significantly effect picture diagnostic quality. A system that solves some of the problems enumerated above also has utility in a package or human examining system searching for dangerous objects, such as guns, knives and bombs. Such a system should present no hazard to operating personnel, expose the package or person being examined to negligible radiation doses and be reliable and compact.

Accordingly, it is an important object of the invention to overcome one or more of the problems enumerated above.

It is another object of the invention to provide a radiographic system that is compact, sensitive to small doses of radiation and yet produces a useful image of hidden objects in real time.

It is another object of the invention to achieve one or more of the preceding objects with a sensitive radiographic-energy-to-visible image transducer that is compact and relatively inexpensive.

It is a further object of the invention to provide a compact system in accordance with one or more of the preceding objects suitable for use as a portable dental radiological system.

It is a further object of the invention to achieve one or more of the preceding objects with a system that is relatively easy and inexpensive to fabricate and capable of operating with a minimum of skilled attention.

SUMMARY OF THE INVENTION

According to the invention, a radiographic system comprises a source of radiographic energy and means

including a thin sodium iodide crystal for converting the radiographic energy into a visible image after the radiographic energy from the source passes through an object. The source may be a low dosage isotope source, such as radioactive iodine-125, conventional X-ray tube or other suitable source. Preferably the system includes a closed circuit television system for amplifying and displaying the visible image of the detector upon the face of a display tube for real time observation. The image may also be intensified and recorded on film.

According to a specific aspect of the invention especially useful for scanning parcels, there is means for scanning the object to be examined. Preferably this means comprises means for scanning the object to be examined with the radioactive source and a number of contiguous ones of said detectors, each with an associated photoelectric transducer for converting the light signal from the associated sodium iodide detector into a corresponding electrical signal. In a preferred form of culling station according to the invention the output of the photo-electric transducers is compared with a reference signal derived from a reference detecting assembly to provide an indication when an unusual item is sensed in the parcel being scanned.

Numerous other features, objects and advantages of the invention will become apparent from the following specification when read in connection with the accompanying drawing in which:

BRIEF DESCRIPTION OF THE DRAWING

FIG. 1 is a block diagram illustrating the logical arrangement of a system according to the invention;

FIG. 2 is a combined block-pictorial diagram of a portable film recording dental radiographic unit;

FIG. 3 is a combined pictorial-block diagram of a preferred source-camera assembly for a dental radiographic system that is exceptionally compact and easy to use;

FIG. 4 is a more detailed view of the dental system source camera unit;

FIG. 5 is a graphical representation of the contrast between cavity and enamel images as a function of X-ray energy representative of required X-ray energy for a given contrast;

FIG. 6 is a graphical representation of X-ray transmission as a function of energy;

FIG. 7 is a graphical representation of detection difference requirements as a function of radioisotope source energy;

FIG. 8 illustrates the geometry of photons produced in the detector crystal;

FIG. 9 is a graphical representation of sodium iodide crystal detector characteristics;

FIG. 10 is a pictorial representation of photon production as a function of energy;

FIG. 11 is a pictorial representation of a parcel culling station according to the invention;

FIG. 12 is a combined block-pictorial representation of an automatic parcel culling station according to the invention; and

FIG. 13 is a combined block-pictorial diagram of a parcel imaging station according to the invention.

DETAILED DESCRIPTION OF PREFERRED EMBODIMENTS

With reference now to the drawing and more particularly FIG. 1 thereof, there is shown a block diagram il-

lustrating the logical arrangement of a system according to the invention. A radiographic energy source assembly 11 emits radiographic energy upon target 12 to produce an image on sodium iodide detector 13 that is amplified by electro-optical amplifier 14 and then displayed on viewing system 15, such as a television picture tube.

The source assembly 11 typically comprises a small diameter source, a housing, a shutter and a radiation-shaping aperture. Source assembly 11 functions to properly illuminate target 12 (a tooth or parcel for example) and controls exposure for minimum operational dosage. The maximum upper limit of dosage may be set by appropriately choosing the radioisotope and setting its initial activity level.

Sodium iodide detector 13 is a thin crystal that converts X- and gamma-radiation into a form of radiant energy that can be electro-optically manipulated. This material is an exceptionally efficient passive detector having the property of emitting violet light when irradiated with gamma or X-rays. By making the crystal thin the crystal reproduces spatial details contained in the probing X- or gamma-ray in violet light that could be visually observed unaided for sufficiently high X- or gamma-ray activity level.

However, to keep radiation levels low, the system preferably includes the electro-optical amplifier 14 using known techniques to achieve electron multiplication and input-output phosphor planes to intensify the image. Simple optical lenses 16 guide the light from the detector crystal 13 to the input surface of the image intensifier 14.

Viewing system 15 comprises a television camera connected to a monitor that may provide some light amplification to aid vision but primarily functions to provide an easily viewed enlarged image at a location convenient to the analyst. Alternately, the image from image intensifier 14 may be recorded on photographic film.

Referring to FIG. 2, there is shown a combined block-pictorial diagram of an exemplary system according to the invention. In FIG. 2 and throughout the drawing corresponding elements are identified by the same reference numeral. Radiographic source assembly 11 includes a shutter 21 that selectively emits radiographic energy that passes through target 12 to reach sodium iodide detector crystal 13 to form an image that is focused by lens 16 upon image intensifier 14. A radiation shield 20 preferably encloses source assembly 11, target 12 and detector crystal 13.

Viewing system 15 comprises a television camera 22, a camera control unit 23 and a television monitor 24' intercoupled by the connecting cables.

Referring to FIG. 3, there is shown a preferred form of the invention for use as a portable dental radiographic unit. The source-camera unit 24 includes the radiographic source assembly 11, the sodium iodide detector crystal 13, the image intensifier 14, the lens 16 and the vidicon camera 22. All the elements of the system fit in the case 26. The system includes camera control 23, television monitor 24, power converter 27 for converting low battery voltages to higher operating potentials and image storage unit 31, which permits a radiographic image to be stored after a short exposure and then viewed continuously on television monitor 24. A system for storing radiographic images on a storage

tube and then viewing it on a television monitor is known in the art and not a part of this invention.

Referring to FIG. 4, there is shown a more detailed view of source-camera unit 24 with portions in diametrical section to illustrate certain features of the invention. Radiographic source assembly 11 includes a radioactive source 32 surrounded by lead rear and front shields 33 and 34, respectively. Front shield 34 is formed with an aperture 35 that is covered by a protective shutter 36 except when exposing a target. Sodium iodide detector crystal 13 is seated as shown at the entrance of a generally J-shaped light pipe 37, preferably of radiation shielding material, facing source 11 for positioning immediately adjacent to target 12, in this case a tooth being X-rayed.

Light pipe 37 includes a lower mirror 41 and an upper mirror 42 at 45° angles relative to the plane of sodium iodide detector crystal 13 to reflect the image on crystal 13 upon the input 43 of image intensifier 14 through objective lens 44 and relay lens 45. With the construction shown the source camera unit 24 may be of the order of only a foot long and less than 2 inches in diameter.

Having discussed the physical arrangement of a system according to the invention, it is appropriate to consider the principles by which the invention produces diagnostic radiographic images of quality comparable to those achieved with contemporary dental X-ray equipment.

The definition of image quality is highly subjective because it involves the ability of an observer to resolve fine detail and depends on the physiological response of the eye and is subject to many variables in the presentation of the image.

In evaluating system capabilities it is helpful to define thickness, density and absorption coefficients of damaged and sound tooth structures in addition to general performance criteria, such as the minimum spatial resolution and visually detectable contrasts. The size of the minimum resolvable tooth structure for diagnostic radiology is important because the required source strength is proportional to the number of resolution elements which make up the image field. Yet, the dimensions of the source are limited by the resolution requirement and the exposure geometry. A preferred source balances these requirements consistent with the specific activity of suitable radio isotopes.

A convenient theoretical framework for such a study is described in an article by Carl O. Henrickson entitled *Iodine-125 as a Radiation Source for Odontological Roentgenology* in ACTA RADIOLOGICA SUPP. 265. (1967). That article presents the pertinent physical parameters for the problem of the detection of a demineralized caries defect and defines the minimum acceptable contrast as a difference in the image density of the cavity and adjacent areas which is statistically significant to two standard deviations. The resolution requirement can be estimated from the observation that the panoramic images produced by that author in conjunction with others using a 0.5 mm diameter Iodine-125 source placed inside the mouth are of acceptable quality for diagnostic purposes. That work is described in a paper by Beronius et al. entitled *The Use of Iodine-125 as an X-ray Source in Roentgen Diagnostics* in 13 INTERN, J. APPL. RADIAT. 253 (1962). In this case the source-film distance is only about three times the maximum object-film distance so that the penum-

bra produced by the finite width of the source broadens the image of each point in the highest object plane by 0.17 mm. Such a resolving power provided by the present system is adequate.

The cheek and tooth under examination attenuate X-rays provided by the source for the detector. The detector crystal converts most of these X-rays to visible light quanta. The fraction of the X-rays which interact in the detector crystal and the number of photons produced in each such interaction along with the geometric efficiency of the optical system at the photoelectron yield are relevant.

A basic limitation of the observable contrast with the present invention is the statistical fluctuation in the number of photoelectrons produced in each resolution element at the photocathode of the image intensifier 14. The remainder of the system may have gain sufficiently high to contribute negligibly to statistical error. By determining the incident flux of X-rays required to overcome the fluctuation and combining that with a number of resolution elements, the strength of the Iodine-125 source to produce the desired image may be determined.

The Henrickson article cited above discusses the effects of the imaging detector on the required flux of X-rays. While this discussion contemplates X-ray film as the detection medium, it also presents the values of the minimum flux required to produce the desired difference for the case of a perfect detector in which the contrast-limiting factor is the fluctuation in the arrival of the X-ray quanta. These minimum flux value determinations are applicable to the present invention by taking into consideration the deviations of the present system from the characteristics of a perfect detector. An important feature of the present invention is that the sodium iodide crystal in combination with the image intensification system according to the invention approaches the ideal detector much more closely than film.

For an ideal detector the desired contrast may be defined in terms of the difference between the number of X-rays, N_c , which emerge from the path containing the cavity and the corresponding number, N_e , for the path containing just enamel. An acceptable contrast criteria is the number of X-rays, N_1 , required to be incident on a 1-millimeter section of the path to produce a difference between N_c and N_e which is twice the estimated statistical fluctuation error. Referring to FIG. 5, there is shown a graphical representation of this difference as a function of X-ray energy.

Referring to FIG. 6, there is shown a graphical representation of X-ray transmission as a function of energy through 1 millimeter of caries, 20 millimeters of cheek, 1 millimeter of enamel and 5 millimeters of enamel. The graphical representations of FIGS. 5 and 6 will be helpful in analyzing a specific example.

Consider the situation at 27.4 keV. At this energy $N_1 = 22$, the number of X-rays required to be incident on the one mm section. From FIG. 6 ($0.96 \times 22 = 21$) are transmitted through the cavity while only $0.45 \times 22 = 10$ survive passage through the enamel. The difference $N_c - N_e = 11$, and its estimated error due to statistical fluctuation is $\sqrt{10 + 21} = 5.5$, so that the difference is twice the estimated error.

For an ideal detector the number of X-rays which the source must emit into each resolution element may be determined by dividing the values shown in FIG. 5 by

the combined transmission of the cheek and the 5 mm of enamel, common to both paths, corresponding to the product of curves B and D in FIG. 6. The results are graphically represented in FIG. 7 and are helpful in evaluating the source strength requirement for various radio isotopes. This graphical representation shows a minimum number of photons per resolution element is required at about 40 keV energy.

Since the present invention does not use a perfect detector, the flux levels in FIG. 7 are increased by a factor related to the deviation of the system from perfection. Some of the X-rays pass completely through the detector crystal without interaction. The fraction which thus penetrates depends upon the crystal thickness and the X-ray energy. The crystal thickness is preferably chosen at the largest value which will not compromise resolution.

Referring to FIG. 8, there is shown the optical geometry at the detector crystal. The detector crystal 13 includes a specular reflecting surface 51 upon which the X-rays are incident. The surface might also be absorbing black to provide better resolution and half the number of photons. Each interaction produces a large number of visible photons, but only a small fraction of these are collected by the optics and focused on the image intensifier photocathode 43. The transfer lens 16 is focused upon the mirror surface 51 with the f3 viewing cone reduced to f5.3 within the sodium iodide because its index of refraction is 1.77. The optical system accepts light quanta emitted into the internal f5.3 cones in either the forward or backward direction as shown for the hypothetical interaction in the center of the detector crystal in FIG. 8. The specular surface 51 reflects the backward cone of light forward.

The preferred maximum thickness of crystal 51 may be calculated from the resolution requirement. The largest blur is produced by interactions which occur in the crystal at the maximum distance from the focal plane. Since the focal plane corresponds to that of the specular surface 51, this maximum distance occurs at point C in FIG. 8 with the resolution element width W being the thickness t of the crystal divided by the f number, 5.3. Thus, with a resolution requirement of 0.17 mm, the crystal may be as thick as 0.9 mm.

Referring to FIG. 9, there is shown a graphical representation of the fraction of the X-rays which interact in the detector crystal for thicknesses of 0.9 and 0.45 millimeter as a function of X-ray energy. The sharp discontinuity at 33 keV is due to the K-edge of the iodine in the crystal. Note that interaction efficiency approaches unity for the thicker layer over most of the energy range. Thus, by dividing the optimum fluxes shown in FIGS. 5 and 7 by the fraction of X-rays which interact, such as shown for sodium iodide of the indicated thicknesses in FIG. 9, and considering the statistical nature of the interaction in the crystal, the statistical significance of the difference ($N_c - N_e$) may be maintained.

Referring to FIG. 10, there is a graphical representation of photons produced per X-ray interaction in the detector crystal and photoelectrons produced by the image intensifier photocathode, both as a function of X-ray energy. The optical system collects approximately $(2\pi/4) (t^2/4\pi f^2)$ where t is the crystal thickness and f is the effective f number in the crystal, or about 4.5×10^{-3} of the light quanta from each interaction for

the crystal of FIG. 8 with a thickness of 1 millimeter and for $f/5.3$.

The reflective losses at each of the optical surfaces are the order of 5–10% and may be further reduced to negligible values by using antireflection coatings. By using quartz lenses with negligible transmission losses at the wavelengths of the scintillation, transmission losses through the lenses are negligible. With this optical collection efficiency and a 20% quantum efficiency of the image intensifier photocathode 43 at the peak of the sodium iodide fluorescence spectrum (4,200Å), an X-ray interaction which produces 1,100 photons in the detector crystal will produce one photoelectron at the photocathode 43 of the image intensifier. The number of electrons produced per X-ray will be greater than one for all energies above 45 keV and about 0.63 at 27.4 keV.

That the number of photoelectrons n' thus produced is of the same order as the number of X-rays (n) which interact in the detector crystal facilitates estimation of the fluctuations of n' . If n' were much smaller than n , the percentage error would be $\delta n'/n' \approx \sqrt{1/n'}$. If it were much larger than n , the statistics of n would dominate, and the percentage error would be approximately $\delta n'/n' \approx 1/n$. In the present case in which the ratio $P = n'/n \approx 1$, the errors produced in n' for an exact ratio P would be approximately equal to those which result from the fluctuations in P for a fixed (i.e., $\bar{P}\delta n \approx n\delta P$). The combined error may be estimated as $\delta n'/n' \approx \sqrt{1/n' + 1/n} = \sqrt{1/n'(P+1)}$ on the assumption that the two distributions are largely independent.

The flux of X-rays N_1 required to be incident upon a 1-mm length of crystal to yield a difference accurate to more than two standard deviations between the number of photoelectrons produced in the two adjacent resolution elements may now be estimated. The number of photoelectrons produced are $n'_e = N_1 T_e \epsilon P$ and $n'_c = N_1 T_c \epsilon P$, where T_e and T_c represent the transmission of 1 millimeter lengths of enamel and cavity, respectively, and ϵ is the fraction of the incident X-rays which interact in the detector crystal. The difference $d = \epsilon P_1 (T_e - T_c)$ and its estimated error is $\delta d = \sqrt{\epsilon P (P+1)} N_1 (T_e + T_c)$. The required two standard deviation accuracy is achieved if $d = 2\delta d$, which occurs when $N_1 = (4/\epsilon P) (P+1) (T_e + T_c) (T_e - T_c)^2$. At 27.4 keV, $\epsilon = 0.895$, $P = 0.63$, $T_e = 0.96$, $T_c = 0.45$ and $N_1 = 62$. This result is approximately 2.8 times as large as the value obtained for N_1 by Henrickson for the ideal detector plotted in FIG. 5.

There are other sources of noise; however, an increase of N_1 from 62 to 68 to maintain the two standard deviation accuracy should be adequate. The effect of this additional noise is further lessened by reducing the exposure, the exposure being much less than one second for much of the useful life of the radioactive source.

Turning now to estimating a suitable strength for an Iodine-125 source. The transmission of the cheek and 5 mm of enamel which X-rays must typically tranverse to reach the last millimeter length of a typical tooth is 8×10^{-3} . The flux per resolution element incident on the cheek is therefore N_1 divided by this transmission, or $68/(8 \times 10^{-3}) = 8430$. With an intrinsic resolution of 0.17 mm, the number of such elements per cm^2 is approximately 3,500 so that net flux of X-rays entering the cheek per cm^2 is about 2.95×10^7 . With the radioactive source at a distance of 5 cm from the specular

surface plane 51, the fractional solid angle subtended by this 1 cm^2 area is $\frac{1}{4}\pi d^2$, or $1/100\pi$. The total emission of the source into 4π steradians must then be $2.95 \times 10^7 \times 314 = 9.3 \times 10^9$ X-rays. If this is produced in 1 second, the equivalent source strength is 250 millicuries, since one curie = 3.7×10^{10} disintegrations per second. The self absorption of the 27.4 keV X-rays in the 0.5 mm diameter sources described in the above-cited Henrickson publication is of the order of 10–20% for densities of about 1 curie/ mm^2 . This is more than offset by the average of 1.39 X-rays in the 27–28 keV range produced per disintegration. Combining these two effects, a 0.5 mm diameter source containing approximately 215 millicuries of activity will provide the desired equivalent source strength and hence produce the desired imagery with a resolution of 0.17 mm.

An important feature of the invention is that this one second exposure time is much less than that required when an Iodine-125 source is used with film as the detector. The exposures expected for film vary from 15–170 seconds, with the longer times required for X-rays of thicker teeth, such as molars, in which the attenuation of the beam in the tooth itself is greater. These values were obtained by correcting those reported in the Henrickson article for various roentgenograms to a source strength of 215 mCi and a source-film distance of 5 cm. Thus, the 20-second exposure reported for panoramic roentgenograms with a 300-mCi source inside the mouth at a distance of 2–4 cm from the film predicts exposures of 40–170 seconds in the present configuration. Similarly the 10–20 second exposure range reported with a 500-mCi source at 6 cm distance predicts values of 15–30 seconds in the present example with film substituted for the crystal detector.

Iodine-125 sources can be fabricated with up to 500 millicuries in the 0.5 mm diameter package. The 0.5 mm size was selected by Beronius in the above-identified article to provide a maximum blur diameter of 0.17 mm in panoramic exposures taken with the source inside the mouth, where the distances to the film vary between 2 and 4 cm. However, the 0.5 mm diameter source is smaller than that which would produce such an image blurring at a distance of 5 cm. A larger diameter source would be tolerable if the distance is fixed at 5 cm as in the exemplary embodiment. If the maximum tooth thickness is 6 mm as assumed herein, the distance between the outside surface of the tooth and the source would be 43 mm. A 0.17 mm penumbral image of a point on the outside surface of the tooth would result from a source size of $0.17 \text{ mm} \times (43/6) = 1.2 \text{ mm}$. Under these conditions the effective area and activity of the source may be increased by a factor of 5.8 without compromising the resolution. The maximum source strength would then be 2.9 curies, and the exposure time would not be increased above 1 second until the activity fell below 215 millicuries. This would require approximately 3.75 half lives so that the useful life of the source would be approximately 220 days.

If the exposure time can be extended to 2 seconds without causing undue hardship on the exposed part, the useful life of the source would extend over more than a year. Alternately, if the cheek can be pushed aside so that the tooth can be irradiated directly, the required exposure time would decrease by more than a factor of two so that the useful life of the source would be further extended.

Now consider the reduced radiation exposure resulting from the present invention. As stated above the equivalent activity of a 27.4 keV source of X-rays for producing the desired two standard deviation difference between adjacent healthy and cavity-containing regions of a tooth in a 1 second exposure is 250 millicuries. With the source placed 5 cm from the detector crystal, it would be approximately 2.5 cm from the skin of the patient. From FIG. 14 on page 34 of the above-cited Henrickson article a 1 millicurie point source would produce a radiation field of 0.1 milliroentgens/hour at a distance of 1 meter. Correcting this value for the source strength and the 40-fold difference in source-skin distance, the source in the exemplary embodiment of this invention would produce but $0.1 \times 250 \times (40)^2/3,600 = 11$ milliroentgens/second at the skin. Since the gram absorption coefficients of air and soft tissue are essentially the same at 27.4 keV, this converts to an equivalent skin exposure of but 9.6 millirads/sec., an exposure that is more than 100 times smaller than the average mean exposure per dental film of 1,138 milliroentgens quoted in an article by Alcox entitled *Diagnostic Radiation Exposures and Dosages in Dentistry* in 76 JADA 1066 (1968) as representative of the state of the art in 1964.

Furthermore, that article explained that this exposure level only defines the properties of the radiation field and does not consider the size of the area irradiated. Another advantage of the invention is that the source is so highly collimated that only the region of cheek directly in front of the teeth being examined is irradiated. The protection of the eyes and other critical organs of the head-neck region from radiation during dental X-ray has been a subject of recent concern. A low figure for a 14-film series of eye radiation is of the order of 300-500 milliroentgens. With the shielding as disclosed in this application the exposure to the eye will be below 1 milliroentgen, hundreds of times less. And while a conventional 14-film dental X-ray survey under optimal conditions produces a dosage of between 0.5 and 1.0 milliroentgen on the gonads, the system according to the invention virtually eliminates this exposure.

Table I is a list of the most commonly used radioisotopes which emit X- or gamma-rays in the energy range of interest and their half-lives. The fourth column lists the equivalent source strength at each energy for producing the desired contrast in a system according to the invention. These values were obtained by scaling the results for Iodine-125 by the curve shown in FIG. 7 for an ideal detector and correcting for the detector-produced increase in the fluctuations as described above. Table I follows this page.

For energies below 20 keV the absorption of radiation in the tooth itself is so large that inordinately high activities are required to provide the required number of X-ray quanta at the detector crystal. For energies above 100 keV the difference in absorption between healthy and diseased tooth structures is so small that a very large flux is required to produce statistically significant contrast between them. In addition the effective utilization of these more energetic emissions is greatly reduced because the conversion efficiency of the detector crystal falls off rapidly with increasing energy.

A preferred balance between the net transmission of the tooth and the detection efficiency comes in the 35-50 keV range. Column 5 of Table I lists the approxi-

mate percentage of the nuclear disintegrations which produce the X-ray of interest in each case. It is seen that Iodine-125 has the especially advantageous property of providing more than one 27.5 keV X-ray per disintegration. The alternate sources may be used less satisfactorily for dental X-ray or with acceptable and perhaps superior performance for certain other applications.

The required isotope activity may be determined by dividing the desired equivalent source strength by the radiative efficiency to produce the results tabulated in the penultimate column of Table I. These results may be compared with the last column listing the source activities which are presently available commercially at reasonable cost. The comparison shows that Iodine-125 is presently available at a source activity and size compatible with dental X-ray in accordance with this aspect of the invention.

Other suitable materials include Thulium-170 and Americium-241 having an extremely long half life. As supplies of these materials become more available and the size of the source package reduced, these materials might be advantageous when used in the dental X-ray aspects of the invention.

Referring to FIG. 11, there is shown another embodiment of the invention for culling parcels. A gamma ray source assembly 61 scans a target parcel 62 to produce an image of the contents along a line of detector crystals 63 backed by an array of photomultipliers 64 to produce detected signals analyzed by detector electronics 65, typically for producing an alarm signal when the parcel contains a gun-like or bomb-like object. While not shown in FIG. 11, parcel 62 may be upon a conveyor that carries parcels across the culling station automatically.

It need hardly be doubted that parcels, baggage and people carrying guns, bombs or other contraband material create great dangers. And detection of such contraband material is difficult, costly and often embarrassing when reputable travelers are subjected to the indignities of a search. The present invention is capable of detecting firearms unambiguously without opening parcels or subjecting them to abnormal physical handling.

An important feature of a preferred system of FIG. 11 is its full compatibility with automated parcel handling systems. The invention relies upon distinguishing between the gamma-ray absorption properties of different materials to detect suspicious parcels. The imaging system uses gamma-rays and electro-optical imaging techniques to identify contraband within the culled parcels. The radiation doses produced by the system present no hazard to personnel, and the parcels receive negligible doses. The system may use electronic techniques well within the present state-of-the-art with an assembly of commercially available components.

An inspection system according to the invention contemplates a two stage process in which the first stage is operated "on-line" to make a coarse determination of the amount of metal contained within a parcel (or person in a personal examining system). This first stage does not require interruption of the transportation system of parcels or humans at all so long as the targets examined do not have a suspicious amount of metal.

Targets having a suspicious amount of metal move to a second stage which displays an image of the target contents. For many applications, the image may be

manually examined to determine whether the target package should be opened or the target person searched. Where many fine examinations must be made, it may be advantageous to have the second stage include pattern recognition apparatus for automatically determining whether the target contains contraband material.

In the first stage to be described in greater detail below, all parcels are examined for dense material (especially metal) content, using relatively soft gamma-ray radiation of very low intensity with electronic logical apparatus culling suspicious targets. These suspicious targets then go to a second station, where the target contents are imaged upon a visual display for identification of contraband by an operator. The "innocent" targets may then return to the mainstream while "illegal" targets are designated for other disposition.

The second station uses soft, low-intensity gamma-rays for obtaining an image of the contents. Radiation exposure is kept to only that absolutely necessary to present the information required for the particular examination. And radiation exposure to the operator does not exceed the dose he receives daily from his natural environment. A feature of the invention is that most radiation-sensitive products that may be shipped, such as very fast photographic film, are essentially unaffected by the radiation level required.

A complete parcel inspection system according to the invention may comprise several metal detector culling stations operating in parallel in the main parcel stream before the sorting stations. A conveyor switch may be arranged to immediately follow each detector section for actuation upon suspicious metal detection to divert the suspect parcel from the main parcel stream. A crossing conveyor may collect all suspect parcels and transport them to a single imaging station for detailed examination. Parcels passing the image inspection may then return to the main parcel stream for further processing with the other "legal" parcels while the illegal parcels may be switched to a storage area for further disposition.

A feature of the invention is that the measurements obtained along with suitable data processing reduces the false alarm rate, thereby reducing the handling rate of the full imaging station.

Another feature of the gamma-ray detecting approach is insensitivity to magnetic effects. In contrast a magnetic detector operates in an uncertain magnetic environment. A sensitive magnetic detector must be located with careful attention to proximity of electrical wiring and switch gear, characteristics of conveyor systems, proximity of machine and other local ferrous metal traffic, such as carts and portable cleaning equipment. And weapons constructed of nonmagnetic materials would not be detected by magnetic detection methods.

The culling station of FIG. 11 is compatible with conveyor speeds currently in use in automatic and semiautomatic parcel handling facilities, these speeds typically being 200 feet/minute.

The gamma ray source assembly 61 may contain cobalt 57 radio isotope as the radioactive source placed on one side of the conveyor belt. Detector bank 63 may comprise a vertical array of 1" x 1" gamma-ray sodium iodide detector crystals, abutting one another. The source radiation beam is preferably shaped to produce radiation fans that cross above the belt, beginning

at the point source and terminating in the line of detectors 63.

There are preferably at least two fan beams as shown to assure complete parcel coverage. Parcels passing through the two scanning beams modulate the transmission to individual detectors in accordance with the absorbing properties of the parcel and its contents. A suitable criterion for defining a "suspicious" parcel is to require counting rates in one or more detectors to fall to less than a predetermined count. A suitable minimum sized object of interest may be a cubic inch.

Referring to FIG. 12, there is shown a combined block-pictorial diagram of a preferred form of culling station for providing a rough indication of suspect parcels. The scan beam control 71 scans a parcel to produce a rough image of the transmissivity of the parcel upon the detector crystals 72 each backed by an associated one of photomultiplier tube assembly 73 to provide signals that are processed and produce an alarm from alarm 74 when the indicated metal content is greater than a predetermined value.

An 80 Hz oscillator 75 drives a motor 76 that rotates upper and lower radioisotope sources 77 and 78, respectively, having their apertures relatively displaced by 180° for illuminating the parcel through upper and lower windows 81 and 82, respectively, in shielding case 83. A rotation sensor 84 provides a signal to the signal processing control unit 85 to synchronize the data processing circuitry with the alternating scan beams.

The detection system comprises a number of detector crystals 72, the ones designated 1 through 14, for receiving a rough image and a reference detector crystal, designated R, for receiving reference radiation. Each of these detector crystals is backed by the photocathode of an associated photomultiplier tube and responds to incident radiation by providing light on the associated photocathode. A radiation conversion produces an electron emission that is amplified within the associated photomultiplier tube. The output of each photomultiplier tube is a series of electrical signal pulses occurring at a rate related to the parcel material attenuation then in the field of view of the detectors.

A light source 86 normally illuminates a photocell 87 that is deenergized when a parcel passes between the light source 86 and the photocell 87 to provide a signal delivered to signal processing control unit 85 indicating that a parcel is present.

The signal processing apparatus comprises an associated one of counters 91 and digital-to-analog converters 92. The channels associated with detector crystals 1-14 also each include a sample and hold circuit and output switch 93 for sequentially delivering held potentials corresponding to the count in the associated counter to the signal input of comparator 94. The reference input of comparator 94 receives a reference potential from threshold control 95 that receives the combination of a manually set potential set by control 96 and a reference potential from the reference detector channel representative of the level of radiation then being provided from the scanning source. Signal processing control 85 provides appropriate signals for resetting the counters, the sample and hold circuit, the output switches and activating the threshold control 95 to provide a reference level.

The signal processing circuitry functionally counts pulses, converts the digital count into an analog level

and compares to establish whether or not a detected pulse rate is to be accepted as an alarm. In the illustrated embodiment, the detector bank generates 14 parallel lines of variable pulse rate information. At a parcel passage rate of 200 feet/minute a complete signal processing cycle of 25 milliseconds is adequate.

The cycle is preferably divided into two parts to permit count totalizing with respect to each scanned beam. By accumulating counts for 12.5 millisecond intervals, each count is representative of the parcel having traveled only 0.5 inch. Thus, in 1 inch of travel the parcel is examined from two different aspects with the detector crystals effectively viewing the parcel through a 1 inch vertical slice that is segmented by the stacking of 14 individual inch-square crystal-windowed PMP assemblies. Thus, a high mass object as small as an inch square will effectively shadow at least one detector to make the detector count so low that an alarm condition is signalled.

Having described the logical arrangement of the system of FIG. 12 and the general functioning of the system, its operation will be described in detail. As motor 76 rotates at constant speed, the beam from source 77 exits through aperture 81 to illuminate detector crystals 72. Sensor 84 provides a signal on input A of signal processing control unit 85 signifying the start of the scan to produce a resetting signal that resets the counters 91 to 0, thereby starting a new counter interval. Counters 91 then receive pulses from associated PMT assemblies 73 and present the advancing count to associated ones of digital-to-analog converters 92. These converters provide an analog signal proportional to the binary count. At the end of half the scan period (typically 12.5 milliseconds) each of sample and hold circuits 93 receives a signal from output D of signal processing control unit 85 that holds the potential thereon to a value representative of the total count accumulated in a counter during the count interval.

As the scanning beam from source 77 is shielded, rotation sensor 84 provides another signal to the A input of signal processing control unit 85 that closes the switches in the sample and hold and switch circuits 93 in sequence to compare the stored voltages in the differential amplifier comparator 94 to provide an alarm when any of the stored potentials is less than a predetermined reference potential provided by threshold control 95. Alarm 74 may provide a signal that results in automatic parcel switchout at the imaging station.

A feature of the invention is "self-check" cycling to introduce an appropriate threshold level under the control of the signal from output line F of signal processing control unit 85 when no parcel is present. When light source 86 illuminates photocell 87 to indicate that no parcel is present on input line B of signal processing unit 85, channels 1-14 continuously respond to radiation. If a predetermined rate is not then produced as determined from comparing the sampled voltages with a reference potential determined by the output signal on output line F of signal processing control unit 85, comparator 94 produces an alarm signal to indicate a malfunction. The time at which such an alarm signal occurred would be indicative of which of the 14 channels was defective.

Another feature of the system is the provision of a reference detector channel to permit object signal count comparison against a floating reference. Half life decay of the radiation source results in general count

level decrease. If this change went uncompensated, there would be an unnoticed change in the parcel reject or accept count limit causing a gradual increase in false alarm rate. The reference detector is positioned so that it directly views the source and continuously provides a signal proportional to source behavior as a floating reference at the comparator. The reject-pass ratio may be adjusted manually by the control 96 shown.

The frequency of oscillator 75, nominally indicated as being at 80 cycles, may be adjusted to control the beam scan to compensate for various conveyor system velocities.

Referring to FIG. 13, there is shown a combined block-pictorial diagram of an imaging station according to the invention for fine examination of parcels rejected in the coarse examination effected with the system of FIG. 12. A gamma ray source 101 illuminates parcel 102 to produce a radiographic image on scintillator detector 103 comprising sodium iodide crystals to produce a visible image of the contents of parcel 102 that is focused by optical system 104 on image intensifier 105 after which image detector 106, typically a vidicon camera, detects the intensified image and delivers it to storage tube 107 through image control unit 108. Image display unit 111, typically a television monitor, receives the stored image from storage tube 107 through storage control unit 112. An intensifier control unit 113 controls image intensifier 105.

The entire display is under the control of system control unit 114 which delivers appropriate control signals to intensifier control unit 113, image control unit 108 and storage control unit 112. Such television display systems using storage tubes are well known in the art and not described in detail here so as to avoid obscuring the principles of the invention.

Having described generally the physical arrangement of the imaging station according to the invention, certain principles of operation will be described. The gamma ray source 101 may be similar to that used at the culling station, typically comprising cobalt-57 radioisotope. The source 101 provides a conical beam of gamma-rays that penetrate portions of the parcel in varying degrees depending on the shape and mass of the internal material. Rays which penetrate the package strike the scintillator detector 103 to produce visible light by photoelectric and Compton processes.

The scintillator detector 103 is preferably a mosaic assembly of small crystal planes of sodium iodide that produces a visible light picture representing the contents of the parcel. The electrooptical system amplifies this visible light picture so that it may be easily observed visually.

Preferably the image detector 106 operates on a single frame basis similar to taking a snapshot with a camera. A gamma-ray parcel exposure occurs while the electron scan beam of the image detector 106 is cut off sufficiently long to build up a charge storage representing an image of good contrast and sufficient intensity for display purposes. A satisfactory charge build-up or integration time interval is approximately 1 second. Then the image detector 106 scans the target image charge with its electron scanning beam to produce a composite electrical video signal such as produced in an ordinary television system.

Although this video signal could be displayed as a visible image directly on a standard television monitor picture tube, the single-frame raster-scanned image

readout event is ordinarily too complex for operator comprehension with one frame in 1/30 second for standard television. Therefore, an image storage tube electronically holds the image for the length of time desired for repetitive scanning and subsequent display and evaluation on the television monitor. Suitable operator controls permit image erase at will to allow new exposures to be made at various parcel positions.

The system described above minimizes radiation exposure requirements for successful parcel content imaging, operates reliably and is relatively easy to assemble.

The resolution requirement for parcel contents imaging is not severe so that many types of optical systems may be employed, such as diffracting, reflecting, Fresnel or combinations of these elements. The specific choice of light transmitting characteristics of the optical system affects the source strength required and may be selected to maximize cost effectiveness.

A suitable source strength of 2.4 curie of Co-57 is adequate, although other sources may be used. For example, cadmium 109 may be used, though with less conversion efficiency. A feature of the invention is the low radiation from the system. The radiation near the operator at 1 meter distance in an 8 hour working day with no shielding is of the order of 0.1 milliroentgenens.

The daily dose acquired by a person from natural causes (cosmic rays and earth radioactivity) is larger and about one mr. The permissible AEC daily dose is 50 mr.

Each millimeter of lead shielding can further reduce the dose by a factor of 30, thus reducing the radiation received by the operator from direct parcel probing to insignificant levels.

Furthermore, the dose to the contents of a parcel is of the order of several magnitudes below that required to change the density of very fast film by 0.1 density units, an effect that is directly unobservable. Such low levels of radiation make the system practical for examination of a person.

Identification of contraband within a parcel or carried by a person may be made by a human observer interpreting a television monitor image. It is within the principles of the invention to incorporate automatic identification through the use of pattern recognition techniques.

There has been described novel radiographic imaging techniques and apparatus characterized by numerous advantages. The system is compact. Radiation doses are low. The invention is useful in solving many problems economically and safely.

It is evident that those skilled in the art may now make numerous uses and modifications of and departures from the specific embodiments described herein without departing from the inventive concepts. Consequently, the invention is to be construed as embracing each and every novel feature and novel combination of features present in or possessed by the apparatus and techniques herein disclosed and limited solely by the spirit and scope of the appended claims.

What is claimed is:

1. Radiographic apparatus comprising, a source of radiographic energy,

means for directing said radiographic energy toward an object,

means including a thin crystal having a flat specular surface facing said source that is a compound from the group consisting of sodium iodide and cesium iodide for converting at least the radiographic energy passing through said object into a visible image on the crystal representative of the transmission characteristics of the object after the radiographic energy from the source passes through an object,

and lens means positioned with its object focal plane including said flat specular surface for producing an image of said visible image,

said lens means including optical means for viewing the image on said crystal accepting rays within a predetermined viewing cone,

said apparatus characterized by a predetermined resolution W,

the thickness of said crystal being substantially equal to the product of said resolution W with the effective *f* number of said viewing cone in said crystal.

2. Radiographic apparatus in accordance with claim 1 wherein said thickness is less than a centimeter.

3. Radiographic apparatus in accordance with claim 1 wherein said thickness is of the order of a millimeter.

4. Radiographic apparatus in accordance with claim 1 and further comprising,

a closed circuit television system for amplifying and displaying the visible image on said crystal upon the face of a display tube for real time observation.

5. Radiographic apparatus in accordance with claim 1 and further comprising,

an image intensifier for amplifying the visible image upon said crystal,

and film means for recording the intensified image.

6. Radiographic apparatus in accordance with claim 1 wherein said source of radiographic energy comprises a low dosage isotope.

7. Radiographic apparatus in accordance with claim 6 wherein said low dosage isotope comprises an isotope from group consisting of radioactive Iodine-125 and Cobalt-57.

8. Radiographic apparatus in accordance with claim 6 and further comprising,

an assembly supporting said source and said crystal to define a region therebetween for accommodating a tooth to be X-rayed,

said assembly including optical means for focusing the visible image on said crystal upon a predetermined visible image plane.

9. Radiographic apparatus in accordance with claim 8 and further comprising image intensifying means located at said image plane.

10. Radiographic apparatus in accordance with claim 9 wherein said optical means comprises a generally J-shaped light pipe with said crystal at the end of the short leg of the J and the image intensifying means at the end of the long leg of the J with said source supported from said long leg.

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