Radiation Safety Review for 511-keV Emitters in Nuclear Medicine

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With the advent of high-energy collimators and dual-head coincidence cameras, standard nuclear medicine facilities will soon begin imaging with PET isotopes. The use of 511-keV emitters raises new radiation safety concerns for technologists traditionally limited to handling ^{99m}Tc and other low-energy isotopes. This article is a basic review of positron emitters, measurement concerns, exposure rates, shielding requirements and external radiation exposure mitigation. Newly developed PET shielding products are presented and regulatory status is discussed briefly.

Key Words: radiation safety; PET isotopes; positron shielding

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Investigation into the use of 511-keV emitters for medical imaging, commonly known as positron emission tomography (PET), began over 20 yr ago with the development of the first positron scanner by Massachusetts General Hospital in conjunction with scientists at Massachusetts Institute of Technology. However, the size, expense and complexity of the cyclotrons then available limited the use of PET to large research institutes. With the development of the new breed of self-shielding cyclotrons in the 1980s, and significant improvements in computer hardware and software for image processing and reconstruction, PET facilities began to move into larger university hospital centers or government hospital facilities and expanded beyond research to a clinically useful tool. PET provides a unique image of the metabolic and physiological process which cannot be duplicated by radiographs, CT, ultrasound or MRI. This is accomplished by tagging organic compounds with short-lived positron emitters. The positron-emitting isotopes and some common clinical applications are listed in Table 1 (*1*).

Positively charged electrons (positrons) are emitted from the nucleus as it undergoes β^+ decay. The positron travels a short distance (a few millimeters), depositing any excess energy before it combines with a free electron. The mass of the e^- and e^+ is completely converted into two photons with an energy of 511 keV each. These annihilation photons are emitted in opposite directions, 180° apart. This is the basis for PET scanners in a process called coincidence counting. PET scanners use detectors in coincidence, so that both annihilation photons generated by a positron can be detected at the same time. Only those photons detected within a predefined coincidence window are recorded. This process is known as coincidence counting. Single-photon events, such as scatter, background and random photons (only one photon is detected), are events that are not recorded. The use of coincidence detection allows for more accurate determination of the angle of interaction of the annihilation photons without the use of a collimator. This greatly improves the sensitivity of PET over gamma camera systems.

Within the last few years, gamma camera manufacturers have begun to produce high-energy collimators that can be used with 511-keV emitters on standard, single-head gamma cameras. These high-energy collimators generally have greater septal thickness, fewer holes and larger diameter holes than low- or medium-energy collimators to reduce scattered radiation and those 511-keV photons not oriented perpendicular to the detector face while still maintaining an acceptable sensitivity. For example, a typical 511-keV collimator may have a thickness of 65 mm, a hole diameter of 3.4 mm and a septal thickness of 3 mm.

The U.S. Food and Drug Administration (FDA), which is responsible for controlling the approval of medical devices in the marketplace, had taken the position that the 511-keV collimator was not necessary and has been slow to approve these new medical devices, or 510(k), submissions. In April 1996, the FDA publicly reversed its position on this issue and stated that the pending 510(k) submissions would be processed collectively. With the new 511-keV collimators, one PET production center can supply longer-lived positron-labeled radiopharmaceuticals (e.g., ^{18}F products) to numerous nearby nuclear medicine facilities, thereby dramatically increasing the availability of selected PET procedures and greatly reducing the unit cost of these procedures.

Camera manufacturers are also developing coincidence counting gamma cameras. These units consist of a dual-head gamma camera whose detectors are fixed at 180° from one another. Following the same principal as the PET scanners,

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TABLE 1
Current PET Applications

Isotope	Form	Application
^{15}O	Water	Cerebral blood flow
^{13}N	Ammonia	Myocardial blood flow
^{11}C	Glucose	Cerebral glucose metabolism
^{11}C	Acetate, palmitate	Myocardial metabolism
^{18}F	Fluorodeoxyglucose	Cerebral metabolism, myocardial metabolism, tumor imaging
^{82}Rb	Rubidium	Myocardial blood flow

TABLE 2
Gamma-Ray Constants

Isotope	Γ (mSv-m²/h)/MBq
^{99m}Tc	3.317E-5
^{201}Tl	2.372E-5
^{131}I	7.647E-5
^{11}C	19.37E-5
^{18}F	18.79E-5

only those 511-keV photons that are simultaneously detected at 180° will be recorded; all other photons will be disregarded. Coincidence cameras offer several distinct advantages when compared to single-head collimated cameras. By removing the collimator, the sensitivity is greatly increased, which reduces the amount of activity needed for good imaging. Also, the intrinsic resolution of the coincidence camera increases with increasing energy, while the extrinsic resolution of the gamma camera using positron emitters and the 511-keV collimator is greatly degraded. Fewer projection angles are required for an image as compared to a collimated camera because the coincidence camera has, in effect, a fan-beam geometry. Among the drawbacks to the coincidence camera is the limited amount of activity that can be in the field of view before the camera saturates. Both systems suffer from the limited detection efficiency of NaI(Tl) for 511-keV photons.

The coincidence cameras may also be used for any standard nuclear medicine imaging and, most importantly, are less than half the cost of a dedicated PET scanner: $600,000 compared to $1.2–$2 million. Coincidence cameras for most manufacturers are still in the beta-testing phase, and commercial distribution is still pending FDA approval. In response to this advance in imaging technology, PET cyclotron sites are joining with radiopharmaceutical companies and developing distribution networks that will provide transportation of ^{18}F compounds, both multidose vials or unit dose, to local nuclear medicine facilities.

The availability of 511-keV collimators, coincidence cameras and PET distribution networks will result in ^{18}F at a reasonable cost to any facility located within geographic proximity to a cyclotron center. The transfer of high-energy isotopes for use in routine nuclear medicine departments raises concern for radiation safety, primarily because shielding devices and safety procedures have historically been developed for use with ^{99m}Tc, which has a photon energy of 140 keV. Positron-emitting isotopes also generate radiation safety concerns.

MEASUREMENT OF POSITRON-EMITTING ISOTOPES

Whether positron-emitting isotopes are used in a dedicated PET center or transferred to a traditional nuclear medicine facility, the activity measurement process remains the same. The annihilation photons are sufficiently high in energy to be readily measured by a standard dose calibrator. No geometry corrections are required for container differences between the primary standard used to generate the calibration number and the radiopharmaceutical vial or syringe used for processing and distribution. Calibration numbers for all PET isotopes should be provided by the dose calibrator manufacturer.

Although a dose calibrator cannot differentiate between photons of different energies, its response does vary since a different amount of current is produced for each energy. The higher the energy, the greater the current. Maximum activity limits are generally specified in terms of ^{99m}Tc. Since photons from PET isotopes are significantly more energetic than the 140-keV photons produced by ^{99m}Tc, the current produced for ^{18}F is about three times greater than ^{99m}Tc for the same activity. Subsequently, maximum activity limits are about one-third less for ^{18}F than for ^{99m}Tc. For this reason, a dose calibrator that has higher activity limits is preferable in PET production centers. For unit doses or multidose vials in the millicurie range, any currently available model is suitable.

In addition to measurement capabilities, the gamma-ray shielding around the dose calibrator must be evaluated. Dose calibrators designed for nuclear medicine applications generally provide 0.3–0.6 cm (⅛–¼ in.) of inherent lead shielding around the ionization chamber. This is insufficient protection for the technologist when measuring PET isotopes. Lead rings specifically designed to surround a dose calibrator are commercially available between 40–60 mm thickness. The combined lead thickness of the shield and external lead rings should be at least 5 cm or greater. Alternately, standard lead bricks can be purchased and stacked in front of the calibrator as a shield. Care should be taken to ensure that the bricks are securely fastened and do not accidentally fall, causing injury the technologist or damaging the calibrator.

EXPOSURE FROM POSITRON-EMITTING ISOTOPES

A comparison of gamma-ray dose constants (Table 2) (*2*) and corresponding dose or exposure rates for the same activity of typical nuclear medicine isotopes and positron emitters clearly demonstrates the safety concerns between the isotopes. The gamma-ray dose constant, Γ, is the dose rate in air for 1 MBq of an isotope at a distance of 1 m.

Many of the positron emitters commonly used in PET have only the 511-keV annihilation photons. PET isotopes are specifically selected from isotopes that do not have significant

TABLE 3
Comparison of Dose Rate and Exposure Rate at 5 and 20 Centimeters

	20 cm		5 cm	
Isotope	Dose rate (mSv/hr)	Exposure rate (mR/hr)	Dose rate (mSv/hr)	Exposure rate (mR/hr)
^{99m}Tc	0.31	35	4.96	568
^{18}F	1.74	199	27.8	3189

Assumes 1 Roentgen = .00873 J/kg in air.

gamma emissions other than the 511-keV photons. To calculate exposure rates, the following formula (*3*) is used and assumes a point source to simplify comparison:

$$X = \frac{\Gamma A}{d^2}.$$

where X = exposure rate; Γ = gamma-ray dose constant in (mSv−m^2/h)/MBq; A = activity in MBq; and d = distance from the source in centimeters.

The typical ^{18}F unit dose activity is about 370 MBq (10 mCi). Applying this formula to a point source with 370 MBq (10 mCi) of activity at a distance of 20 cm (8 in.) results in dose and exposure rates that are correspondingly six times greater for the positron emitters compared to low-energy nuclear medicine isotopes as shown in Table 3.

Twenty centimeters is about the distance of extremity exposure using typical short-handled tongs. To estimate the exposure near the surface of an unshielded vial, reducing the distance to about 5 cm (2 in.) results in the dose and exposure rates as shown in Table 3.

RADIATION EXPOSURE REDUCTION

The traditional methods of time, distance and shielding are the first step in reducing radiation exposure. Decreasing the amount of time an exposure occurs will decrease the exposure in a linear fashion (*3*):

$$\text{mR/hr} \times \text{hr} = \text{total mR}.$$

All technologists should practice transfering, drawing and injecting procedures using a nonradioactive solution before using positron-emitting isotopes. You should gain experience using the shielding products for high-energy isotopes that are heavier, bulkier and more awkward to handle. Evaluate current dose handling protocols and ask the following questions: Is there a way to decrease the exposure time? Is either a vial or loaded syringe unshielded for any length of time? Is there a way to improve the efficiency of current procedures? Increased awareness of the radiation risks and a careful review of established methods can often reduce handling time, thus reducing radiation exposure and dose. The greatest source of radiation exposure to the technologist is prolonged contact with the patient. Technologists who work in PET centers realize that it is necessary to minimize the time spent in close contact with the patient.

Distance is one of the most important methods of radiation protection. Increasing the distance between the source and the technologist will decrease the exposure rate by the square of the distance (assuming a point source) (*3*):

$$\frac{X_1}{X_2} = \frac{d_1^2}{d_2^2} \quad \text{or} \quad X_1 d_2^2 = X_2 d_1^2$$

where X = exposure rate and d = distance from the source.

If the exposure rate for the same activity at the same distance is six times greater for 511-keV emitters than for ^{99m}Tc, then increasing the distance by the square root of 6, or 2.4 times, will reduce the exposure rate to that of ^{99m}Tc. However, this is not always easy to accomplish. Typical remote handling tongs in nuclear medicine are about 23 cm (9 in.) long. Tongs with longer handles are readily available for vial handling or vial transfers. Vials can be comfortably moved from one storage container or transporter to another with tongs as long as 61 cm (24 in.). Assuming 8 cm to hold the tongs, this increases the distance from 15 to 53 cm from source to fingers. By increasing the distance 3.5 times, the exposure rate decreases by a factor of 12. Although increasing distance is possible when handling vials, it is not as easy to increase the distance between the source and the technologist when drawing a dose from a vial into a syringe, or administering a dose from a syringe into a patient without the addition of specialized equipment to provide added shielding.

Decreasing time and increasing distance alone cannot adequately reduce radiation exposure from positron emitters for all handling procedures. Additional shielding is mandatory. Dedicated PET isotope production centers have addressed the shielding and handling requirements for 511-keV emitters in several ways. The central processing control units (CPCUs) are generally housed in cellular mini-sized hot cells shielded with 60 mm of lead. The final elution product is stored in thick vial shields and transferred to larger hot cells with remote handling manipulators or shielded hoods with 60-mm-thick protective walls or lead barrier shields. Once the unit doses have been prepared, the syringe is transported quickly to the scanning room through a variety of syringe shields and transporters specifically designed for high-energy isotopes. A short-term intravenous line should be inserted so doses may be injected while shielded. The entire PET laboratory is designed for the exclusive use and handling of high-activity, high-energy isotopes.

The style and thickness of shielded products found in nuclear medicine facilities are designed for protection against 140-keV photons generated by ^{99m}Tc. For example, a typical syringe shield consists of 0.32-cm (⅛-in.)-thick lead or equivalent. Vial shields are usually 0.64–1.3-cm (¼–½-in.)-thick lead or equivalent. Lead barrier shields are usually manufactured with 1.3 cm (½ in.) lead and 0.64-cm (¼-in.)-thick leaded glass. This thickness provides adequate protection for low-energy isotopes but is inadequate in shielding 511-keV emitters.

TABLE 4
Comparison of Linear Attenuation Coefficients for Lead

Isotope	μ (cm^{-1}) for lead
^{99m}Tc	30.146
^{201}Tl	26.211
^{131}I	2.409
^{11}C	1.695
^{18}F	1.827

A comparison of linear attenuation coefficients for lead (Table 4) (*2*) and corresponding shielding requirements for the same activity of typical nuclear medicine isotopes and positron emitters clearly demonstrates the shielding differences between the isotope groups. The linear attenuation coefficient, μ, is defined as the fraction of photons removed from the radiation field per centimeter of absorber. Consequently, the value for μ changes with different materials and different photon energies.

Note that the linear attenuation coefficient for the positron-emitting isotopes is much smaller than that of ^{99m}Tc. To calculate dose or exposure rates, the following formula (*3*) is used and assumes a point source to simplify comparison:

$$X = BX_i e^{-\mu t}$$

where X = exposure, shielding present; B = buildup factor; X_i = initial dose or exposure rate, no shielding; μ = linear attenuation coefficient; and t = thickness of lead.

The buildup factor is an experimentally determined correction factor that accounts for the fact that the measured exposure value is greater than predicted by the linear attenuation coefficient alone. As the photons traverse the material, some are scattered along the path generating energetic electrons that contribute to dose buildup. Tables of buildup factors can be found in most health physics references (*4*). Table 5 compares the difference in shielding required to reduce an activity of 370 MBq (10 mCi) at 20 cm to the same dose rate after shielding. To provide the same amount of attenuation for ^{18}F as is currently provided by ^{99m}Tc, the thickness of lead must increase by a factor of 16.

An alternate way to compare the shielding requirements for ^{99m}Tc and ^{18}F is to compare the HVL (half-value layer). The HVL is that thickness of material that will decrease the amount of exposure by one-half. The HVL for 140-keV photons in lead is 0.026 cm (0.01 in.), as compared to 0.402 cm (0.16 in.) for 511-keV photons. As expected, the HVL for positron emitters is about 16 times greater than for ^{99m}Tc.

TABLE 5
Comparison of Shielding Requirements

Isotope	Dose rate mSv/hr	Lead thickness	Dose rate after shielding
^{99m}Tc	0.31	0.159 cm (1/16 in)	0.04
^{18}F	1.74	2.54 cm (1 in)	0.04

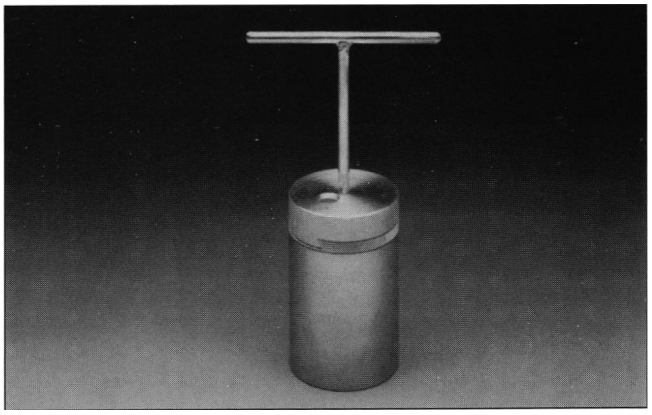

FIGURE 1. PET tungsten vial shield.

PET SHIELDING PRODUCTS

Attenuation of 511-keV photons in lead is equally dependent on the atomic number (Z) and electron density because the attenuation due to the photoelectric effect is about equal to that due to Compton scattering (*5*). Increasing the thickness of a lead shield by 16 times is impractical due to weight or size restrictions, therefore, tungsten, which has higher Z and electron density compared to lead, is a suitable alternative. A comparison of the HVL values for lead and tungsten indicates the relative thickness required to establish equivalent shielding protection. Again, the influence of the buildup factor changes the values at different energies from that predicted by the linear attenuation exponential component alone. At 511 keV, the HVL for lead is about 0.4 cm (0.16 in.) and 0.28 cm (0.109 in.) for tungsten. Tungsten provides about 1.4 times the shielding capabilities for the same thickness as lead.

For positron shielding of cylindrical shapes for which weight is an important consideration, such as syringe shields and vial shields, tungsten is the material of choice. The decreased thickness required reduces the outer diameter and, consequently, the total weight significantly. Some shielded containers allow both dose drawing and dose administration without excessive exposure to the technologist. High-energy tungsten versions of syringe shields, vial shields and dose drawing syringe shields are commercially available. Leaded "PET thick" lead barrier shields and a dose drawing station, which holds a shielded vial and rotates, facilitate use. The only current commercial manufacturer of PET shielded products is Capintec, Inc. (Pittsburgh, PA).

The vial can be transported in a shielded 1-in. lead-equivalent tungsten vial shield (Fig. 1). The shielded vial is then transferred into the dose drawing station. (Fig. 2). With the aid of the dose drawing syringe shield (Fig. 3), the dose can be safely withdrawn from the vial. A complete unit, including the lead-block shield, is shown in Figure 4. To transport and inject the dose, remote injection devices are available. With the addition of a few specially designed items, doses can be safely

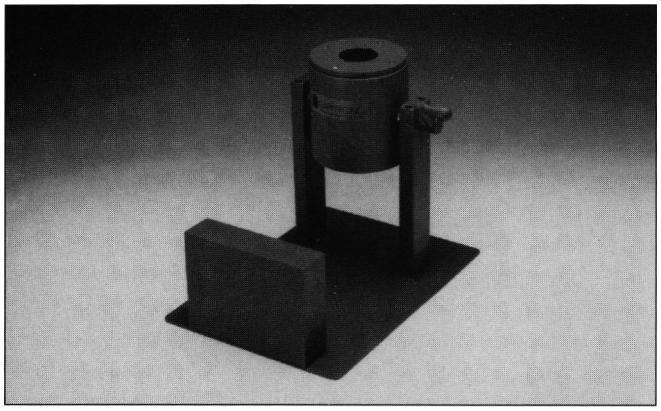

FIGURE 2. Dose drawing station.

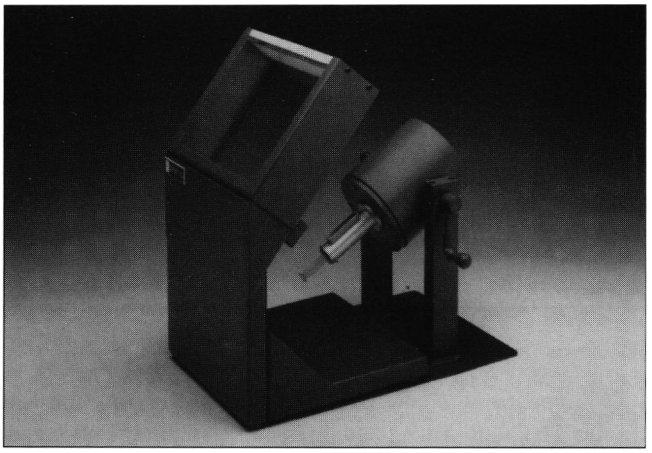

FIGURE 4. Dose drawing station with lead block shield.

transported, drawn and injected without undue exposure to the technologist.

OTHER RADIATION SAFETY ISSUES

To evaluate other radiation safety issues related to the use of positron emitters in PET applications, including dosed patients, spills and waste disposal, the half-lives of the isotopes, as shown in Table 6 (*5*), should be used as a guideline. Clearly, the short half-lives of ^{15}O, ^{13}N and ^{11}C will preclude their use in any facility that is not directly associated with a cyclotron. However, ^{18}F is sufficiently long-lived to allow transport to nuclear medicine facilities within close proximity. The short half-life also provides a tremendous advantage in dealing with postadministration exposure control.

Waste disposal includes contaminated syringes, gloves and other protective clothing and absorbent pads for work areas. All ^{18}F waste can be easily accommodated by decay in storage for 1 day (10 half-lives = 18 hr). Waste from ^{18}F can also be combined with other radiopharmaceutical waste, which includes isotopes with short half-lives, such as ^{99m}Tc but excluding ^{131}I. The waste should then be surveyed with a standard survey meter to ensure that there is no measurable exposure above background. Once radiation hazards have been eliminated, the waste must then be categorized as to whether or not it is biohazardous. Waste that is not biohazardous can be transferred as ordinary cold, or nonradioactive, waste and handled accordingly. Items that are considered biohazardous, once transferred to nonradioactive status, must be handled in accordance with the local facility safety program, applicable OSHA and state health regulations.

The short half-life of ^{18}F provides easy control in accidental spill situations. Since most spills will be either dropped vials or syringes, the volume of liquid involved is fairly small: 10–30 ml. Fluorine-18-FDG is excreted through the kidneys. Patients who are incontinent or catheterized can be a problem with regard to contamination from radioactive urine. In addition, some oncologic procedures require the catheterization of patients for emptying the bladder. Technologists must be just as diligent as they are with patients receiving bone imaging. Evacuating the immediate area for the remainder of the day would provide the easiest way to prevent any unwanted exposure from an accidental spill. In some instances, where evacuation of a room for a prolonged period would prove too costly in lost services or procedures, a combined approach of evacuation for a short period of time (1 or 2 hr) and standard spill clean-up procedures would be effective. After the activity has decayed somewhat, use remote handling devices (long-handled tongs) to manipulate the absorbent material used to soak up the spill and transfer the contaminated materials to a shielded waste container. Survey the contaminated area and any potentially contaminated personnel. If residual activity remains, clean again with a disposable dampened cloth. Survey the area again and repeat as necessary. Once the spill area has been cleaned, remove protective clothing and survey any personnel who may have been contaminated. This approach would permit use of a room within a few hours after a spill, if economically necessary.

FIGURE 3. PET drawing syringe shield.

TABLE 6
Comparison of Half-Lives

Isotope	Half-life
^{15}O	122.2 sec
^{13}N	9.97 min
^{11}C	20.4 min
^{18}F	109.7 min
^{99m}Tc	360.4 min

Exposure can also occur from patients. Technologists should always use the method of distance when possible to reduce their dose from the patient. Pregnant technologists should consult with their radiation safety officers and be monitored to remain within the limits in 10CFR Part 20. In instances of unusual radiation concern (e.g., pregnant spouse, nursing mother), apppropriate counseling is recommended.

REGULATORY ISSUES

Positron-emitting isotopes are produced by cyclotron accelerators and, as such, come under the jurisdiction of individual states rather than the NRC. The Conference of Radiation Control Program Directors, Inc. (CRCPD) technical advisory group for the Healing Arts (Task Group SR-6) met five years ago to discuss the need for additional regulations for the use of PET isotopes (*6*). They concluded that current state regulations sufficiently addressed any safety issues and no new regulations were required. An advisory notice alerting nuclear medicine facilities on the safe handling of PET isotopes may be issued after ^{18}F becomes readily available.

CONCLUSION

With the advent of high-energy collimators, coincidence cameras and PET distribution networks, expanded use of ^{18}F in nuclear medicine facilities is expected to begin sometime in 1997. This new application requires a review and modification of current safety procedures and equipment that were designed around the use of traditional lower energy radiopharmaceuticals. Fluorine-18 generates a dose rate six times greater than that for ^{99m}Tc at the same distance and activity. Time, distance and shielding are still valid and applicable safety protocols for external exposure control. However, the high energy of ^{18}F requires approximately 16 times more lead for the equivalent shielding effect as ^{99m}Tc, thereby necessitating the use of additional shielding equipment to ensure adequate protection. Lead equivalent products designed from tungsten may be used if available to limit the weight and size of the shields, especially for cylindrical shapes that will be picked up or carried. Specially made vial shields, syringe shields, dose drawing stations and dose administration shields specifically designed for PET isotopes have become commercially available.

The half-life of PET isotopes is extremely short, with the half-life of ^{18}F equal to 110 min. This greatly reduces the radiation safety hazards from waste, spills and postadministration patient control. Current suggested state radiation safety regulations are adequate for application to PET isotopes for medical use, but the new 10CFR Part 20 gives the NRC jurisdiction to keep occupational dose within regulated limits. The use of ^{18}F can be safely expanded to nuclear medicine facilities with minimal administrative changes and the addition of special shielding containers specifically designed to handle high-energy isotopes.

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