BNWL-SA - 4761

CONF- 731123--3

A BIOMEDICAL PROBE USING A FIBER OPTIC COUPLED SCINTILLATOR*

K. L. Swinth+
Battelle
Pacific Northwest Laboratories
Richland, Washington 99352

and

J. H. Ewins
Laboratory of Nuclear Medicine and Radiation
Biology
University of California at Los Angeles
Los Angeles, California 90024

* This paper is based on work performed under United States Atomic Energy Commission contracts AT(45-1)-1830 and AT(04-1) GEN-12.

+ On temporary assignment at Laboratory of Nuclear Medicine and Radiation Biology, UCLA.

DISCLAIMER

Introduction

A biomedical scintillation counter similar in size to a fiber-optic gastroscope was built to count low concentrations of internal radioactivity. The probe was built specifically to count the uranium L x-rays (average energy 17 keV) emitted in the decay of plutonium located in the tracheobronchial lymph nodes. Plutonium is found in the tracheobronchial lymph nodes due to translocation of inhaled insoluble plutonium compounds.(1) Efforts with the scintillation technique resulted in a probe with higher sensitivity than a previously developed solid-state avalanche diode probe.(2) Although the probe was developed specifically for counting plutonium, it can be readily applied to counting other isotopes with low energy photons such as ^{131}I, ^{125}I, ^{197}Hg, ^{67}Ga, ^{133}Xe, ^{99m}Tc, etc.

The purpose of this paper is to describe both the physical features and the performance of a fiber optic coupled scintillation counter and to compare the probe with similar solid-state probes. Previous fiber optic probes have been built for dosimetry;(3) however, the present probe is designed for low background, low energy, high sensitivity photon counting.

Description of Probe

Coupling a scintillation crystal to a photomultiplier with fiber optics causes an 80-90% loss in the scintillation intensity reaching the photomultiplier. At low energies this is serious because it causes a loss of resolution and moves the scintillation spectra into the noise region of the phototube. For low background counting and an improved signal to noise ratio, one must reduce the background from the photomultiplier. This was done by selecting a low-noise, high-gain phototube, magnetically limiting the active area of the photocathode and by reducing the number of fast rise time pulses by a pulse-shape discrimination technique.(4) Use of these techniques reduces the background count rate in the energy range from 6.5-40 keV by a factor of 50 or more.

Figure 1 shows a sectional drawing of the detector and of the probe. The NaI(Tl) crystal, surrounded by MgO is encased in a thin aluminum shell (0.25 mm) with a glass optical window sealed in place by epoxy. The detector is coupled to the fiber optic light guide by an optically transparent epoxy, an aluminum ferrule for mechanical strength and an optically opaque epoxy which also occludes light and fills the remaining void.

The fiber optic light guide is constructed of a 4.7 mm diameter bundle of 50 micron glass fibers having a numerical aperture of 0.66. The ends of the bundles are secured, bonded and polished in stainless steel ferrules to which an opaque outer sheath of polyvinyl chloride is attached.

The completed assembly is coated with a pure polyurethane resin and cured slowly at room temperature and high humidity. This material protects the probe from the caustic fluids within the digestive tract and softens the edges and corners of the probe. Due primarily to the thermal sensitivity of the NaI(Tl) crystal the probe cannot be autoclaved; however, gas sterilization has been used without degrading the performance or longevity of the probe.

Figure 2 shows a photograph of the assembled probe with the detector and photomultiplier; a preamplifier is encased in the housing with the phototube. This unit uses a 2.5 cm long by 4.7 mm diameter crystal coupled to the photomultiplier by 60 cm of fiber optics. Three other probes using a 90 cm fiber optic coupling have been built and resulted in performance comparable to the present unit. Probes could be built in a large variety of sizes and shapes constrained primarily by light collection in the crystal geometry selected and by light transmission of the fiber optics. In general, suitable operation of larger volume detectors [NaI(Tl)] operating at low energies (10-40 keV) would be difficult whereas smaller volume detectors could be easily engineered as could probes for higher energies.

Probe Performance

The calculated absorption in the detector for normally incident photons is 100% below 80 keV; however, at low energies attenuation in the aluminum encapsulation becomes important. For example, the intensity of the uranium L x-rays emitted in the decay of plutonium (13.6, 17.0 and 20.2 keV) is reduced 38% by the 0.25 mm thickness of aluminum encapsulation. Above 80 keV the efficiency begins to fall and is reduced to 65% at 150 keV.

Figure 3 shows ^{99m}Tc spectra taken with the probe. The poor resolution is caused by light losses resulting from the fiber optic coupling of the crystal. Measured resolution at 59.6 keV with the probe crystal directly coupled to the photomultiplier is 17%; however, with the fiber optics, the resolution is reduced to 54%.

Table 1 shows measured count rates (10-150 keV) for various sources located 5 cm from the probe in air. Shielded and unshielded background count rates over three energy intervals are listed in Table 2. Although pulse-shape discrimination does not appear important in reducing the background for the probe, this is not generally the case. This was a phototube with excellent low noise characteristics and tests on other phototubes required the use of pulse-shape discrimination for ultimate background reduction.

The count rate with the scintillation probe immersed in a 500 ml beaker containing a solution of ^{99m}Tc (Figure 3) is 5.6×10^{6} cpm/μCi/ml. The count rate for a GaAs biomedical probe under similar conditions is 200 cpm/μCi/ml[5] and results published for a large volume CdTe biomedication probe[6] (50 mm^3) indicate a count rate of 6.3×10^{4} cpm/μCi/ml in a more favorable geometry (smaller beaker). It must be noted that both the CdTe (50 mm^3) and GaAs (0.19 mm^3) probes have much smaller volumes than the scintillation probe with an active volume of 450 mm^3. The probe could be built in a smaller volume configuration; however, it was designed for a problem where high counting efficiency is required.

Previous esophageal probes for plutonium detection were built using avalanche silicon diodes as detectors.[(2)] These diodes make excellent detectors, but are limited in ultimate sensitivity by their low absorption of incident photons. The calculated intrinsic efficiency of these diodes is approximately 12% based on normally incident L x-rays from plutonium decay. The diodes have extremely low background rates (∿1 cpm) which help compensate for their poor efficiency. In contrast, thin detectors of NaI(Tl) with approximately 100% efficiency have long been used to detect these x-rays. Table 3 shows typical backgrounds, sensitivities and minimum detectable amounts (MDA) for avalanche probes and a NaI(Tl) probe in a thorax phantom representing the human tracheobronchial lymph node geometry. The avalanche diode probes in Table 3 consist of a three diode array to improve angular response.[(2)]

All plutonium consists of its various isotopes[(6)] and ^{241}Am, a daughter of ^{241}Pu. Radiations from the ^{241}Am are an interference and must be corrected for in interpretation of results. The 59.6 keV gamma from ^{241}Am can reach the esophageal detector from the lung field whereas the L x-rays from the lung are heavily attenuated. Avalanche probes have a poor efficiency at 59.6 keV (∿1/2%) which reduces the interference while pulse-height selection can be used to control the interference for the scintillation proble. The 59.6 keV sensitivities listed in Table 3 are for materials located in the lymph nodes.

Probe Application

The probe has been used to count plutonium translocated to the tracheobronchial lymph nodes of dogs which had been exposed to a respirable aerosol of $^{239}PuO_2$.(1) Previous results(2) with the avalanche probes indicated that a sensitivity of greater than 1 cpm/nCi could be expected based on the relative sensitivities of the two types of probes. Figure 4 shows one of the probes located near the tracheal bifurcation at the position of maximum count rate while Figure 5 shows a scan of count rate versus position referenced to the incisors. Four dogs were counted and based on thoraic lymph node burdens estimated by extrapolation from measured build up in sacrificed animals and whole body counts, the sensitivity varied from 0.45 to 1.9 cpm/nCi with an average of 1.2 cpm/nCi. These results are reasonable considering the large errors in the extrapolations due to biological variability and due to possible errors in whole body counting. Actual sensitivities will be higher since tracheobronchial burdens constitute only a portion of the thoracic (tracheobronchial, mediastinal, sternal) lymph node burdens used in estimating sensitivities.

Conclusions and Discussion

Probes have been built for use in clinical medicine which utilize silicon p-n and p-i-n detectors;(8,9,10) however, their use has been primarily in beta counting. Noise levels from such detectors are usually too high to permit efficient counting of the plutonium x-rays. As larger volume solid-state detectors such as CdTe, GaAs and HgI_2 become available they will be applied to plutonium detection problems and they are already finding some use in medical applications.(5,6) HgI_2,with its high absorption, could lead to small volume gamma probes with high efficiency.(11) The excellent pulse-height resolution and durability of solid-state detectors are their major attributes.

The large volume scintillation probe showed excellent performance and has a higher sensitivity than present biomedical solid-state probes including one designed for the same application. Although the probe was designed to estimate lymph node burdens in humans, it could readily be adapted to other biomedical measurements such as detection of esophageal cancer. Kobayashi(5) and Takayanagi(8) have described many applications for biomedical probes to which the present probe could be adapted.

Acknowledgments

The authors would like to thank G. C. Huth for his support and L. S. Graham, Medical Physics group at UCLA, for his assistance in performing the ^{99m}Tc measurements.

References

(1) J. F. Park, Health Physics, 22, 803 (1972).

(2) K. L. Swinth, J. F. Park and P. J. Moldofsky, Health Physics, 22, 899 (1972).

(3) J. Byfield, P. Chan and J. E. Morgan, Radiology, 92, 1120 (1969).

(4) K. L. Swinth, IEEE Transactions on Nuclear Science, NS-21 (1974), to be published.

(5) T. Kobayashi, T. Sugita, S. Takayanagi, M. Iio and Y. Sasaki, IEEE Transactions on Nuclear Science, NS-20, 310 (1973)

(6) G. V. Walford and R. P. Parker, IEEE Transactions on Nuclear Science, NS-20, 318 (1973).

(7) K. L. Swinth, IEEE Transactions on Nuclear Science, NS-18, 125 (1971).

(8) S. Takayanagi, T. Kobayashi, T. Sugita, M. Ieo and Y. Sasaki, IEEE Transactions on Nuclear Science, NS-19, 226 (1972).

(9) J. H. Ewins, S. S. Friedland, A. Magevsky and H. S. Katzenstein, IEEE Transactions on Nuclear Science, NS-17, 275 (1970).

(10) J. H. Ewins, C. Marcus, M. Cosgrove, H. Katzenstein and S. Friedland, IEEE Transactions on Nuclear Science, NS-20, 329 (1973).

(11) L. Lavoie, Physics in Medicine and Biology, 18, 120 (1973).

Table 1: Probe Count Rates for Various Sources

Source	Count Rate (10- 150 keV) (Counts/minute/μCi)	Principal Energies (keV)
Am-241	3870 ± 28	17, 60
Co-57	5330 ± 56	122, 136, 14.4
Cd-109	5670 ± 46	23, 88
"Mock" I-131	~8440 ± 41	30, 80
Pu-239*	70 ± 2	17
Pu-239+	462 ± 5	17, 60

* Isotopically Pure Pu-239
\+ Typically encountered "Pu-239" with other Pu isotopes and Am-241 daughter.

Table 2: Background Count Rates

Pulse Shape Discrimination	Shield*	Background Counts (cpm) 10-40 KeV	40-90 KeV	90-150 KeV
No	No	19.9 ± 1.2	51.9 ± 1.9	73.9 ± 2.2
Yes	No	15 ± 1	56.0 ± 1.9	70.6 ± 2.2
No	Yes	7.1 ± 0.7	1.3 ± 0.3	7.5 ± 0.7
Yes	Yes	3.9 ± 0.5	1.5 ± 0.3	9.1 ± 0.8

* 3.9 cm lead shield around detector.

Table 3: Comparison of Probe Sensitivities in Human Thorax Phantom

Probe	Background, B^+ (cpm)	Sensitivity, S (cpm/nCi)	Minimum Detectable Amount (nCi)*
Avalanche 1	5.05	0.025	52
Avalanche 2	5.56	0.039	34
Fiber Optics			
X-Ray Energies	$18.4\ (9.20)^+$	0.807	3.0
γ-Ray Energies	$38.7\ (3.06)^+$	1.22	2.9

---

*Calculated at 90% confidence level with a 20 min counting period

i.e., $MDA = \frac{2.56}{S}\sqrt{\frac{B}{20}}$

+Values in parentheses are for shielded detector.

Figure Captions

Figure 1: Sectional Drawing of Detector and of Probe

Figure 2: Photograph of probe showing the detector, fiber optics and photomultiplier housing.

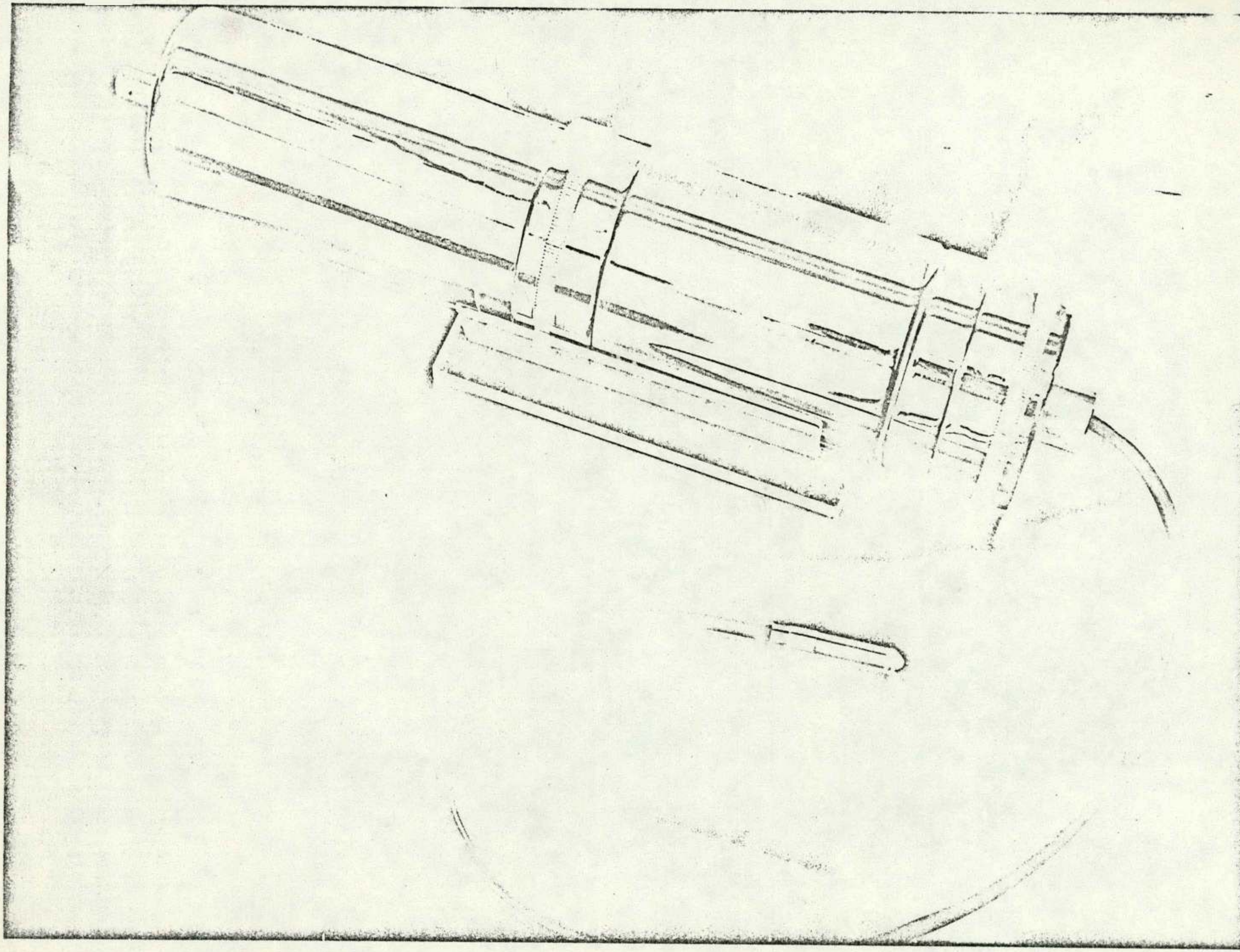
Figure 3: Spectra taken with sources of ^{99m}Tc

Figure 4: X-ray showing probe in the esophagus of a dog and near the tracheal bifurcation

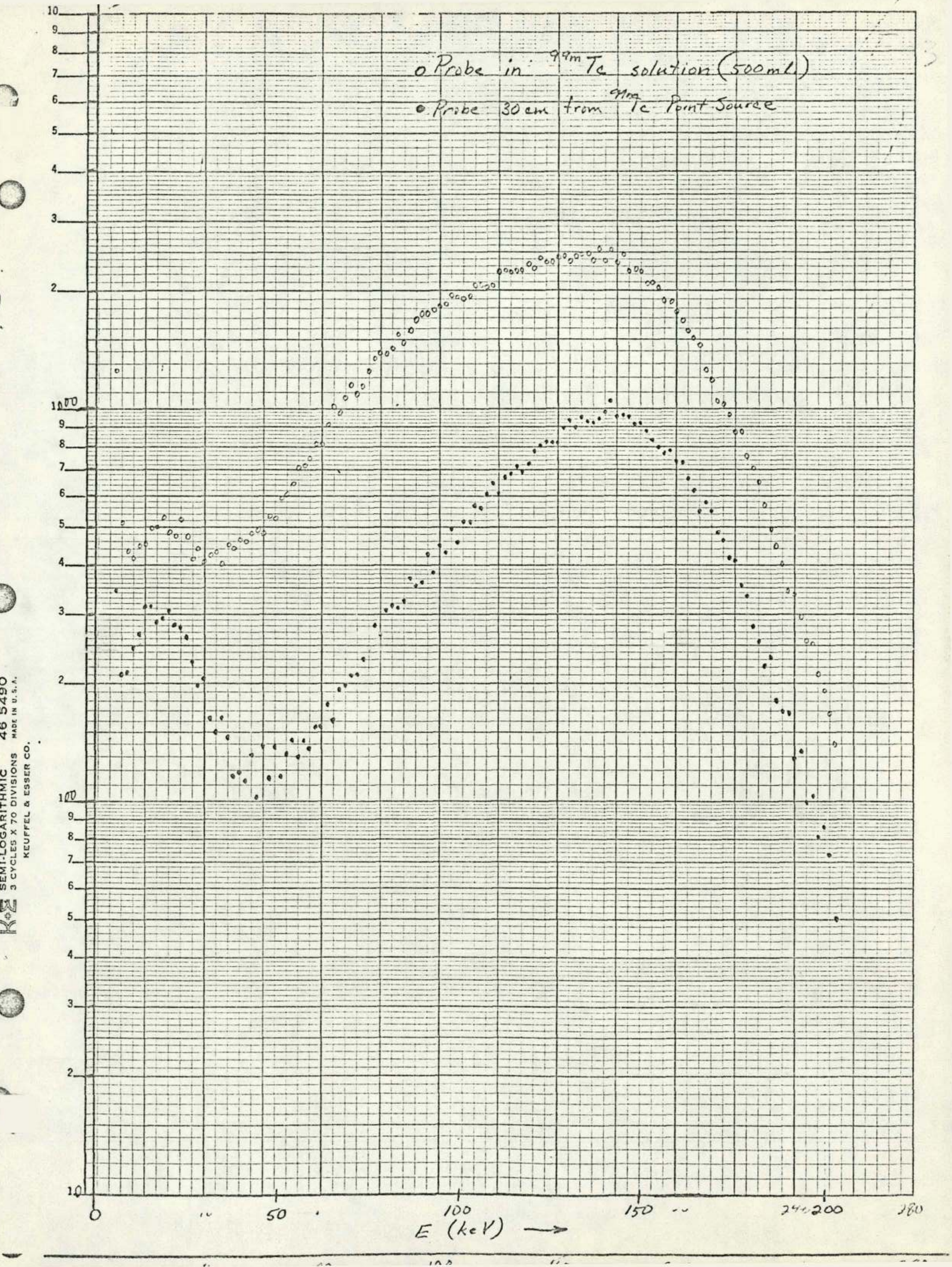
Figure 5: Count rate vs postion as the probe is moved down the esophagus of a dog with plutonium in its tracheobronchial lymph nodes

FIGURE 1 still in preparation.

PNL 73178 -3CN

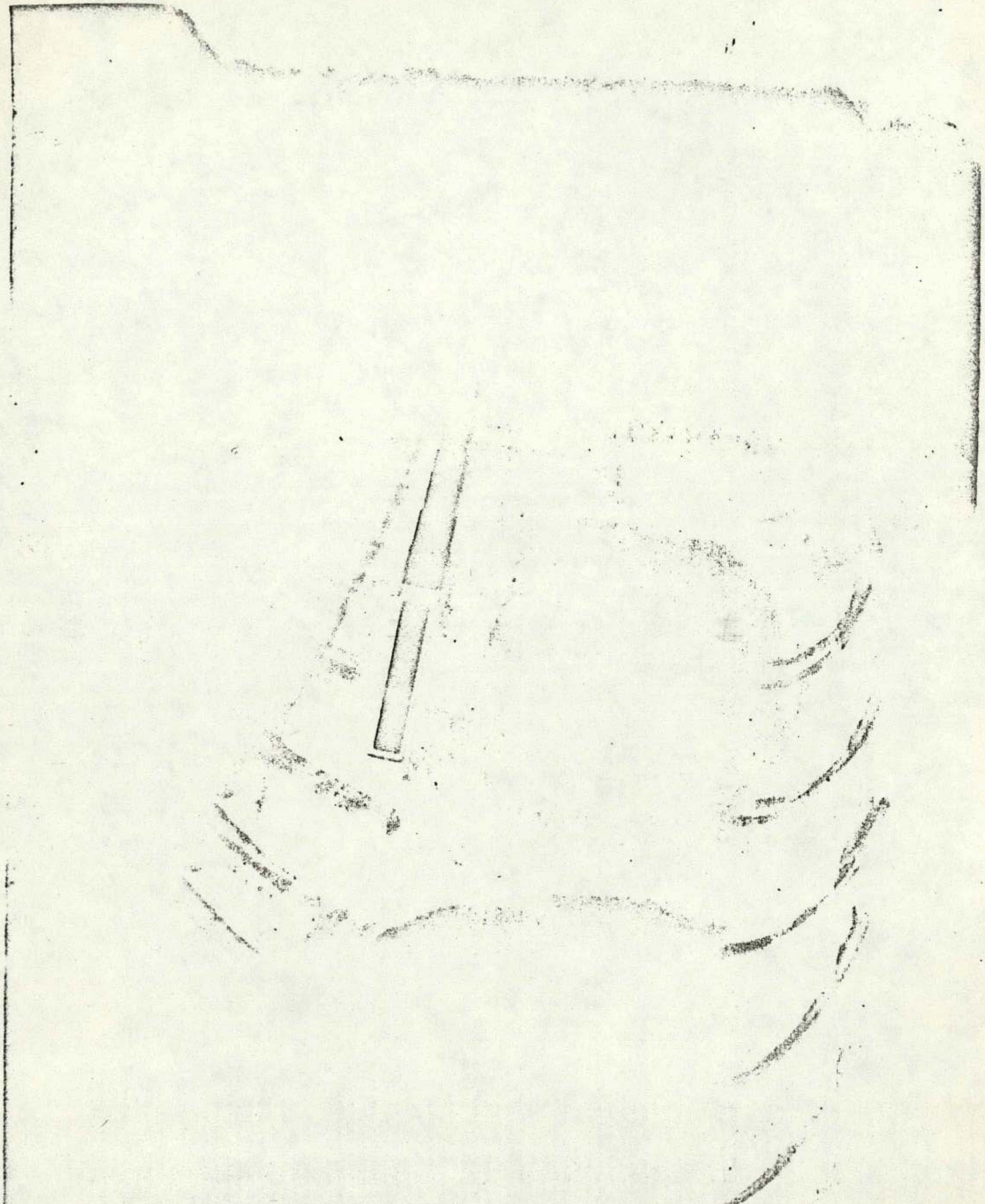


F-2

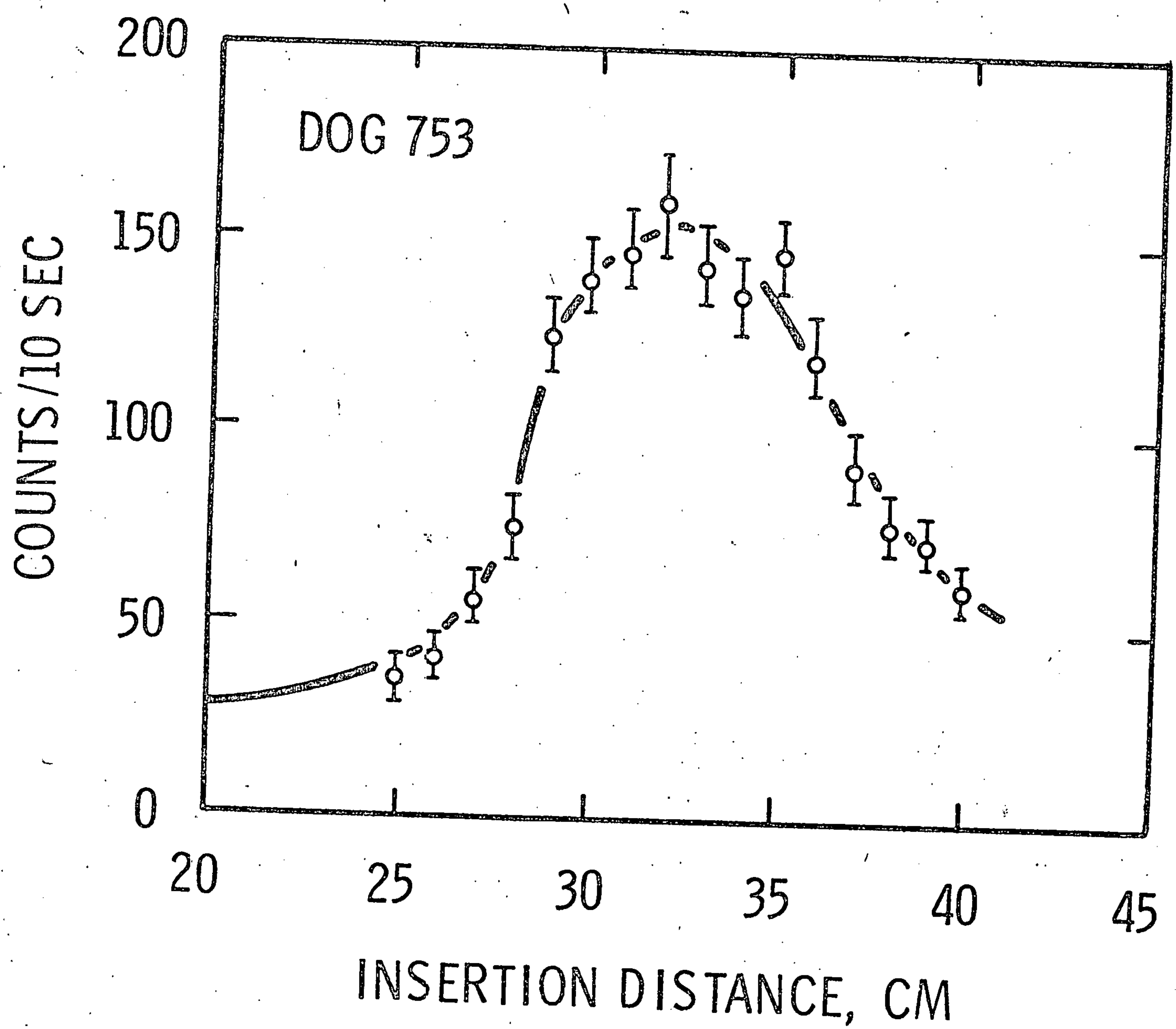


o Probe in 99m Tc solution (500 ml)
• Probe 30 cm from 99m Tc Point Source
E (keV) →
0
50
100
150
200
1000
100
10
K&E SEMI-LOGARITHMIC 46 5490
3 CYCLES X 70 DIVISIONS MADE IN U.S.A.
KEUFFEL & ESSER CO.

F-4



PNL 7 779-2



DOG 753
COUNTS/10 SEC
200
150
100
50
0
20
25
30
35
40
45
INSERTION DISTANCE, CM