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BIOPHYSICAL STUDIES WITH THE CO_2 LASER

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The high power levels available from CO_2 lasersare of potential value in biology and medicine. Investigations of the mechanisms of interaction of CO_2 laser radiation with tissue is required for meaningful application to biology and medicine and for the demarcation of both short and long term hazards (1). This report summarizes a heat flow analysis of threshold lesions in rodent skin and tissue ablation in rabbits produced by CO_2 laser radiation.

Abdomens of shaved and depilated C-57 black mice were exposed to 10 micron radiation at incident power densities of 5 watts/cm^2, 3.3 watts/cm^2 and 2 watts/cm^2. Exposure times were varied between 20 milliseconds and 5 seconds. Approximate 100 per cent lesion frequency thresholds (Figures 1,2) were established on the basis of 40 exposures at 5 watts/cm^2, 50 exposures at 3.3 watts/cm^2 and 80 exposures at 2 watts/cm^2. In each case, a lesion was recorded as " immediate" if it appeared as a whitish area within one hour after the exposure. These "immediate" lesions were 3-5 mm in diameter in contrast to the 6.5-8 mm beam diameter used in these exposures.

Incident power densities of $5.0 \pm 0.8 \frac{w}{cm^2}$ at 0.5 seconds exposure, $3.3 \pm 0.5 \frac{w}{cm^2}$ at 1.25 seconds exposure and $2.0 \pm 0.2 \frac{w}{cm^2}$ at 3.0 seconds exposure almost always resulted in an "immediate" visible lesion. This experimental data is shown in Figures 1 and 2. "Immediate" visible lesions were also observed in 50 per cent of the trials at $5.0 \pm 0.8 \frac{w}{cm^2}$ for 0.16 seconds exposure, and at $2.0 \pm 0.2 \frac{w}{cm^2}$ for 1.10 seconds exposure.

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If the irradiated skin and underlying tissue is assumed to be a uniform, semi-infinite medium with absorption coefficient, α cm^{-1}, then the following equation for one-dimensional heat flow holds for the maximum surface temperature rise at the center of the irradiated spot.

$$\Delta T = (1-R)h \left\{ \frac{\sqrt{\tau}}{3.70\sqrt{K\rho C}} - \left[\frac{1+\exp(\alpha^2 D\tau) \cdot \operatorname{erfc}(\alpha\sqrt{D\tau})}{\alpha K} \right] \right\} \quad (1)$$

In using equation 1 for the analysis of threshold lesions, the value of $\sqrt{K\rho C}$ was chosen as 0.042 c.g.s. units, based on measurements of the thermal conductivity $K = 1.76 \times 10^{-3}$ c.g.s. and the volumetric heat capacity, $\rho C = 1$ by Davies (2). The incident irradiance, h, is measured in watts/cm^2; the exposure time, τ, in seconds. The thermal diffusivity is given by $D = \frac{K}{\rho c}$. The reflectivity, $R \approx 5\%$, is based on observations by Hardy (3) and on a value of the refractive index, $n \approx 1.6$, obtained experimentally.

In calculating a maximum surface temperature, errors resulting from radial heat flow during the pulse are less than 10 per cent at the beam center if the square of the beam radius, a, is greater than $4D\tau$. In these experimental investigations, the ratio $4D\tau/a^2$ was approximately 0.14. Consequently, radial heat flow is negligible near the beam center during the radiant pulse, and one-dimensional heat flow can be considered to apply in this region.

The absorption coefficients for tissues are high in the wavelength region about 10 microns. Some of these data are summarized in Figure 3 and compared to measured absorption in several rat tissues. Since the absorption coefficients at 10.6 microns (Figure 3) range from 130 cm^{-1} to 270 cm^{-1}, radiation at this wavelength is attenuated in tissue by 60

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per cent at depths ranging from 37 to 77 microns. A reasonable approximation for a heat flow analysis is, therefore, one in which the tissue is considered as completely opaque. Consequently, in the limit as $\alpha \rightarrow \infty$ equation 1 reduces to the following form:

$$\Delta T = \frac{(1 - R)\, h \sqrt{\tau}}{3.70 \sqrt{K \rho C}} \qquad (2)$$

Based on the thermal constants given above, equation 2 predicts a maximum surface temperature rise at the lesion center of 22°C ± 15 per cent for the thresholds shown in Figure 1. Actual temperature measurements utilizing a thermocouple injected following the pulse were 22°C ± 20 per cent, in good agreement with calculated values. Using this value of 22°C, a constant temperature curve (isotherm) is drawn in Figure 1 representing the probability of inducing a gross, immediate lesion in almost all cases. This isotherm is in excellent agreement with the experimental data. Values of h and τ to the right of this isotherm should result in an immediate lesion approaching 100 per cent probability.

The absorption coefficient that most accurately fits the experimental data shown in Figure 1 is $\alpha = \infty$. Using equation 1, which assumes a non-opaque medium, additional isotherms for 22°C are drawn in Figure 1 for $\alpha = 150\ cm^{-1}$ and $\alpha = 270\ cm^{-1}$. For long pulse durations ($\tau > 1$ sec.) there is little difference between the isotherms derived from the opaque and non-opaque models and, in fact, for $\alpha > 200\ cm^{-1}$ the isotherms are within experimental accuracy. However, from equation 1, and as shown by Figure 1, for shorter exposure times, $\tau \approx (4D\alpha^2)^{-1}$, it will be necessary to know α within close tolerances since an opaque model will not describe

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the surface boundary conditions with sufficient accuracy. The spectroscopic values of α shown in Figure 3 ($\alpha \approx 150$ cm^{-1}) are low compared to the *in vivo* estimates of α shown in Figure 1 ($\alpha > 200$ cm^{-1}). Dehydration of *in vitro* samples during spectroscopy may be responsible for these low values of absorption. Consequently, in developing a non-opaque model for tissue for short exposure times, it may be necessary to determine a true absorption coefficient from *in vivo* thermal data.

In summary, based on experimental data, one can estimate the values of CO_2 laser power density and exposure time that will result in approximately 100 per cent probability of producing a mild, immediate skin lesion in mice over the range in Figure 1 by calculating an exposure that satisfies the equation $h = 3.60/\sqrt{\tau}$. These equations are of the proper form for an opaque, homogeneous model for tissue. The experimental data in Figure 2 shows that the total energy density required to produce an "immediate" visible lesion decreases with time. For shorter pulse durations, a diathermanous model is required. Although a specific temperature has been used in the above discussion, the thermal model employed has general applicability in relating h and τ for other gross tissue injury.

Tissue ablation in rabbits, dogs and monkeys has been studied on irradiation at 10 microns. Figure 4 contains a set of data points for ablative penetration of *in vivo* rabbit skin at a power density of 100 watts/cm^2. The solid curve represents a model developed by Landau (4) for calculating material losses in melting and ablating solids. This model assumes one-dimensional heat flow and a constant ablation rate. Curve A is drawn using the thermal constants for tissue given above with $\rho = 1.0$ gm/cm^3 (Figure 4). Additional assumptions for Curve A include a latent heat of vaporization

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for tissue, L = 540 cal/gm and a critical temperature for tissue carbonization of T = 300°C (5). Carbonization and ablation are assumed to occur at the same critical temperature. Curve B is drawn for water "ablating" at 100°C. The theoretical curves agree with most of the data points within a factor of two. An improvement in this model should be achieved through an investigation of the thermal constants of tissue above 100°C and by a closer evaluation of the temperature associated with the onset of ablation.

The simplified tissue optical problem at 10.6 microns results in a heat flow analysis that is manageable. For laser wavelengths in the visible and near infrared, lower absorption coefficients and radiation scatter within tissue present serious problems when a heat flow analysis is attempted. Investigations with the CO_2 laser should yield information pertinent to the development of thermal models for immediate, gross tissue injury for wavelengths in the visible and infrared. Since the absorption coefficient at ultraviolet wavelengths is high, studies at 10 microns will be applicable to immediate gross thermal injury in studies with UV lasers. Photoreactive injury in the UV region will require further consideration.

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LIST OF FIGURES

Figure 1 - Isotherms for approximate 100% frequency for visible threshold lesions of rodent skin calculated from opaque and non-opaque thermal models (log-log plot).

Figure 2 - Energy density for approximately 100% lesion frequency threshold for rodent skin. Experimental data curve calculated from opaque compared to thermal model.

Figure 3 - Absorption coefficients for *in vitro* biological tissues.

Figure 4 - Experimental data and theoretical models for ablative penetration in rabbit tissue at an incident power density of 100 watts/cm^2.

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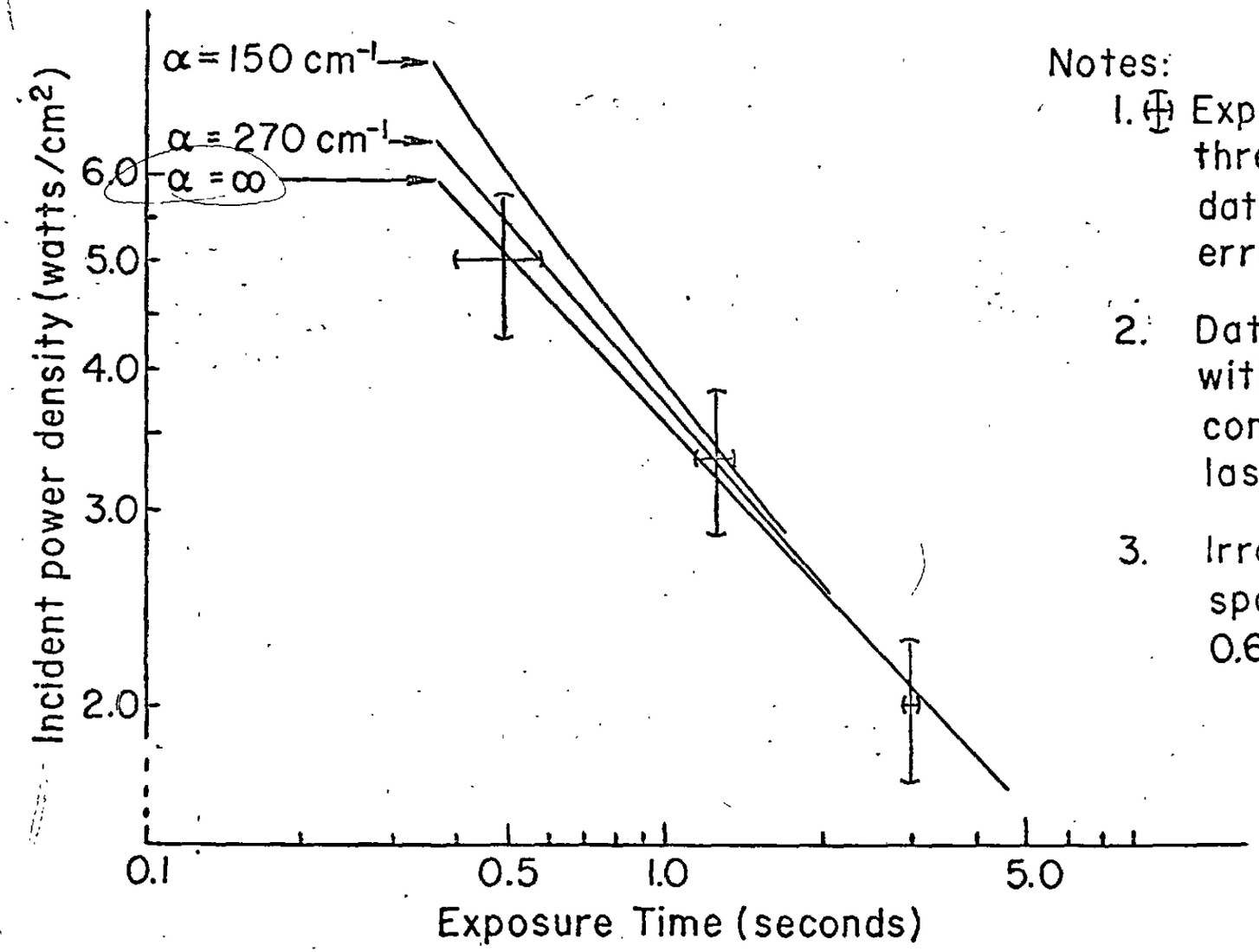

Notes:

1. ⊕ Experimental threshold data with error bars
2. Data obtained with 20 watt continous CO_2 laser
3. Irradiation spot size 0.65-0.80 cm

Fig. 1

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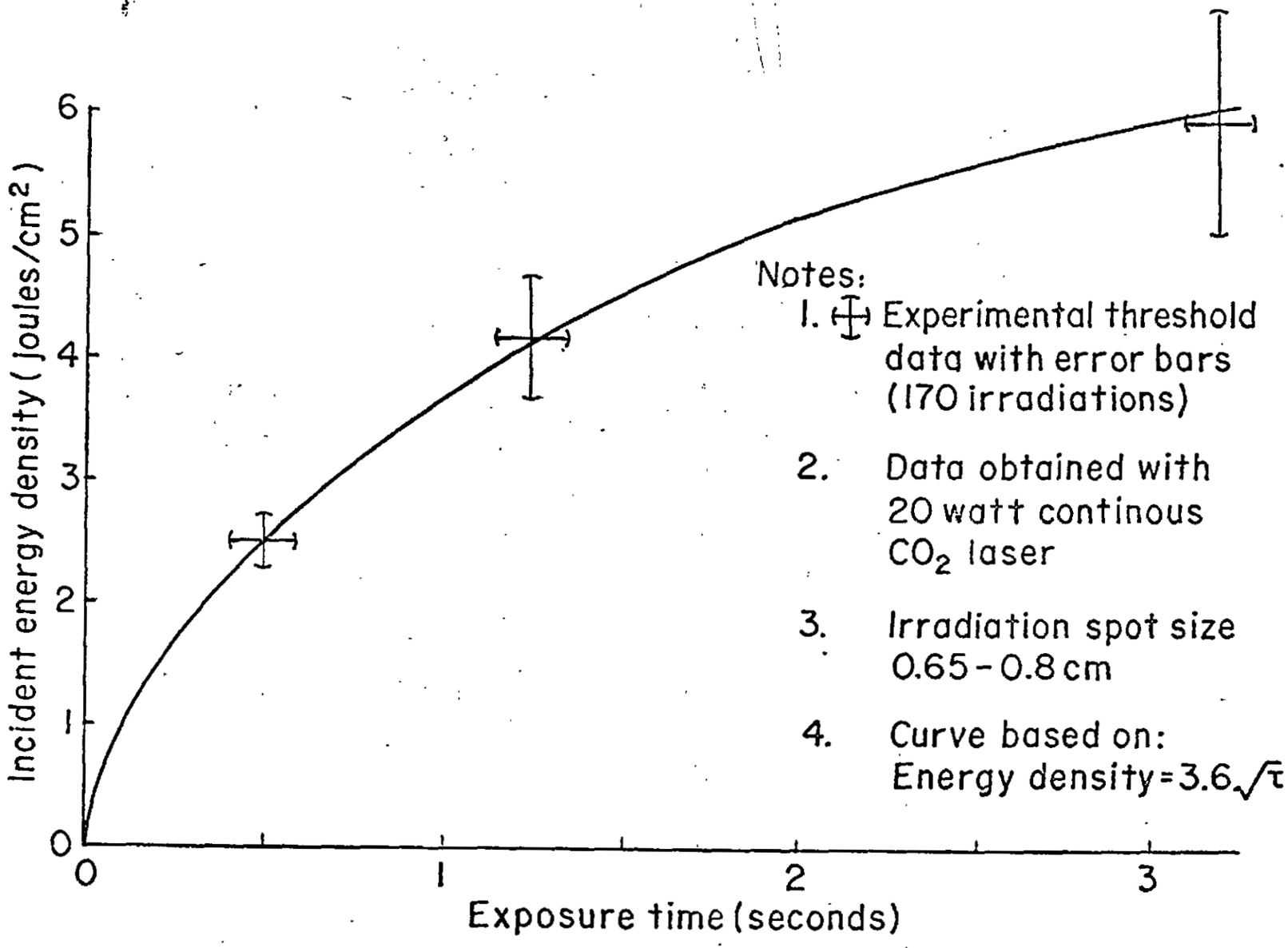

Fig. 2

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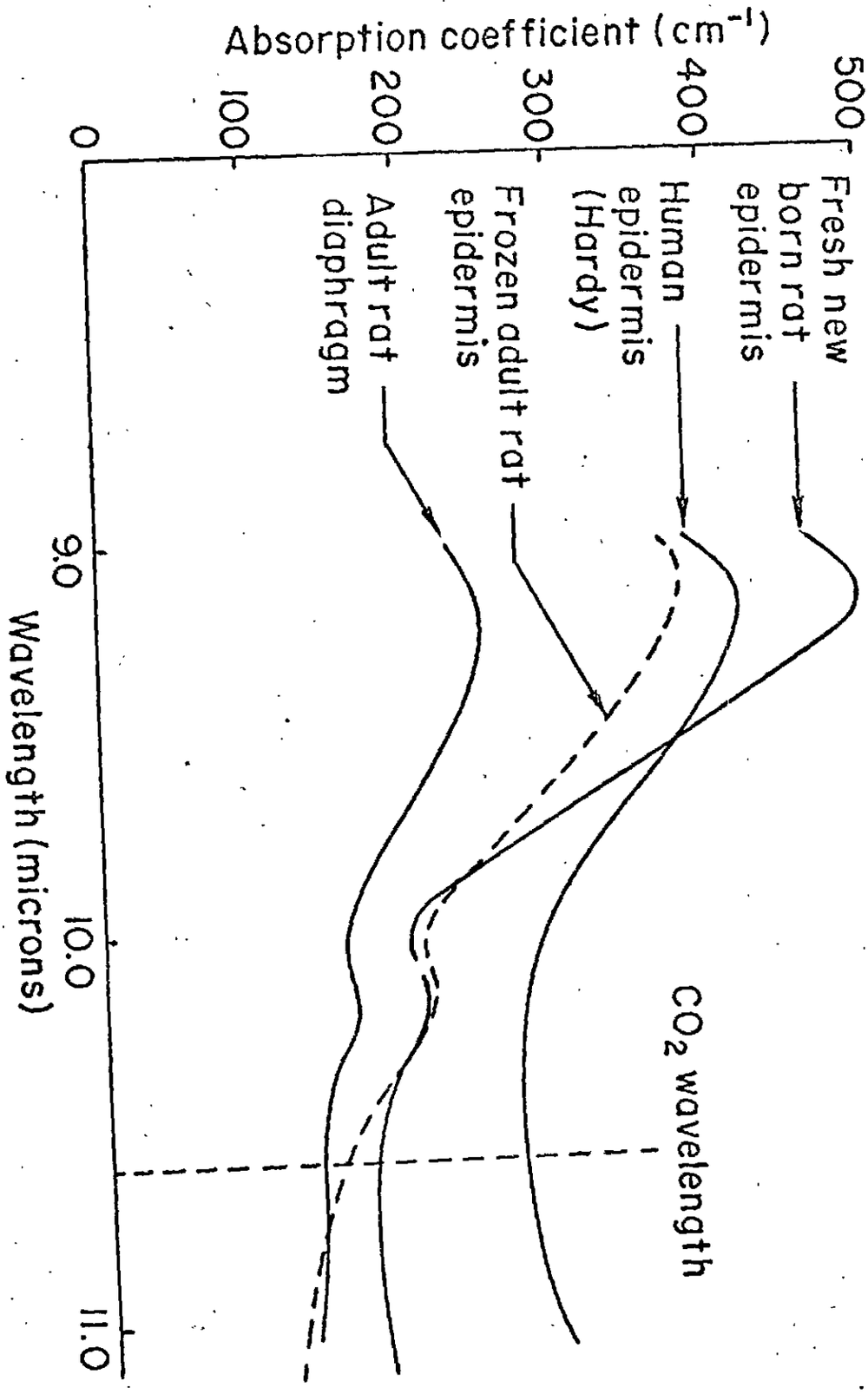

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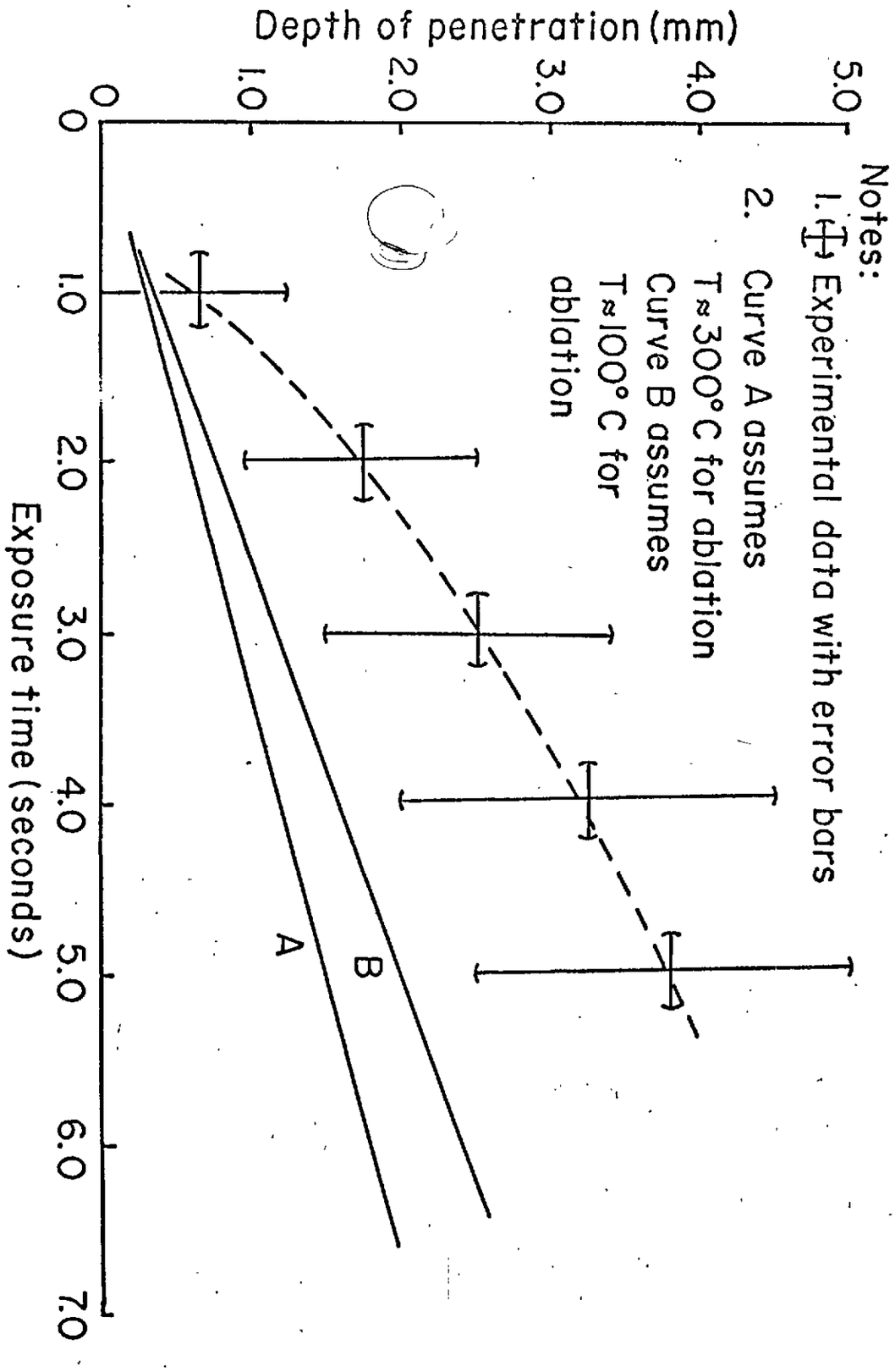

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