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THERAPEUTIC APPLICATIONS OF ELECTROMAGNETIC POWER

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BY

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Therapeutic Applications of Electromagnetic Power

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Abstract—The use of electromagnetic (EM) power for therapeutic applications has existed since EM sources have been available to man. Physical medicine has been a major user of both shortwave (27.33 MHz) and microwave (2450 MHz) diathermy over the decades in which the EM power has been used to heat deep tissues for stimulating various medically beneficial physiologic responses in the relief of certain pathological conditions. Experimental and clinical research indicates that these responses will occur as a result of elevating the tissue temperatures in the range 41° to 45°C requiring absorbed power densities from 50 to 170 W/kg in the deep tissues where treatment is desired. The combination of pain responses and a large reserve of blood cooling capacity seems to be sufficient for limiting the heating to safe, but therapeutic levels in vasculated and innervated tissue. Recent research has shown that the use of the industrial, scientific, and medical (ISM) frequency of 915 MHz is more efficient than the currently used 2450-MHz microwave frequency in terms of maximum power transfer to deep tissues. The results also show that in addition to thermal applications, microwave energy can be used for the controlled transcutaneous stimulation of nerve action potentials via implanted miniature microwave diodes.

I. Introduction

DIATHERMY is a technique used for producing therapeutic heating in tissue by the conversion of physical forms of energy such as ultrasound, EM shortwave or microwaves into heat after being transmitted transcutaneously to deep afflicted tissue areas. The technique has been used in physical medicine from the time that the physical energy sources have been available to man. The various techniques have been covered extensively in the literature; for example, see Licht [1], Schwan [2]–[4], Rogoff [5], Scott [6], Moor [7], and Lehmann [8], [9]. Diathermy is used in the clinic to treat afflictions that normally respond to heat but cannot be reached by surface heating. Though some physicians have achieved considerable success with diathermy, others have not seemed to recognize any benefits at all. It appears that the major problem has been the lack of a good scientific approach in the design and use of the diathermy apparatus for optimal results. Unfortunately, only the earliest and far from optimum equipment, which was designed at a time when the interaction of the EM and ultrasonic energies with biological media was not well understood or quantified, has been available to the therapists over the many decades of popular usage. The shortcomings of this equipment, coupled with public fear of the hazards of the wave energies, has discouraged many practitioners from using diathermy. In this paper, these shortcomings will be discussed and some remedial possibilities will be presented. Since the major concern of this special issue of the PROCEEDINGS is microwave applications, ultrasound diathermy will not be covered, but shortwave diathermy will be discussed, since it is a major competitor of microwave diathermy and provides a means for quantitative comparison.

The evolution of therapeutic heating with electromagnetic (EM) energy has kept close pace with the development of

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EM sources. Knowledge of this historic evolution is important in the elucidation of present-day problems on the medical use and biological effects of microwaves. Interest in the interaction of EM energy with biological tissue dates back to the first man-made EM sources. D'Arsonval, a physician–physiologist, in connection with his work on developing methods for producing high-frequency currents, found, in 1892, that currents of frequency 10 kHz or greater would produce a sensation of warmth without the painful muscular contractions or fatal consequences that could occur at lower frequencies [10]. This led to the use of radio-frequency energy by physicians for the therapeutic heating of diseased tissue. The use of EM energy rather than external heat sources for heating the tissue became popular since the high-frequency currents could penetrate deeply and produce heat in subcutaneous tissues through resistance losses.

The word "diathermy" ($\delta\iota\alpha$ through $\theta\epsilon\rho\mu\eta$ heat) was introduced by Nagelschmidt in 1907 [10] to describe the relatively uniform heating produced in the tissue by the conversion of high-frequency currents into heat. This form of therapy was intended to heat the muscle and joint tissue in contrast to the superficial heating of the cutaneous and subcutaneous tissue obtained by infrared radiation and hot packs. By 1900 physicians were using high-frequency currents between 0.5 and 3 MHz (longwave diathermy) and by 1935 frequencies as high as 10 MHz were used for this purpose. The earliest diathermy sources were large and noisy low-frequency spark-gap generators which required direct electrode connections to the body in order to introduce sufficient current density for therapeutic heating [1]. These spark-gap diathermy generators, which were popular up to as late as 1940, were eventually outlawed by international agreement because of interference with radio services and they were replaced by higher frequency vacuum-tube generators.

In 1928, EM radiations as high as 100 MHz (shortwave diathermy) were being produced by Esau and used clinically by Schliephake [10]. Eventually, at the International Radio Conference held in Atlantic City, N. J., in 1947, it was decided to allocate finite frequency bands for industrial, scientific, and medical (ISM) purposes with the hope that every country would legislate the use of the frequencies uniformly. During the same year, the FCC allocated three frequencies in harmonic progression for shortwave diathermy: 13.66, 27.33, and 40.98 MHz, with respective bandwidths of 15, 320, and 40 kHz. Since generators with wider tolerance were easier and less expensive to construct, the 320-kHz-wide 27.33-MHz band quickly became popular for shortwave diathermy and is still widely used today. In 1937, it was reported by Williams [11] that EM waves with wavelengths of a few centimeters could be focused, and Southworth [12] pointed out that such radiation could be directed along tubes.

The idea of using microwaves for therapy actually originated in Germany in 1938 and 1939 when Holmann [13], [14] discussed the possible application of radio waves of 25-cm wavelength for therapeutics and predicted that these waves could be focused to produce heating of the deep tissues without excessive heating of the skin. Hemingway and Stenstrom [15], in the United States, also suggested the possibility of using the higher frequency radiation for diathermy because of the ability to beam the energy to a selected tissue region. No equipment was available at that time, however, for providing sufficient output for biological work.

Interest developed in the Mayo Clinic [16], [17] for the application of this centimeter-wavelength energy to medicine. In 1938, the magnetron tube, capable of generating microwave frequencies, was developed at Bell Laboratories, but the available power it generated was only 2 to 3 W. Later that year, RCA developed a magnetron capable of generating 20 W and promised that 100 W could be produced. In 1938, a klystron tube was developed at Stanford University, and promises were made that the tube could soon be used for therapeutic purposes. Suddenly, at this time when tubes of sufficient power for therapeutic application were known to exist, they became mysteriously unavailable. It was not until the secret of radar was finally revealed that the medical community realized that such tubes had become frozen for military use during World War II. During that time, the first studies of the effects of microwaves on living mammalian tissue were conducted by the U. S. Armed Forces to dispel fears of possible ill effects of EM radiation upon personnel connected with radar work. The work conducted by Daily, 1943 [18]; Follis, 1946 [19]; and Lidman and Cohn, 1945 [20], on experimental animals indicated no ill effects when the subjects were exposed to radar pulses. After the war, a magnetron tube was developed at M.I.T. capable of generating 400 W at 3000 MHz and made available for medical use. In June, 1946, the Raytheon Company supplied an apparatus using the new tube to the Mayo Clinic for medical research.

The first work on the therapeutic applications of microwaves, started at the Mayo Clinic in 1946 by Krusen *et al.* [16] and Leden *et al.* [17], involved the exposure of test animals to 65 W of 3000-MHz radiation. Trained dogs were instrumented with thermocouples so that the temperature distribution in the thigh could be measured before and after a period of exposure. (The thermocouples were removed during the period of radiation.) The work indicated that the deep tissues could indeed be heated, resulting in a number of physiological responses and increased blood flow to the area treated. It was noted in these experiments that the average temperature rise was greater in the skin and subcutaneous fat than in the deeper muscle tissue. The final temperature in the muscle, however, was higher. This work launched the use of microwave diathermy for application to physical medicine. The focusing characteristics of microwaves at that time were believed to be advantageous in that they provided a means of achieving a wide variety of heating patterns with improved flexibility in therapeutic applications. The fact that the patient was completely free to move away from the director at any time and the freedom from pads, encumbering cables, and toweling commonly used with shortwave diathermy were considered advantageous. The experimental results seemed to indicate that true deep heating was achieved without undue heating of the cutaneous surface. It must be remembered, however, that these conclusions were based on the use of dogs which have thinner layers of subcutaneous fat and muscle than humans. This new microwave modality gained further credence from dielectric data published as a result of research done at M.I.T. during World War II. The data indicated that the absorption of microwaves at a frequency of 2450 MHz in water at 100°F was in the order of 7000 times greater than the absorption at the commonly used shortwave diathermy frequency of 27 MHz. As a result, in 1946 the FCC assigned the frequency of 2450 MHz to physical medicine based on its alleged superiority in therapeutic value. This is a classic example of how the historic lack of engineering in medicine has prolongated ill-conceived practices, not only in medicine, but also in nonrelated industrial applications. The research teams did not consider the fact that electrical properties and geometry of tissue, as well as the wavelength in the tissue, are far more important than the absorption and focusing characteristics of waves in the generation of therapeutic heating patterns. Today, after 28 years, there is no commercial microwave diathermy apparatus available other than that which was conceived at that time. As a matter of fact, the majority of the microwave ovens and industrial food processing units being manufactured and used today operate on 2450 MHz based on the historic microwave diathermy frequency allocation.

After 1950, research on the use of microwaves for diathermy and also on the hazardous biological effects mushroomed. Though the most obvious effects of microwaves were thermal in nature, evidence was sought for the explanation of possible nonthermal effects. Complete references on the biological effects research since 1950 are too numerous to discuss here, but the work has recently been documented in a special issue of the IEEE TRANSACTIONS ON MICROWAVE THEORY AND TECHNIQUES [21], and in papers by Michaelson [22], and Johnson and Guy [23]. Significant work on engineering approaches in quantifying the various effects was done by Schwan at the University of Pennsylvania. Schwan's work on the dielectric properties of biological media and wave propagation and absorption by various tissue geometries deserves considerable attention [2], [3], [24]–[26]. During this period, it was demonstrated theoretically by Schwan that 2450 MHz was not a good choice of frequency for diathermy. He pointed out the major deficiencies: 1) excessive heating in the subcutaneous fat due to standing waves; 2) poor penetration of energy into the muscle tissue due to small skin depth; and 3) poor control and knowledge of energy absorbed by patients due to large variations in electrical thickness (compared to a wavelength) of subcutaneous tissues. He recommended that the frequency be changed to 900 MHz, or less.

Between 1960 and 1966, Lehmann *et al.* [27]–[29] and Guy [30] experimentally verified Schwan's earlier theoretical prediction that 900 MHz or lower frequencies could produce better therapeutic heating patterns than obtained with 2450-MHz energy. Since 1966, Lehmann *et al.* [31], DeLateur [32], and Guy [33], [34] have developed and clinically tested a new direct-contact 915-MHz diathermy aperture source which appears to be therapeutically more effective and safer in terms of leakage radiation than the existing 2450-MHz equipment. Unfortunately, the lower frequency equipment is not commercially available at this time, and there is also great pressure to reduce the bandwidth of the 915-MHz ISM band to allow more spectrum for other uses [35]. It is important that maximum bandwidth be maintained for this useful ISM frequency since it has been demonstrated that it is the optimum allocated frequency for providing the maximum penetration of EM energy into human tissue with a reasonable size source. A wide bandwidth is necessary to allow practical and inexpensive sources to be used for varying clinical conditions.

The future promises many improved and new therapeutic applications of EM energy in the microwave range. In addition to the improved methods of therapeutic heating discussed here, microwaves have been suggested for production of differential hyperthermia in connection with the treatment of cancer [36], reversing a patient's induced hypothermia state in conjunction with open-heart surgery [23], and the transcutaneous transmission of microwave pulses into nervous tissue for conversion into dc pulses by implanted diodes for stimulating nerves. The latter, which will be discussed briefly in this paper, may be used to block pain signals and provide control signals for reactivating lost neuromuscular function.

II. Benefits of Therapeutic Heating

Therapeutic heating is an important technique used in physical medicine, regardless of the source of heat. When local heat is applied to living tissue, the resulting temperature rise will produce many physiologic responses, partly due to direct action on the tissue cells and partly due to thermal action on local nerve receptors. One of the responses is an increase in blood flow due to vasodilatation accompanied by increases in capillary pressure, cellular membrane permeability, and metabolic rate. The latter could result in a further increase in tissue temperature. It is believed that the responses can increase healing rate in diseased or damaged tissue by increasing the transfer of metabolites across cell membranes, providing for greater concentra of white cells and antibodies, and increasing the transport rate of toxins, engulfed bacteria, and debris away from the treated area [37]. The heating can promote relaxation in muscles, reduce pain and provide relief of muscle "spasms" [38], [9]. Heating can also produce changes in the properties of collagenous tissues, as found in tendon, joint capsule, and scarred synovium. As the collagenous tissue is heated to therapeutic levels, the property of viscous flow becomes predominant and tension is reduced [39]. If a physical therapy program of stretch is used in conjunction with heating, as in patients with hip and shoulder limitations, one can take advantage of the increase in extensibility and produce significant increases in range of motion [9], [40]–[42]. Joint stiffness can also be relieved by heating. Backlund and Tiselius have measured the joint stiffness of rheumatoid patients and have shown a decrease in the hysteresis loop after heating the joint [43].

The temperature of the tissues is the most important factor in determining the extent of the physiologic response to heat. Lehmann [9] has shown by animal studies the relationship between the percentage of hyperemia (increase in blood volume) and the temperature (Fig. 1). The results indicate that the tissue temperature must be raised above 41°C to produce any significant reaction and a temperature near 45°C is needed for maximum reaction. The overall body metabolic rate will also initially increase with increased temperature ΔT. The factor of increase is approximately $(1.1)^{\Delta T}$ within physiological limits [44] (e.g., with an initial tissue temperature of 34°C, raising the temperature to 40°C would produce a 77-percent increase in metabolic rate assuming that the increase in a specific tissue metabolic rate is comparable to the metabolic rate increase caused by an increase in the total body temperature, the upper temperature of 45°C probably corresponds closely to the safe upper limit where a further increase could sharply reduce the metabolic rate or stop it altogether [45], [46]). The threshold of thermal pain corresponds to a skin temperature of 45°C with the pain intensity increasing to a maximum at about 65°C. The threshold for irreversible skin tissue damage is also 45°C when the heat is applied for a sufficiently long period of time [47]. For short periods of heat application, the skin can tolerate higher temperatures without damage. For most other tissues, 45°C also appears to be the maximum safe tempera-

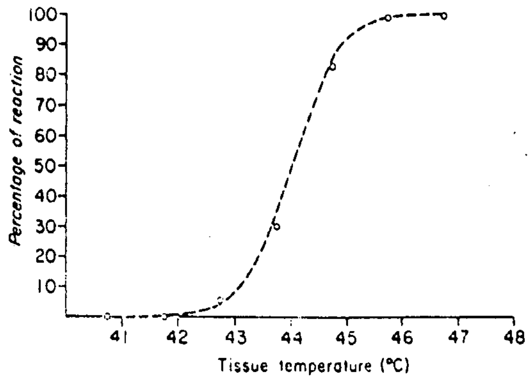

Fig. 1. Dependence of hyperemia on tissue temperature. From Lehmann [73].

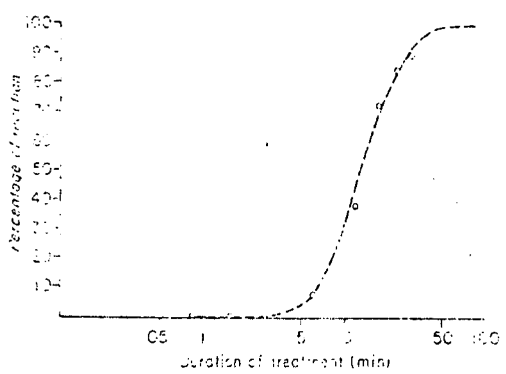

Fig. 2. Dependence of hyperemia on duration of treatment. From Lehmann [73].

ture tolerated without damage [48]. Certain tissues appear to have a lower tolerance, however. For example, the testicles which are normally much lower in temperature than other portions of the body can be affected adversely at temperatures equal to the normal 37°C body temperature [49]. The lens of the eyes are especially vulnerable to radiant-type heating and irreversible damage can occur at elevated temperatures due to the lack of blood circulation and poor tissue repair capabilities [50]. Thus it appears from Fig. 1 that the therapeutic temperature range is not only narrow, but very close to the damaging temperature level.

Lehmann [9] has also shown that the duration of tissue temperature is important in determining the extent of the biological reaction (Fig. 2). The figure indicates that a minimal effective duration of elevation is 3 to 5 min, whereas complete reactions may be obtained with a 30-min application. It is clear that the rate of rise of the temperature plays an important role in determining the extent of the biological response since in the total duration of application only the period where the effective temperature level is obtained would be therapeutically beneficial. Also, the physiologic responses of the nervous system temperature receptors seem to be more pronounced when the rate of temperature elevation is rapid [51], [52].

When the surface heat is applied, the skin temperature will elevate with the accompanying reactions described above. Though this is beneficial for treatment of pathologic lesions at or near the tissue surface, it is ineffective for treating deeper pathologic conditions. The subcutaneous fat will act as a thermal barrier which, combined with the increased surface vasodilatation, will prevent heat flow into the deeper musculature. No increase in deep tissue blood flow will result, and there may even be vasoconstriction to compensate for the increase in surface blood. Nerve reflexes due to the surface heating can produce consensual temperature increases in other parts of the body, e.g., the surface of the opposite extremity, but are less pronounced than the primary increases [38]. Relaxation of the striated skeletal muscles may occur and muscle spasms may be resolved by the surface heating due to reflex nerve reactions from surface temperature receptors. Thus, in general, the surface heating provides only mild physiologic and therapeutic reactions and any effects on the deeper pathologic conditions are only reflex in nature.

Lehmann [53] has demonstrated (Fig. 3), through infrared radiation of the thigh of humans, that surface heating will produce only negligible increases in deep tissue temperature. The measurements illustrated in the figure were made through implanted thermistors while the thigh was exposed to radiation levels where only mild pain or discomfort was produced during a short period of occlusion of blood flow. Immediately after initiation of the radiation (point A), there was a sharp rise in skin temperature. The rate of temperature rise decreased with time until a final temperature of 42°C was reached (point B). After this time, there was a slight decrease in skin temperature until equilibrium was reached throughout the specimen (point C). Shortly after equilibrium, the blood flow was occluded by the application of a tourniquet which caused the skin temperature to increase sharply. A short time later, the power was turned off and the temperature decreased rapidly. Finally, with restored blood flow, there was an even more rapid decrease in temperature. The figure clearly shows the negligible rise in temperature in the deep tissues in contrast to the substantial increase at the surface. The significant increase in blood flow in the superficial tissues is evidenced by the drop in superficial temperature after the peak value of 42°C was reached and the changes in temperature with the blood occlusion. The same results were also typical for both long and short infrared frequencies, each applied to three subjects. Similar results were also obtained through the application of hot packs to human thighs (Fig. 4). Since the hot packs cooled after a short period, the application was repeated every 10 min. It is significant to note that with each pack application, the temperature in the superficial tissues never returned to the original high value produced with the initial pack. This provides more evidence of the cooling effect due to the increase in blood flow in the superficial tissue. Unfortunately, with this type of heating, damaging temperatures would be needed in the superficial tissue in order to obtain therapeutic levels in the deep tissue.

Therapeutic heat treatment of deeper pathologic conditions and chronic disease processes such as joint contractures, chronic pelvic inflammatory disease, arthritis, muscle trauma, fibrositis, sprains and strains, as well as others, can only be done with diathermy. The diathermy technique used must raise the temperatures of the deeper tissues up to therapeutic levels (40°–45°C) without buring the intervening tissues (exceeding 45°C). This is a challenging problem to both the diathermy designer and the clinician since in general the applied energy must be transmitted through a layer of subcutaneous fat, which has little vasculation and cooling capacity, into muscle tissue, which has considerable more vasculation and cooling capability. In addition, the specific heat of the fat layer is lower than that of deeper more vasculated tissues, resulting in a greater temperature rise per unit heat input. Therefore, to produce the same temperature rise,

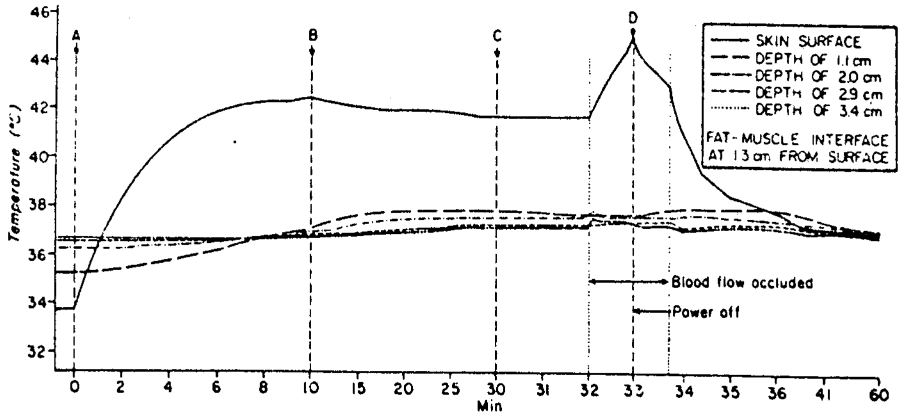

Fig. 3. Temperatures recorded in human thigh during exposure to infrared radiation from 250-W Mazda lamp (red bulb). *A*, temperature distribution before heat is applied; *B*, peak temperature value at skin surface; *C*, temperature equivalent approached throughout specimen; *D*, blood flow obstructed by tourniquet just before heat is discontinued. From Lehmann *et al.* [53].

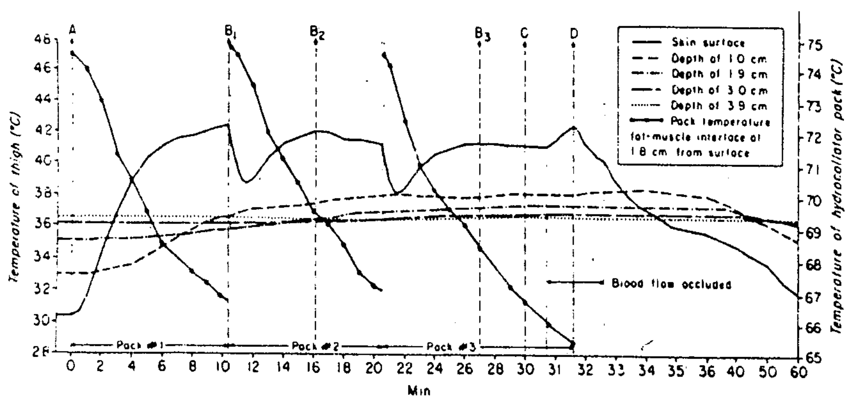

Fig. 4. Temperatures recorded in the human thigh during application with hydrocollator hot packs, repeating the application every 10 min. From Lehmann *et al.* [53].

a much greater amount of energy must be converted to heat in the muscle than in the intervening fat layer. The failure of some diathermy modalities to achieve this is one of the major reasons for their therapeutic ineffectiveness and discontinued use. These include shortwave capacitor-type applicators and 2450-MHz microwave applicators for many clinical problems where deep heating is desired. The use of ultrasound diathermy has proved far more effective in providing deep heating to joint structures such as capsule and synovium. With proper design and selection of frequency, however, microwave can be more effective in heating deep muscular tissue. Another reason is the inadequate understanding by the equipment manufacturer or the user of what the machine output and therapeutic dose relationship is. This relationship will be covered in detail in the following sections.

III. Thermal and Electrical Properties of Tissues

In order to evaluate and understand the therapeutic effectiveness of an applied dose of diathermy energy, one must know the relationship between the absorbed energy, the tissue cooling mechanisms, and the temperature. The energy equation for the time rate of change of temperature (°C/s) per unit volume of subcutaneous tissue heated with diathermy is

$$\frac{d(\Delta T)}{dt} = \frac{0.239 \times 10^{-3}}{c}\left[W_a + W_m - W_c - W_b\right] \quad (1)$$

where W_a is the absorbed power density, W_m is the metabolic heating rate, W_c is the power dissipated by thermal conduction, and W_b is the power dissipated by blood flow, all expressed in W/kg; c is the specific heat of the tissue in kcal/kg·°C and $\Delta T = T - T_0$ is the difference between the tissue temperature T and the initial tissue temperature T_0 prior to treatment.

The absorbed power density for tissue exposed to an EM diathermy source is

$$W_a = 10^{-3}\frac{\sigma}{\rho}\,|E|^2 \quad (2)$$

where σ is the electrical conductivity in mhos/m, ρ is the density in g/cm³, and E is the rms value of the electric field in V/m in the tissue. Within the safe temperature range, the metabolic heating rate may be expressed as

$$W_m = W_0(1.1)^{\Delta T} \quad (3)$$

where W_0 is the initial metabolic heating rate.

The thermal conduction term may be expressed as

$$W_c = \frac{k_c}{\rho}\nabla^2 T \quad (4)$$

where k_c is the thermal conductivity of the tissue in mW/cm·°C and ∇ is the gradient operator.

If it is assumed that blood enters the tissue at arterial temperature T_a and leaves at tissue temperature T,

TABLE I
THERMAL AND PHYSICAL PROPERTIES OF HUMAN TISSUES

Tissue	Subscript	Specific Heat[a] c (kcal/kg·°C)	Density[b] ρ (g/cm³)	Metabolic Rate[c] W_0 (W/kg)	Blood Flow Rate[c] m (ml/100 gm·min)	Thermal Conductivity[d] k_c (mW/cm·°C)
Skeletal muscle (excised)	m		1.07			4.4
Skeletal muscle (living)	m	0.83		0.7	2.7	6.42
Fat	f	0.54	0.937			2.1[b]
Bone (cortical)	bc	0.3	1.79			14.6[b]
Bone (spongy)	bs	0.71	1.25			
Blood	bl	0.93	1.06[e]			5.06
Heart muscle	m			33	84	
Brain (excised)	br					5.0
Brain (living)	br			11	54	8.05
Kidney	k			20	420	
Liver	l			6.7	57.7	
Skin (excised)	s					2.5
Skin (living)	s			1	12.8	4.42
Whole body				1.3	8.6	

[a] Reference [55].
[b] For pig [8].
[c] Calculated from data in [56].
[d] Reference [57] except where noted.
[e] For humans [54].

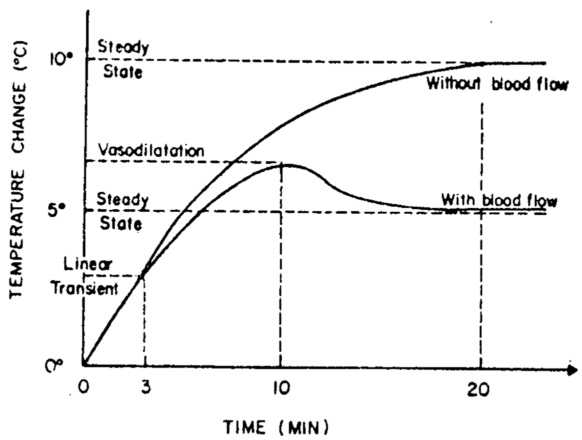

Fig. 5. Schematic representation of transient and steady-state temperature for a typical tissue under diathermy exposure.

$$W_b = \frac{k_2 m c_b}{\rho_b} \Delta T' \qquad (5)$$

where $\Delta T' = T - T_a$, c_b is the specific heat of blood, ρ_b is the density of blood in g/cm³, m is the blood flow rate in milliliters per 100 g·min, and the constant $k_2 = 0.698$.

Prior to the time the diathermy is applied, it is assumed that a steady-state condition exists where $W_a = \Delta T = d(\Delta T)/dt = 0$, requiring

$$W_m - W_c - W_b = 0. \qquad (6)$$

According to the typical values of the physical and thermal properties of tissues given in Table I, [8], [54]-[57], the equilibrium values of the terms in (6) under normal conditions are in the order of 1 W/kg for typical resting muscle. When a therapeutic level of EM power, $50 < W_a < 170$ W/kg, is absorbed, ΔT will increase as shown in Fig. 5 with an initial linear transient period typically lasting about 3 min, where by (1)

$$\frac{d(\Delta T)}{dt} \approx \frac{0.239 \times 10^{-3}}{c} W_a. \qquad (7)$$

This period is followed by a nonlinear transient period usually lasting for another 7 to 10 min where ΔT becomes sufficiently large that blood flow and thermal conduction become important in dissipating the applied power. In tissues with negligible or insufficient blood flow, the temperature will monotonically approach a steady-state value dictated by the magnitude of W_a as shown on the upper curve where equilibrium is reached when $W_a = W_c$. For vasculated tissues, however, blood flow plays a significant part in heat dissipation limiting the slope of the $d(\Delta T)/dt$ curve. In addition, for vasculated tissues, a marked increase in blood flow will occur when the temperature passes through the range 42° to 44°C due to vasodilatation. As a result, the temperature will drop and approach a steady-state value at a somewhat lower level, as shown in the figure, when $W_a = W_c + W_b$, indicating a significant reserve of blood cooling capacity. For proper and safe therapeutic action, it is necessary to raise the temperature sufficiently in the deeper vasculated tissue to trigger the vasodilatation without exceeding safe levels in the poorly vasculated intervening subcutaneous fat layer. Clinical experience has shown that when normal vasculated tissue is exposed to a diathermy source, pain will be noted by the patient before any tissue damage can occur. In fact, the pain may be used as a guide to indicate that the tissue temperature has reached the required 43° to 45°C for vasodilatation and associated therapeutic benefits.

The power absorption density W_a must be sufficiently high so that the therapeutic level of temperature can be maintained over the major portion of the treatment period. If too little power is applied, the period of elevated temperature will be too short for any benefits. If too high a level is applied, the temperature can overshoot the safe level before the vasodilatation can take effect. The pain sensors are a reliable and sensitive means for detecting this temperature range, however, and if the applied power level is set so that only mild pain or discomfort are first experienced by the patient, the vasodilatation will be sufficient to limit or even lower the temperature to a level that is both tolerable and therapeutically effective. If the effective temperature is reached at the surface, it is felt as a mild burning sensation. On the other hand, if it is reached in the deeper tissues, it is felt as a dull aching type of pain. Hardy [47] has shown that the intensity

TABLE II

ELECTRICAL PROPERTIES OF HUMAN TISSUES

	Muscle					Fat			
Frequency (MHz)	Wavelength in Air (cm)	Dielectric Constant ϵ_m	Conductivity σ_m (mho/m)	Depth of Penetration (cm)	Wavelength in Tissue (cm)	Dielectric Constant ϵ_f	Conductivity σ_f (mho/m)	Depth of Penetration[a] (cm)	Wavelength in Tissue[a] (cm)
27.12	1106	113	0.612	14.3	68.1	20	10.9–43.2	159	241
40.68	738	97.3	0.693	11.2	51.3	14.6	12.6–52.8	118	187
100	300	71.7	0.889	6.66	27	7.45	19.1–75.9	60.4	106
433	69.3	53	1.43	3.57	8.76	5.6	37.9–118	26.2	28.8
750	40	52	1.54	3.18	5.34	5.6	49.8–138	23	16.8
915	32.8	51	1.60	3.04	4.46	5.6	55.6–147	17.7	13.7
1500	20	49	1.77	2.42	2.81	5.6	70.8–171	13.9	8.41
2450	12.2	47	2.21	1.70	1.76	5.5	96.4–213	11.2	5.21

[a] Typical values.

Fig. 6. Shortwave diathermy application with condensor pads to back with spacing between skin and electrodes provided by layers of terry cloth.

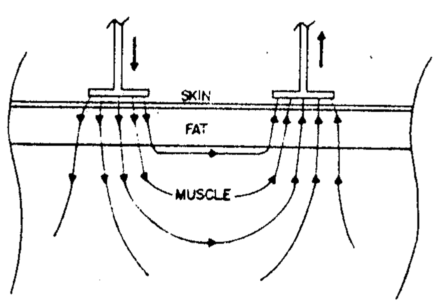

Fig. 7. Cross-sectional sketch showing fields in layered tissue exposed to shortwave diathermy capacitor-type electrodes.

of absorbed nonpenetrating radiation at the threshold of pain in the skin is 0.045 g/cal/cm²/s, or 188 mW/cm². Various methods of achieving therapeutic levels of absorbed power by penetrating EM fields are discussed below.

IV. SHORTWAVE DIATHERMY

The earliest diathermy equipment consisted of a high-frequency generator from which currents were applied directly to the tissues by contacting electrodes. As a result of uneven or poor contact, one of the greatest hazards was the production of burns localized at the electrode–tissue interface. As frequency was increased, the electrodes were designed so that they did not have to make direct contact, since displacement currents between the electrode plates and the tissue surface were sufficient to couple energy to the tissue. Although capacitor electrode arrangements such as those shown in Fig. 6 are still used to treat patients with present-day 27.33-MHz diathermy equipment, there are some fundamental problems. Fig. 7 illustrates how induced conduction currents in the tissue will produce much greater power absorption in the subcutaneous fat than in the skin and muscle tissue, and how the divergence of the current will tend to concentrate the power absorption in the superficial tissue next to the electrodes. For example, if we neglect the spreading of the fields, and note that the electric fields are predominantly normal to the tissue interfaces, the relationship between the fields in the air E_0 and those in the subcutaneous fat E_f and muscle E_m is

$$E_0 = \epsilon_f^* E_f = \epsilon_m^* E_m \tag{8}$$

where $\epsilon_{f,m} = (\epsilon_{f,m} - j\sigma_{f,m}/\omega\epsilon_0)$ are the complex dielectric constants of the fat and muscle, respectively, ω is the angular frequency, and ϵ_0 is the permittivity of free space. Evaluation of the power absorption density P_f and P_m in the fat and muscle, using the physical parameters from Table I and the electrical parameters from Table II, gives [2]–[4], [58]–[61]

$$P_f = \frac{\sigma_f}{\rho_f} \frac{E_0^2}{|\epsilon_f^*|^2} \times 10^{-3} = 3.50 \times 10^{-8} E_0^2 \tag{9}$$

$$P_m = \frac{\sigma_m}{\rho_m} \frac{E_0^2}{|\epsilon_m^*|^2} \times 10^{-3} = 3.68 \times 10^{-9} E_0^2. \tag{10}$$

The results show an order-of-magnitude greater heating in the subcutaneous fat than in the muscle or the skin. Additional selective heating would occur in the fat due to the spreading of the fields as a function of distance from the electrodes. This, along with the fact that the specific heat and density are lower for the fat, as indicated in Table I, would produce a more than 17 times greater rate of temperature rise in the fat than in the muscle. In addition, the blood cooling rate would be significantly less in the fat so that the final steady-state temperature would be considerably higher in that tissue.

Other types of shortwave diathermy applicators are the induction coil arrangements (Figs. 8 and 9) which induce circular eddy currents in the tissues by magnetic induction. The former is a large coil of insulated cable separated from the patient by toweling. The latter, called a "monode" by the manufacturer, is a more compact coil and condensor combination that may be spaced at various distances from the patient by an adjustable supporting arm (not shown). A cross-sectional view of the induced currents (Fig. 10) illustrates the superiority of the inductive applicator over that of the elec-

Fig. 8. Shortwave diathermy application to back with induction coil ("pancake" coil). Spacing between coil and skin is provided by layers of terry cloth.

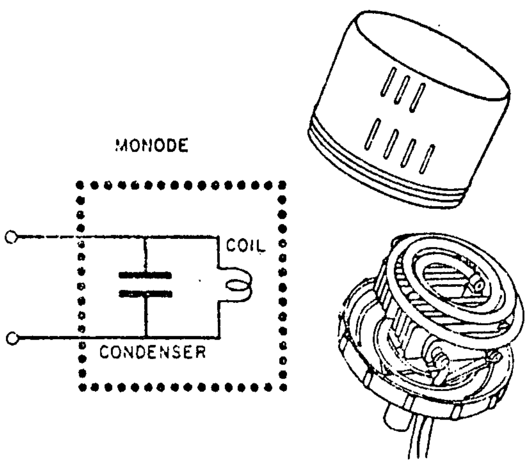

Fig. 9. Compact-type induction coil with wiring arrangement (courtesy of Siemens-Reiniger Werke Ag.).

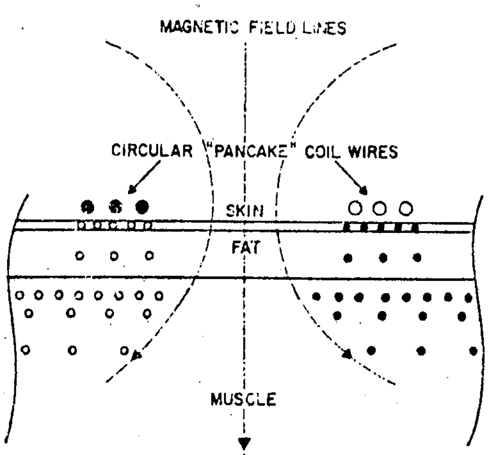

Fig. 10. Cross-sectional sketch showing magnetically induced current in tissue exposed to shortwave diathermy "pancake" coil. Dark dots indicate current density vectors directed into the paper and open circles indicate vectors out of the paper.

trode type. For this case, the induced fields and currents are tangential to the tissue interfaces and are not greatly modified by the tissue boundaries, which was the case for the electrode-type applicator. Ideally, the current density and heating will be higher in the muscle tissue where the conductivity is maximum, as shown schematically in Fig. 10.

Under certain conditions where the diameter and the spacing of the coil turns are excessive, or when the coil is placed too close to the tissue, more energy may be coupled to the subcutaneous fat than to the deeper vasculated tissues. This is caused by the sharp increase in magnetic fields near the coils and the high electric field between the coil turns. This latter coupling is illustrated in Fig. 11.

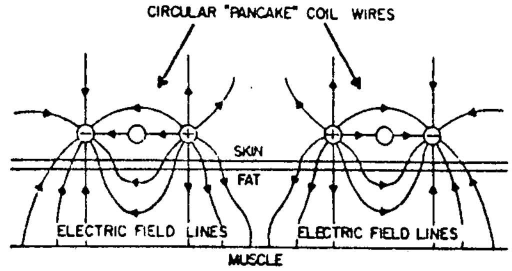

Fig. 11. Cross-sectional sketch showing induced fields in tissue due to intercoil potentials of shortwave diathermy "pancake" coil.

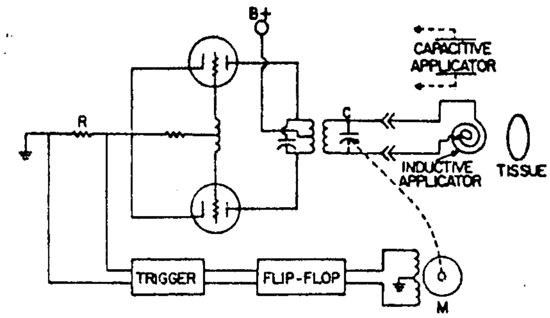

Fig. 12. Simplified circuit schematic of shortwave diathermy generator.

A typical circuit for shortwave diathermy apparatus is shown in a simplified schematic form in Fig. 12. The tank circuit of a high-power high-frequency generator tuned near 27.33 MHz is coupled to a second parallel resonant circuit with variable tuning. Depending on the method of application, the circuit may be coupled to a pair of capacitor electrodes, an inductive coil, or an inductive coil and capacitor combination, as shown in Figs. 6, 8, and 9. Under different clinical conditions, the capacitance between the electrodes and the patient will vary requiring adjustments in generator tuning. This is generally done automatically; for example, in the manner shown in Fig. 12. A motor M is used to continually rotate the tuning capacitor C. As the circuit is tuned through resonance, the change in the plate current sensed by the voltage across resistor R triggers a flip-flop circuit to reverse the direction of rotation of the tuning motor so that the capacitor is again driven through resonance. The hunting action of the tuning capacitor across resonance insures that the circuit stays tuned under varying clinical conditions. A variable-output power switch is usually provided and a timer controls the exposure time of a patient. It is almost impossible, however, for the physician to determine the amount of power a patient is absorbing from the various applicators for different spacings and various power settings.

A great deal of insight and some quantitative information concerning absorbed power can be gained through a simple theoretical analysis of the coupling characteristics between the patient and the applicator. Since the inductive coupling appears most effective, we will examine the case of a planar skin-fat-muscle tissue geometry exposed to a flat "pancake" coil with coordinates and parameters as defined in Fig. 13. Since the size of the coil is small compared to the 11-m wave-

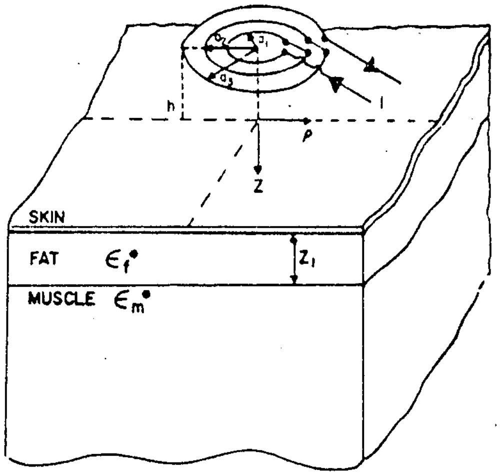

Fig. 13. Geometry and coordinates for a skin-fat-muscle tissue geometry exposed to a flat "pancake" diathermy induction coil.

length, the mathematics can be greatly simplified by approximating the actual spiral coil with perfect concentric loops connected in series and assuming quasi-stationary field conditions. The well-known vector potential and magnetic field expressions for a single closed loop [62] may be used to express the vector potential A_ϕ and magnetic field component H_z of the coil:

$$A_\phi = \frac{\mu I}{\pi(\rho)^{1/2}} \sum_{i=1}^{n} \frac{(a_i)^{1/2}}{k_i}\left(1 - \frac{1}{2}k_i^2\right)[K(k_i) - E(k_i)] \quad (11)$$

$$H_z = \frac{\mu I}{4\pi(\rho)^{1/2}} \sum_{i=1}^{n} \frac{k_i}{(a_i)^{1/2}} \cdot \left[K(k_i) + \frac{a_i^2 - \rho^2 - (z+h)^2}{(a_i-\rho)^2 + (z+h)^2}\right]E(k_i) \quad (12)$$

where

$$k_i^2 = \frac{4\rho a_i}{(\rho + a_i)^2 + (z+h)^2} \quad (13)$$

and where $K(k_i)$ and $E(k_i)$ are elliptical integrals of the first and second kind, a_i is the radius of the ith loop, n is the number of loops, I is the loop current, μ is the permeability of free space, and ρ and z are cylindrical coordinates of the point of observation. The magnetically induced electric field component E_ϕ may be expressed as $E_\phi = -j\omega A_\phi$ which at short-wave diathermy frequencies can be assumed to penetrate the tissues without significant perturbation since the tissues are nearly transparent to the near-field inductive components of the coil. There will also be a significant radial and axial component of electric field $\overline{E} = -\nabla\phi$ originating from the potential ϕ due to the interturn voltages of the coil. Since the fields are maximum in the radial direction between the concentric turns, we will make a first-order approximation th each turn i is at a constant potential V_i, but the voltage between turns I and $i+1$ is

$$V_i - V_{i-1} = 2\pi f L_i I \quad (14)$$

where

$$L_i = \mu \sum_{j=1}^{n} (a_i + a_j - a_0)\left(1 - \frac{k_{ij}}{2}\right)[K(k_{ij}) - E(k_{ij})] \quad (15)$$

is the inductance of the ith turn

$$k_{ij}^2 = \frac{4a_i(a_j - a_0)}{(a_i + a_j - a_0)^2} \quad (16)$$

and a_0 is the radius of the coil conductor.

The potential of the nth turn V_n is assumed to be

$$V_n = V/2 \quad (17)$$

where

$$V = 2\pi f I \sum_{i=2}^{n} L_i \quad (18)$$

and the inductance of the coil is

$$L = \sum_{i=1}^{n} L_i. \quad (19)$$

A rigorous solution of the potential field Φ in the tissues would require proper accounting of the images of the coil due to tissue interfaces. Since both the fat and the muscle have high dielectric constants, however, the potential distributions above the tissue may be approximated by considering a single perfect image along with the actual coil [62], giving

$$\Phi = \sum_{n=1}^{n} A_i \left\{\frac{K(k_i^+)}{[(\rho + a_i)^2 + (z+h)^2]^{1/2}} - \frac{K(k_i^-)}{[(\rho + a_i)^2 + (z-h)^2]^{1/2}}\right\} \quad (20)$$

where

$$k_i^+ = \frac{4\rho a_i}{(\rho + a_i)^2 + (z+h)^2} \quad (21)$$

$$k_i^- = \frac{4\rho a_i}{(\rho + a_i)^2 + (z-h)^2} \quad (22)$$

and the values of A_i are found by evaluating the set of simultaneous equations obtained when (20) is evaluated at $z = -h$ and $\rho = a_1, a_2, \cdots, a_n$ for the known voltages $V_1, V_2, \cdots, V_n$.

The field normal to the surface of the tissue may be evaluated from the gradient of the potential distribution at $z=0$. Then, since it was shown previously that the major heating due to a field perpendicular to the tissue interface occurs in the subcutaneous fat, we obtain

$$E_z = \frac{1}{|\epsilon_f^*|} \sum_{i=1}^{n} A_i \left\{\frac{2h}{[(\rho + a_i)^2 + h^2]^{3/2}}\left[K(k_i) - \frac{4\rho a_i}{(\rho - a_i)^2 + h^2} B(k_i)\right]\right\} \quad (23)$$

in the fat where

$$B(k_i) = \frac{E(k_i)}{k_i^2} - \frac{1 - k_i^2}{k_i^2} K(k_i) \quad (24)$$

and k_i is simply $k_i^\pm$ evaluated at $z=0$.

If we ignore the field spreading and other quasi-static field components because of the close proximity of the fat-muscle interface, we may obtain an estimate of the absorbed

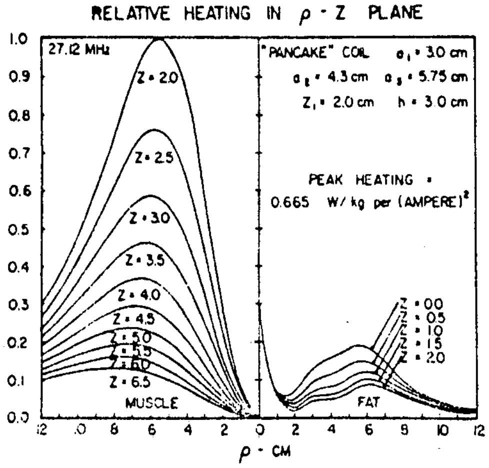

Fig. 14. Calculated absorbed power patterns in plane geometry and muscle tissue layers exposed to shortwave diathermy induction coil.

power in the fat

$$W_a = \frac{\sigma_f}{\rho_f}\left[E_i^2 + E_o^2\right] \times 10^{-3} \qquad (25)$$

due to both the induction field and the significant component of the quasi-static field.

When evaluating (25), one should keep in mind that the most desirable heating or absorption patterns for therapeutic purposes correspond to minimum relative heating in the fat with maximum relative heating and depth of penetration into the muscle. Fig. 14 illustrates the calculated results for tissues exposed to a flat coil with the same wire thickness and radii of turns as the commercial applicator (Fig. 9). Three concentric loops provide the closest approximation for this case. With such few turns, it is more convenient to assume that the total applied voltage calculated from the coil current and inductance was distributed equally between the center and the inner and outer loops. A coil current of 1 A, a fat thickness of $z_1 = 2$ cm, and a spacing of 3 cm between the applicator coil and the surface of the fat are assumed. The results show that the coil induces a toroidal heating pattern with a maximum heating of 0.665 W/kg in the muscle at a radial distance $\rho = 5.5$ from the coil axis with a penetration depth (depth where heating drops by a factor of e^{-2} from the maximum) into the muscle of about 4 cm. The maximum heating in the fat which occurs on the axis is approximately one-third that of the muscle. A second lower peak occurs in the fat at $\rho = 5.8$ cm. The former is due to the coupling from the electric field between the loops, while the latter is due to the electric field induced by magnetic coupling. The value of heating for other values of coil current may be obtained by multiplying the results given in the figure by the square of the coil current. The value of coil current varies according to the power output setting of the generator, the spacing between the coil and the patient, and the geometry of the exposed tissue. Typical values for the five power settings available for the equipment (Siemens Ultratherm 608) powering the type of applicator chosen are listed in Table III. These were determined by comparing magnetic fields measured with a small shielded loop along the axis of the applicator and calculating the equivalent current from (12) for the theoretical coil which would produce the same field. It is convenient to reduce (12) to

TABLE III

MEASURED MAGNETIC FIELDS (SURFACE OF EXPOSED SUBJECT) AND CALCULATED EQUIVALENT INDUCTION COIL CURRENT AS A FUNCTION OF POWER-SETTING FOR SIEMENS ULTRATHERM 608 SHORTWAVE DIATHERMY EQUIPMENT

Power Setting	Coil to Subject Spacing (cm)	Unloaded Coil		Exposing Human Thigh	
		I (A)	H_z (A/m)	I (A)	H_z (A/m)
1	5.0	3.40	31.5		
2	5.0	5.15	47.5		
3	3.0	7.12	128	5.44	97.8
	5.0	7.12	65.7		
4	5.0	8.85	81.7		
5	1.5	9.87	291	7.39	218
	2.0	9.87	278		
	3.0	9.87	182	7.54	139
	3.5	9.87	152	8.22	127
	4.0	9.87	126		
	5.0	9.87	91.0		

Note: Coil inductance: 0.8 μH (measured); 0.83 μH (theory).

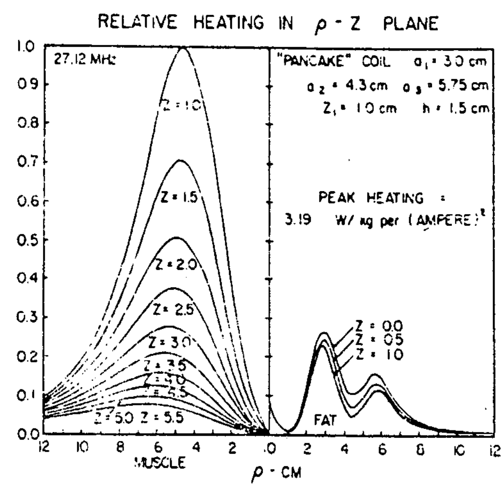

Fig. 15. Calculated absorbed power patterns in plane geometry fat and muscle tissue layers exposed to shortwave diathermy induction coil.

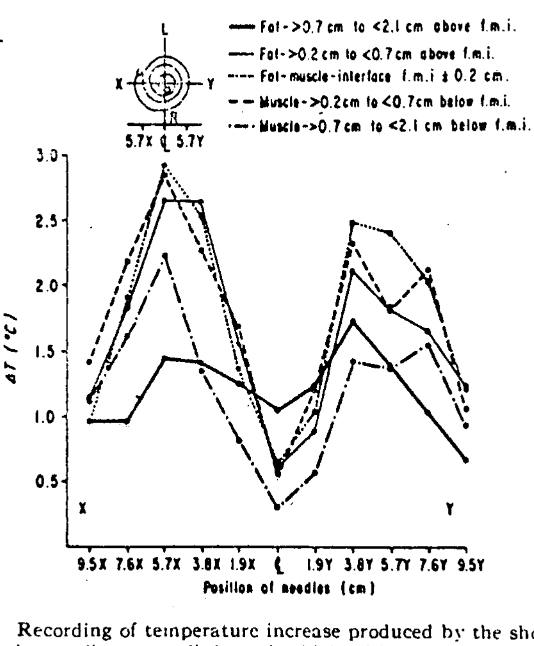

Fig. 16. Recording of temperature increase produced by the shortwave inductive applicator applied to pig thigh with $z_1 = 2$ and $h = 3$. From Lehmann *et al.* [63].

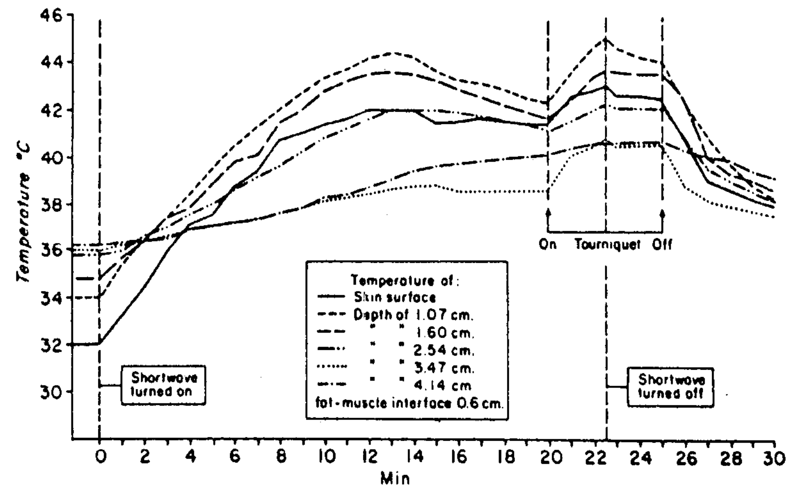

Fig. 17. Recording of temperature increase produced by the shortwave inductive applicator applied to human thigh with $z_1 = 0.6$ and $h = 3.5$. From Lehmann *et al.* [64].

$$H_z = \frac{I}{2}\sum_{i=1}^{n} \frac{a_i^2}{(a_i^2 + z^2)^{3/2}} \tag{26}$$

for this case. Fig. 15 illustrates the heating patterns calculated for the coils spaced 1.5 cm from the tissue surface for a 1-cm-thick fat layer. The peak heating for this case is greater by more than a factor of four, the relative fat heating has increased slightly, the depth of penetration has decreased to approximately 3 cm, and the radius for maximum absorption in the muscle decreased by 1 cm. Lehmann [63] has exposed large specimens of thighs from freshly slaughtered pigs to experimentally determine the heating patterns of the commercial applicator, with the results shown in Fig. 16. The specimens were large compared to the applicator and contained 2 cm of subcutaneous fat. The applicator coils were placed 3 cm from the specimen, and power (position 3 setting) was applied for 5 min. The shapes of the measured temperature curves in the muscle correspond closely to the relative heating predicted by theory. The absolute values cannot be compared, however, since the actual current was unknown and exposure time was too long during the experiment for the linear transient relation to hold. The results of a later experiment conducted by Lehmann [64], however, on live human tissue are shown in Fig. 17, where the temperature in the exposed tissue in the region of maximum absorption was measured as a function of time. It should be noted, however, that the temperatures were measured with implanted thermistors during the period of exposure. Though recent studies [23], [65] have shown that serious artifacts can result from the use of metallic probes in the presence of EM fields, the miniature thermistor used was designed with small-diameter high-resistance leads placed perpendicular to the circulating eddy currents. The absence of appreciable artifact was verified by comparison of temperatures measured in the field by thermistors and alcohol thermometers. Extreme care was exercised to maintain artifact-free measurements [63]. Maximum power available from the shortwave equipment (position 5, $I = 8.22$ A) and a spacing between coil and tissue of 3.5 cm was used. This also corresponded to a tolerable dose where only mild pain was experienced by the patient after 12 min of exposure. The initial transient rise for each temperature curve was used to obtain the power absorption density from (7) at each point of measurement and was compared with the theoretical calculations, with the results shown in Table IV. The skin absorption is not compared since the temperature is much more dependent on uncontrollable surface conditions. The complete theoretical curves are shown in Fig. 18. After 20-min exposure to the shortwave applicator, a tourniquet was inflated to obstruct arterial blood flow. After 2.5 min, the power was turned off and the blood flow was later restored. The curve for the temperature approximately 0.5 cm below the fat–muscle interface indicates a maximum initial linear increase of nearly 1.1°C/min. This corresponds to a calculated absorbed power $W_a = 70$ W/kg. After about 12 min, when the tissue temperature reached 44°C, correspond-

TABLE IV

COMPARISON OF MEASURED AND THEORETICAL SHORTWAVE POWER ABSORPTION AND CALCULATED BLOOD FLOW RATE IN HUMAN THIGH MUSCLE

Distance (cm) z	1.07	1.60	2.54	3.47	4.14
Calculated W_a	61	42.5	21.9	14.9	11
Measured W_a	70	50	24	10	6
Blood flow heat dissipation W_b	81	81	25.4	—	—
Estimated blood flow rate m	13.7	13.7	4.3	—	—

Note: $z_1 = 0.6$ cm, $h = 3.5$ cm, and $I = 8.22$ A.

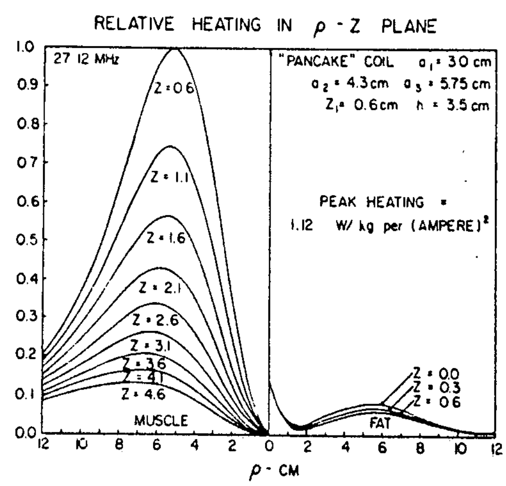

Fig. 18. Calculated absorbed power patterns in plane geometry fat and muscle tissue layers exposed to a shortwave diathermy induction coil.

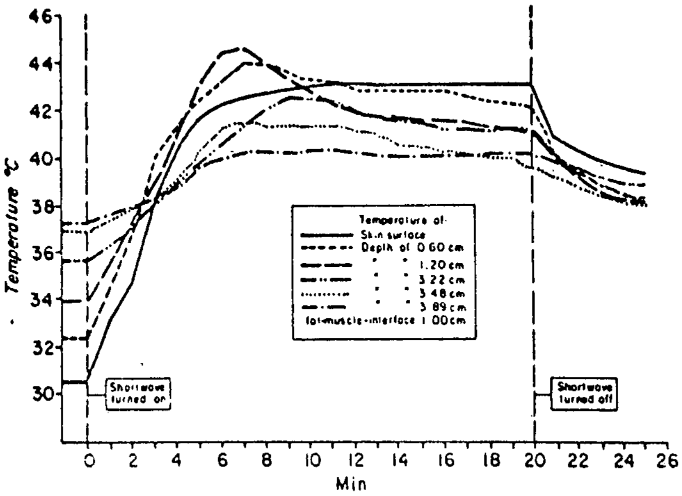

Fig. 19. Recording of temperature increase produced by the shortwave inductive applicator applied to human thigh with $z_1 = 1.0$ and $h = 1.8$. From Lehmann *et al.* [64].

ing to the point of impending discomfort by the patient, there was a marked change in $d(\Delta T)/dt$ to a negative 0.3°C/min, or an estimated 17 W/kg more heat dissipation is needed to maintain a steady-state condition. This necessarily implies by (1) that the blood cooling rate W_b has increased substantially due to an increase in the blood flow rate m. When the arterial blood flow was occluded, $d(\Delta T)/dt$ again changed sharply and became a positive 1.1°C/min, indicating the 70 W/kg of absorbed shortwave power. The results imply that a total heat dissipation of 87 W/kg was provided by blood cooling. If one assumes that the arterial blood arrives at the tissue site at core temperature (approximately 6.0°C below local tissue temperature), we may estimate a flow rate by (5) of approximately 13.5 ml per 100 g·min, which is substantially greater than the initial value estimated from Table I. The increase in temperature at a depth of 3.47 cm after occlusion is believed to be due to a transient displacement of heated blood to that particular site by the occlusion process. According to the results, there is sufficient blood cooling reserve under the particular conditions to maintain a constant tissue temperature below 44°C, even with an increase of 10 to 20 percent in applied power. The slow rate of cooling after the applied power is removed indicates the much stronger role of the blood flow cooling over that of conduction.

One of the recommended methods of using this particular shortwave applicator is to place the cover (1.5 cm from coils) directly against the patient with an intervening 0.3-cm-thick terry cloth spacer. The theoretical heating patterns for a case close to this are given in Fig. 15, and the measured temperature changes in the human thigh exposed for the same power setting (position 5, $I = 7.39$ A) with a 1-cm fat thickness are shown in Fig. 19. The theoretical curves for the $h = 1.5$-cm case can be used for comparison with the $h = 1.8$-cm measurements, provided 0.3 cm is added to the desired value of z. These results clearly show (Table V) the increased absorbed power due to the closer proximity of the coil to the tissue. The results again indicate a triggering of blood flow when the muscle temperature rises above 44°C, which is followed by a $d(\Delta T)/dt = 0.6$°C/min at a depth of 1.2 cm. In this case, however, the transition took place after only 6 min of exposure, coincident with pain that exceeded the threshold level. For this reason, the power applicator had to be terminated prematurely with approximately half the subjects tested under these conditions. From the energy balance equation (1) and from (5), the blood cooling rate was estimated to be 131 W/kg which would require a flow rate of 22 ml per 100 g·min or more. We may note in the first case with the $h = 3$-cm spacing, the deeper muscle tissue reached higher temperature than the surface; whereas, in this latter case, the reverse is true. It appears that the insulating characteristics of the terry cloth in preventing skin cooling may be responsible. The higher measured than theoretical subcutaneous fat heating can be due to 1) errors in measurement resulting from a high thermal gradient near the fat–muscle interface, 2) a higher electrical conductivity in the live fat than obtained from measurements on dead tissue upon which the theory is based, or 3) inaccuracies due to the approximations used in the theoretical equations. We may conclude from the above study that 1) inductor-type shortwave diathermy applicators are effective in elevating the temperature of deep tissue while maintaining cooler surface tissues; 2) the shortwave technique is capable of producing 70 to 100 W/kg of power absorption in the musculature, thereby triggering vigorous blood flow; and 3) the blood flow is more than adequate to maintain safe steady-state temperature within the therapeutic range at or below the pain tolerance level. Perhaps one of the major disadvantages of the inductive shortwave applicator is the nonuniformity of the toroidal-type heating pattern which is awkward to use for treating a small area of tissue.

TABLE V

COMPARISON OF MEASURED AND THEORETICAL SHORTWAVE POWER ABSORPTION IN HUMAN THIGH TISSUE

	Fat	Muscle			
Distance (cm) z	0.6	1.2	3.22	3.48	3.89
Calculated W_a (W/kg)	32.6	126	35	27.2	23.5
Measured W_a (W/kg)	99	102	44	26	20

Note: $z_1 = 1.0$ cm, $h = 1.8$ cm, and $I = 7.54$ A.

V. Microwave Diathermy

When microwave diathermy was first introduced in 1946, there was great hope that it would provide significant improvements in heating patterns over those of the shortwave diathermy described above. The shorter wavelength provided one with the capability to direct and focus the power and couple it to the patient by direct radiation from a compact small-size applicator. This was originally believed to be a distinct improvement over quasi-static and induction field coupling provided by the cumbersome capacitor and coil-type applicators. The cross-sectional area of the directed power could be made smaller and used to provide much more flexibility in controlling the size of the area treated.

Microwave diathermy had been used for a considerable number of years before any quantitative evaluation had been made of the modality. The initial engineering work was done by Schwan [2]–[4], who measured the dielectric properties of human tissues over a wide frequency range (from audio through microwave frequencies) from which much of the data in Table II are derived. Using these results, Schwan theoretically demonstrated the dependence of relative heating in the tissue on the thickness of the skin, the thickness of the subcutaneous fat, and the frequency of a plane wave normally

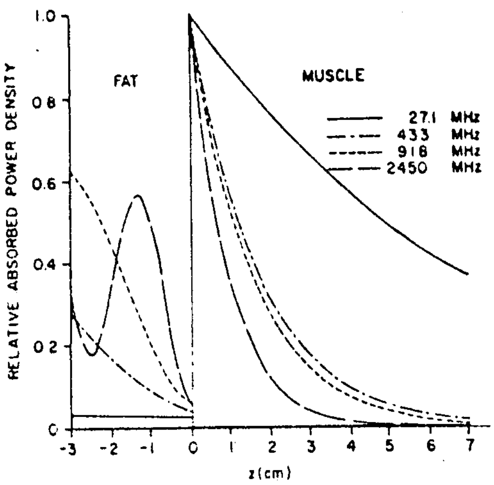

Fig. 20. Relative absorbed power density patterns in plane geometry fat and muscle layers exposed to a plane wave source. From Johnson and Guy [23].

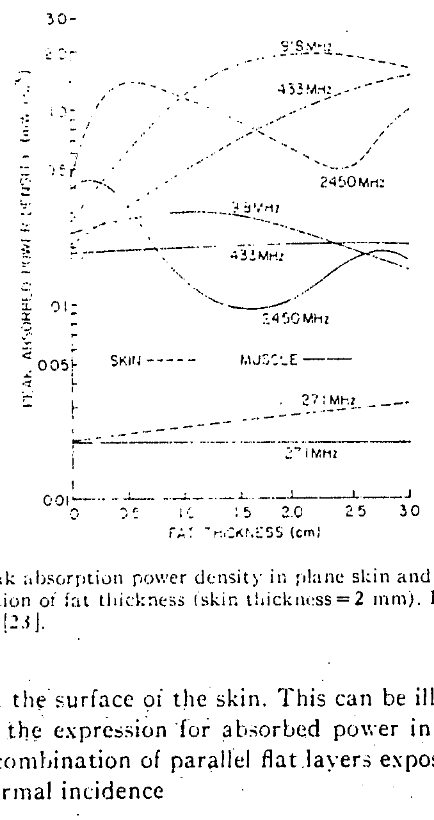

Fig. 21. Peak absorption power density in plane skin and muscle layers as a function of fat thickness (skin thickness = 2 mm). From Johnson and Guy [23].

incident on the surface of the skin. This can be illustrated by evaluating the expression for absorbed power in each tissue layer of a combination of parallel flat layers exposed to plane wave at normal incidence

$$W_{a_i} = \frac{\sigma_i E_i^2}{2\rho_i}\left[e^{-2\alpha_i z_i} + r_i^2 e^{2\alpha_i z_i} 2r_i \cos 2\beta_i z_i + \phi\right] \quad (27)$$

where E_i is the peak magnitude of the field transmitted into the layer, z_i is the distance from the interface with the following layer, $\rho_i = r_i e^{j\phi}$ is the reflection coefficient at the interface with the following layer, and $\beta_i - j\alpha$ is the complex propagation constant of a wave in the tissue layer. The complete absorbed power distributions can be evaluated by (27) for each layer where E_i and ρ_i may be determined from standard transmission line equations as a function of incident power density. The results recently obtained by Johnson and Guy [23] from (27) are illustrated in Figs. 20 and 21. The results show typical power absoprtion characteristics from plane wave irradiation of the tissue for various diathermy frequencies (433 MHz authorized only for European use). Fig. 20 illustrates the results for a wave transmitted through a subcutaneous fat medium into a muscle medium. The absorption is normalized to unity in the muscle at the fat–muscle interface. The relative absorption curves in the fat will remain the same for smaller fat thicknesses (e.g., the portion of the curves between −2 and 0 would correspond to a 2-cm-thick fat layer). Fig. 21 illustrates the absorbed power density in the muscle interface and in a 2-mm-thick skin layer as a function of fat thickness for an incident power intensity of 1 mW/cm². The values may be used to determine the absorbed power at other locations in the muscle and fat by relating them to the curves in Fig. 20. The peak absorbed power density is always maximum in the skin layer for this type of tissue model. The curves illustrate the major deficiencies of the 2450-MHz diathermy as originally demonstrated by Schwan: 1) absorption is so great in the muscle layer that the depth of penetration is only 1.7 cm; 2) the severe discontinuity at the fat–muscle interface produces a large standing wave resulting in a "hot spot" in the fat layer one-quarter wavelength from the muscle surface; and 3) the absorbed power density in the deep tissues varies considerably with fat thickness, making it difficult to predict the proper therapeutic level for different patients having a wide variation of fat thicknesses. The curves indicate, however, that these undesirable conditions may be partially eliminated by using lower frequencies since the depth of penetration will increase and the fat and skin thickness become proportionally smaller compared to a wavelength. The commercially available 2450-MHz diathermy equipment consists of a 0- to 100-W magnetron generator controlled by a variable-power control calibrated in percentage of the total power. Various types of standardized dipole and monopole applicators used with this generator are illustrated in Fig. 22. The radiation power density of the most widely used *C* director is shown in Fig. 23 as a function of distance from the applicator and percentage of power output from the generator. The relation between percent power and actual power delivered to the antenna is shown in Fig. 24. It should be noted that the measurements are of "indicated" power density based on the square of the electric field perpendicular to the direction of propagation along the line of maximum intensity (from *T* feed section of dipole). The measurements were made with distance measured from the protective plastic dipole cover (1.3 cm from dipole). The characteristics of the power density survey meter are such that field components in the direction of propagation and wave impedances different than 120π ohms are not accounted for. Using a meter recently developed by the National Bureau of Standards (EDM-1-C4), one can measure the total field in terms of stored energy and compare it to the field oriented in the direction transverse to propagation as shown in Fig. 25. The results show that the error of neglecting the fields parallel to the direction of propagation can be appreciable for the normal spacing of 5 cm or more used in the clinic. Thus the plane wave analysis will not be completely valid, however, for predicting the absorbed power when the applicator is so close to the tissue since the field and field impedance conditions are considerably different. Though some approximate expressions have been derived to predict absorption in tissue exposed to the near-zone type of field [66], [67], they are not fully applicable to this case. The most expedient method for analyzing the absorption for such sources is by direct measurement through thermographic techniques and phantom models which have been discussed extensively in the

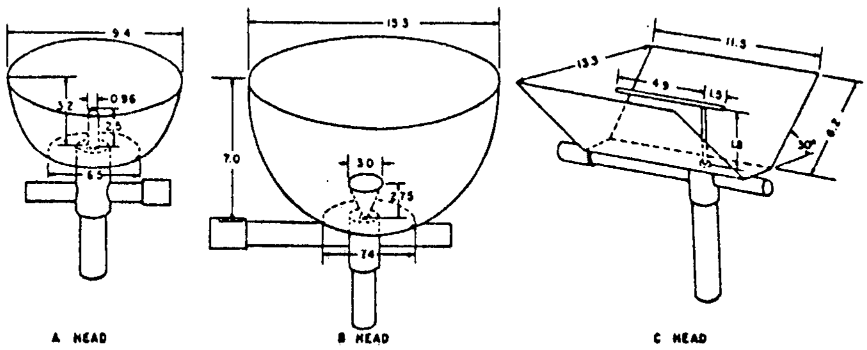

Fig. 22. Applicators used with 2450-MHz diathermy apparatus.

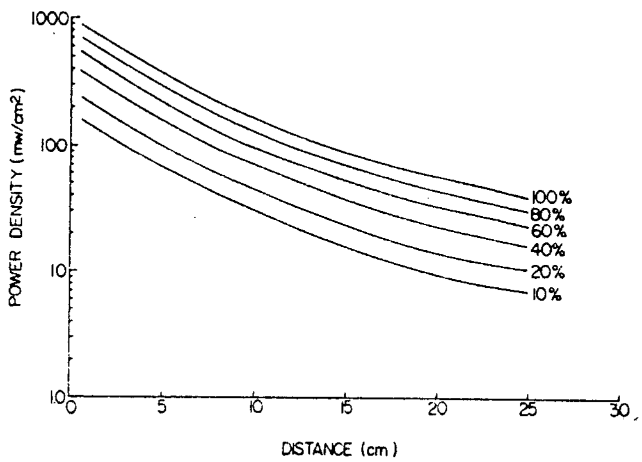

Fig. 23. Measured power density versus distance along axis of maximum intensity for Burdick 2450-MHz C director. (Measurements made with Narda microline electromagnetic radiation monitor Model 8100 with distance measured from dielectric cover 1.3 cm from dipole.)

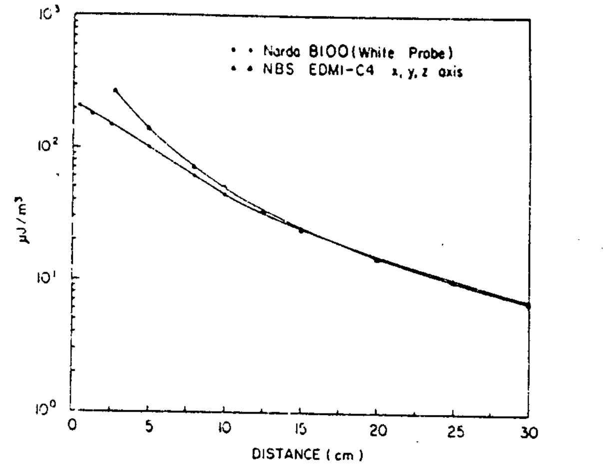

Fig. 25. Electric field energy density versus distance along axis of maximum intensity for Burdick 2450-MHz C director. (Measurement made with NBS electric energy density Meter 1-C4 and Narda 8100 electromagnetic radiation monitor with distance measured from dielectric cover 1.2 cm from dipole.)

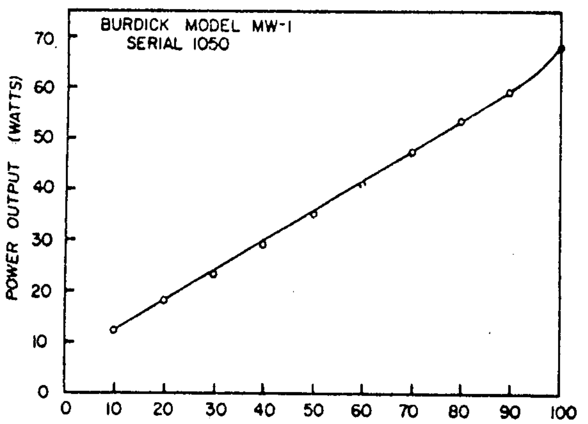

Fig. 24. Power delivered to C director as a function of percentage output for Burdick 2450-MHz diathermy apparatus.

literature [33], [34]. The technique makes use of phantom models which have dielectric properties equivalent to those of actual tissue. These models are exposed for a short time to the EM source under test so that there is an initial linear temperature rise in the model. The model is then quickly disassembled, exposing an internal surface where the temperature distribution can be recorded with a thermograph camera. The information is then converted to absorbed power density by (7). Fig. 26 illustrates the absorption patterns measured in this manner for plane fat-muscle tissue layers exposed to the diathermy C director for different fat thicknesses. The spacing between the applicator (plastic cover) and tissue surface was set to the clinically recommended value of 5 cm. The phantom models used for these studies were assembled by first constructing a 30-cm by 14-cm box with 1/4-in-thick Plexiglas sides and top and bottom surfaces consisting of solid synthetic fat of uniform thickness. The box was then separated into two 30-cm by 15-cm by 14-cm halves, each filled with synthetic muscle. The exposed cut surfaces were covered with a 0.00254-cm-thick polyethylene film to prevent loss of moisture. The models were constructed with fat thicknesses of 1.42, 2.00, 2.47, and 3.6 cm, and muscle thicknesses greater than 10 cm. The experimental data were taken by first exposing the center of the assembled model to the applicator so that the polarization of the electric field was parallel to the plane of separation for the model. The applicator was then energized with sufficient power over a duration of 5 to 60 s, so that the internal temperature rise of the model was sufficient to obtain a thermographic photograph of the plane of separation. The thermograph camera was set to obtain a C scan, displaying a two-dimensional picture of the entire area heated (intensity proportional to temperature) as shown by the large photographs in Fig. 26. The scale on the oscilloscope indicator was set so that one large division is equal to 2 cm. The horizontal midline with the small subdivisions on the photographs corresponds to a line through the geometric center of the applicator and perpendicular to the flat interface of the phantom tissues. The vertical midline with the small subdivisions corresponds to the fat and muscle

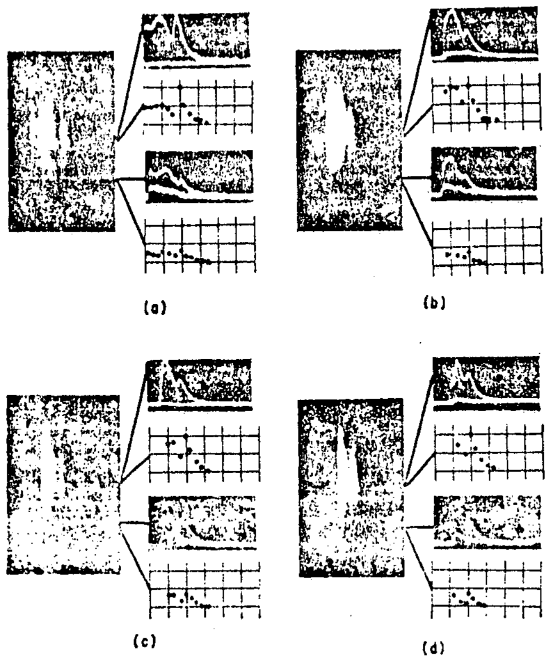

Fig. 26. Thermograms and relative power absorption density patterns in rectangular tissue model exposed to 2450-MHz C director. (Vertical scale B scans, 2.5°C/div at midline for $z_1 = 3.6$ cm and 4 cm below midline for $z_1 = 3.6$, 2.47, 2.0, and 1.12 cm; 5.0°C/div for midline $z_1 = 2.47$, 2.06, and 1.42 cm; input power, 1000 W for 10 s; spacing, 5 cm.) (a) $z_1 = 3.6$ cm. Peak absorption = 2 W/kg per watt input. (b) $z_1 = 2.47$ cm. Peak absorption = 1.84 W/kg per watt input. (c) $z_1 = 2.0$ cm. Peak absorption = 2 W/kg per watt input. (d) $z_1 = 1.42$ cm. Peak absorption = 2.3 W/kg per watt input. · · · · ·, calculated power absorption density.

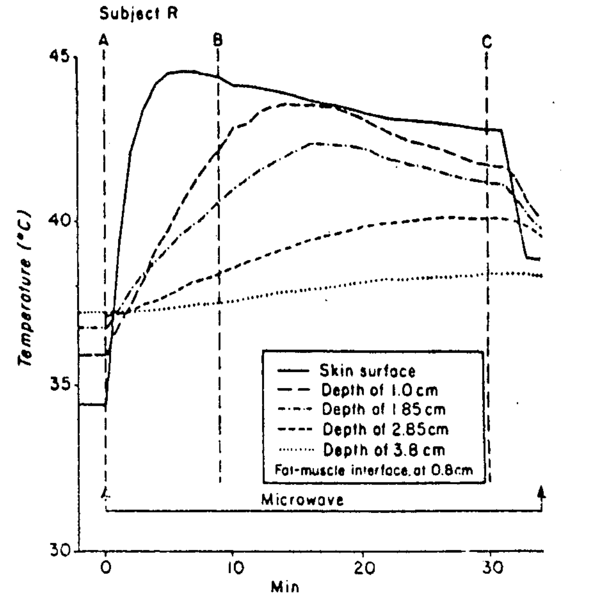

Fig. 27. Temperature recorded in the human thigh during exposure to microwaves at 2456 MHz applied with C director.

TABLE VI

ABSORBED POWER DENSITY AS FUNCTION OF DEPTH IN HUMAN THIGH EXPOSED TO MEASURED POWER DENSITY OF 160 mW/cm²

Depth cm	1.0	1.85	2.85	3.85
C director spaced 2 cm, measured	48	30	12	6
Plane wave, theoretical	22	9.0	2.4	1.2

Note: $z_1 = 0.8$ cm.

interface. After each C scan photo was taken, the model was allowed to cool and the ambient temperature was first recorded by photographing the B scans along the horizontal midline and also along a parallel line 4 cm below the midline of the model. The B scans consisted of one-dimensional scans with the same horizontal scale as the C scans and a vertical deflection proportional to temperature (scale is given under each figure). The model was then reassembled, exposed to the EM source, again disassembled, and B scans were repeated and photographically superimposed on the previously taken B scans. The photographs of the composite B scans and their relation to the C scans are shown at the right of the large photographs in Fig. 26. If the differences in specific heat and density of the synthetic fat and muscle are taken into account by (7), where c_m and ρ_m are the specific heat and density of the synthetic muscle and c_f and ρ_f are the specific heat and density of the synthetic fat, the temperature difference ΔT between the superimposed B scan deflections is approximately proportional to the power absorption density distribution over the region scanned. The accuracy of the estimated specific heat pattern along the midline can be further improved by correcting the error due to heat flow across the interface between the high-temperature muscle and the low-temperature fat observed on all of the thermographs. This can be done in a manner previously described [34] by noting that the power absorption discontinuity at the fat–muscle interface must be proportional to the ratio of electrical conductivities of the two media. The corrected curve is shown by the dotted lines in the figure. The peak absorbed power density per mW/cm² of incident power density normally measured 5 cm from the applicator is shown under each group. The values are substantially greater than that predicted for a plane wave with the same power density as obtained from the graphs in Figs. 20 and 21, using the applicator properties from Figs. 23 and 24. This shows that the power density measurements of the near-zone fields of the C director cannot be compared at all to the same level measured for a plane wave type of source due to multiple reflections between the applicator and the tissue and different impedance conditions. This apparently accounts for the reduced sensitivity of the absorbed power to fat thickness from that of the plane wave source. The experimental data do indicate, however, the problem of excessive heating in the fat due to standing waves and the relatively small depth of penetration into the muscle. Though the results show that the applicator is capable of producing in excess of 130 W/kg maximum absorption at the surface of the muscle, the superficial heating is excessive and it has been clinically demonstrated that tolerance level is dictated by surface heating rather than deep heating. There are no data in the literature on the absorbed power density in actual human tissues exposed to the C director at the 5-cm spacing. This has been done, however, by Lehmann *et al.* [29] for a fat thickness of 0.8 cm and an applicator spacing of 2 cm from the tissue with the results shown in Fig. 27. For this case, the power was adjusted to the point where discomfort or mild pain was felt temporarily at the surface of the skin corresponding to an input power of 17 W (approximately 17-percent output setting on machine) or a measured incident power level of 160 mW/cm². Table VI gives the calculated absorbed power density with depth as compared to that which would be produced by a plane wave with identical measured power density.

The deeper power absorption levels are somewhat lower

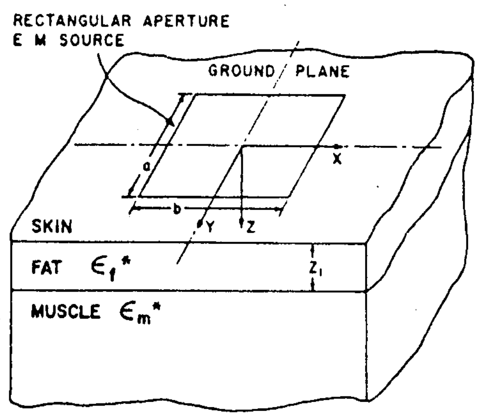

Fig. 28. Aperture source and tissue geometry.

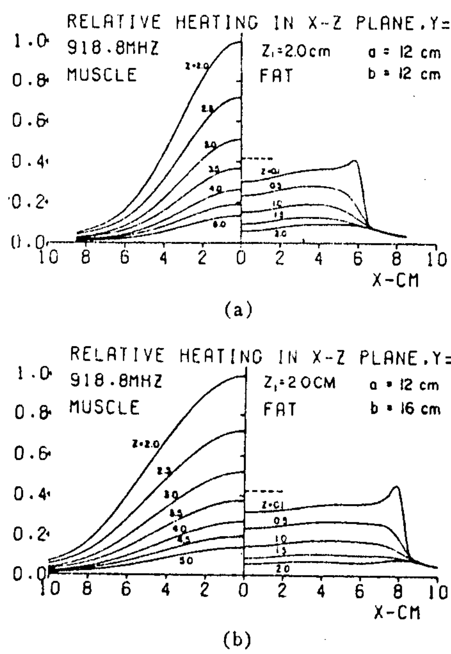

Fig. 29. Relative absorption patterns for 918.8-MHz TE_{10} mode aperture source.

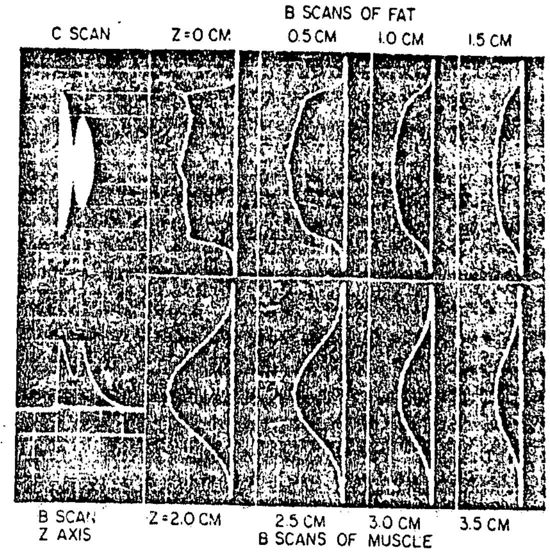

Fig. 30. Thermograms of plane fat and muscle phantom exposed to 13-cm by 13-cm aperture source. Maximum absorption in muscle is 3.27 W/kg per watt input. x-z plane thermograms. $z_1 = 2.00$ cm, $f = 918$ MHz. 13-cm by 13-cm TE_{10} mode source with radome.

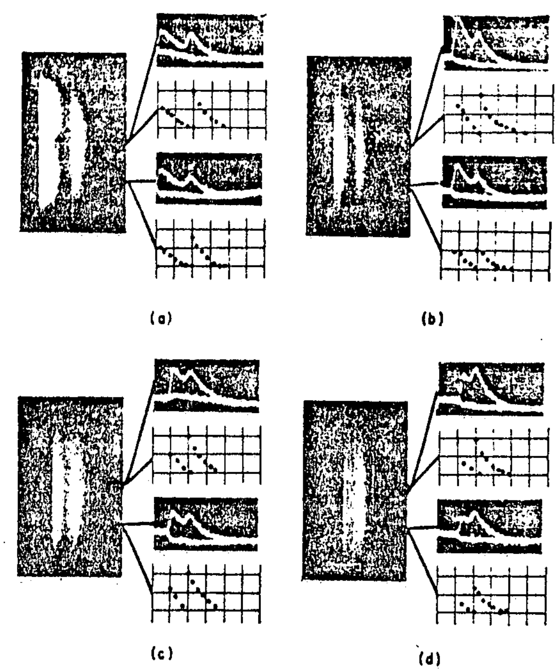

Fig. 31. Thermograms and relative heating patterns of rectangular model exposed to a 12-cm by 16-cm 915-MHz direct-contact aperture. (Vertical scale B scans, 2.5°C/div; input power, 130 W for 40 s.) (a) $Z_1 = 3.6$ cm. Peak absorption = 2.42 W/kg per watt input. (b) $Z_1 = 2.47$ cm. Peak absorption = 2.76 W/kg per watt input. (c) $Z_1 = 2.0$ cm. Peak absorption = 2.42 W/kg per watt input. (d) $Z_1 = 1.42$ cm. Peak absorption = 3.46 W/kg per watt input. · · · · ·, calculated power absorption density.

than that obtained for the shortwave inductor applicator. With all aspects considered, the heating characteristics of the shortwave diathermy discussed in the previous section appear to be superior in terms of therapeutic value to those of the 2450-MHz modality.

VI. 915-MHz Diathermy

The plane wave power absorption characteristics shown in Figs. 20 and 21 clearly indicate the superiority of frequencies lower than 2450 MHz in terms of desirable therapeutic heating characteristics. Plane wave or radiating-type sources at these lower frequencies become impractical to use, however, since the energy is impossible to focus into a beam with reasonable size applicators and the near-zone fields of the applicators extend to greater distances. Under these conditions, a pure radiation or far-zone field can be maintained only by placing the applicator at distances where large areas of the body would be exposed and excessive power levels would be required. Thus, in order to obtain selective heating with reasonable input power levels (50 to 100 W), one must necessarily expose the tissues to the near-zone fields of the source. The induced fields in the tissue are then highly dependent on the source field distribution and frequency and may be considerably different from those induced by a plane wave or radiation field. The aperture source provides a reasonable method for studying the effect of source distribution, size, and frequency on the induced fields in the tissues. With such a source, a degree of control can be exercised over the extent and distribution of the induced fields, resulting in improved diathermy applicators and better control of the absorbed power.

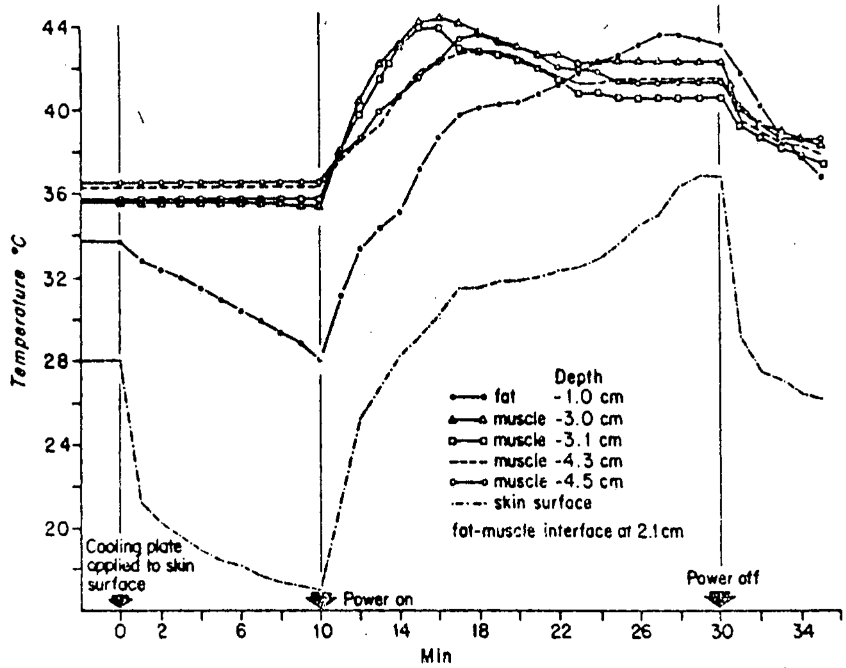

Fig. 32. Temperatures in a human thigh at various depths of tissue resulting from treatment with a 915-MHz direct-contact aperture source with surface cooling. From B. J. DeLateur et al. [32].

Guy [33] has analyzed this case for a planar fat and muscle tissue geometry as shown in Fig. 28. The results showed that the optimum frequency for therapeutic heating with a clinical size 12-cm by 16-cm applicator was 750 MHz, but the results that can be obtained at the authorized ISM frequency of 915 MHz were nearly as good [68]. Fig. 29 illustrates the calculated power absorption or relative heating patterns for a 12-cm by 16-cm and a 12-cm by 12-cm aperture with a TE_{10} mode waveguide electric field aperture distribution. The format of the figure is similar to that of Fig. 14 for the shortwave applicator with half of the symmetrical heating patterns shown for each tissue. The dashed line above the fat heating curves illustrates the level of maximum fat heating due to a plane wave incident on the tissue. The penetration characteristics into the muscle are identical to those of a plane wave. Guy's studies indicated that a 13-cm by 13-cm square aperture with a TE_{10} waveguide mode source distribution should give optimal results for the maximum muscle-to-fat heating ratio with the minimum size source at 915 MHz. Fig. 30 illustrates a set of thermograms taken of the plane fat and muscle phantom model exposed to the aperture. A 3-mm-thick radome cover was placed over the aperture to protect the tissue from localized heating due to the edge of the applicator. The results show that the maximum absorption in the muscle is 3.27 W/kg per watt input to the applicator. Taking density and specific heat into account, the maximum heating in the fat was approximately 40 percent of this value. Again, the maximum absorption density per unit of incident power density is greater than that for a plane wave. This is expected since the applicator was designed to couple all of the transmitted energy to the tissue, whereas, with a plane wave source, a considerable amount is reflected from the surface. The penetration and minimal fat heating characteristics compare favorably with those of the shortwave diathermy, with the additional advantage that the heating pattern is reasonably uniform in contrast to the undesirable toroidal pattern of the latter. Fig. 31 illustrates the experimental measurements of the absorption patterns made in phantom models exposed to a larger 12-cm by 16-cm rectangular aperture direct-contact source of the type discussed in other work [31], [66].

One of the disadvantages of an applicator such as discussed above when it is applied in direct contact with the tissue surface is that it prevents surface cooling normally present due to convection and evaporation. Since the skin normally experiences the greatest power absorption (Fig. 21), excessive skin and fat temperatures could result if some means of cooling is not applied. The 12-cm by 16-cm applicator discussed above was modified by placing a dielectric plate with a circulating dielectric coolant fluid contained within it between the aperture and the tissue to be treated. DeLateur et al. [32] exposed the thighs of human volunteers with such an applicator. In the experiments, coolant was continually circulated through the cooling plate. After a period of 10 min of cooling, 55 W of 915-MHz power was applied for 20 min to the 12-cm by 16-cm aperture applicator. This corresponded to a maximum power density of 573 mW/cm² at the center of the applicator. Fig. 32 shows the results of a typical experiment for a person with 2 cm of subcutaneous fat. The initial 10 min of cooling lowered the skin temperature below 18°C and, to a lesser degree, the fat temperature to 28°C. The muscle tissue was unaffected. When the power was applied, the usual linear transient was observed from which an absorbed power density of 171 W/kg was calculated for a 1-cm depth in the muscle and 75 W/kg was calculated for a 3-cm depth in the muscle. A power density of 96 W/kg was calculated for the center of the fat layer. During the 20-min heating period, the muscle temperature versus time curves followed the characteristic trend. When the temperature reached 44°C, an increase in blood cooling occurred resulting in a decrease in tissue temperature. It may be noted, however, that all sites monitored in the muscle attained temperatures in the therapeutic range above 40°C. It is interesting to note for this case that the blood cooling was sufficient to stabilize

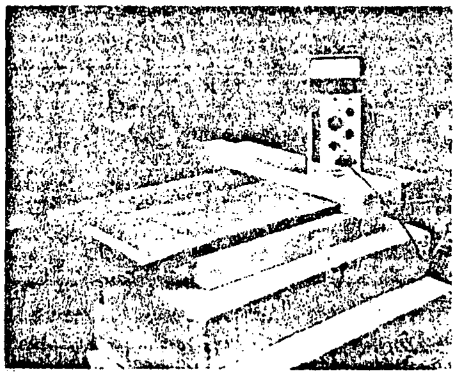

Fig. 33. Photograph of a one-half cross section of a phantom model of a human back exposed to a diathermy applicator. (Probe for measuring leakage fields is shown.)

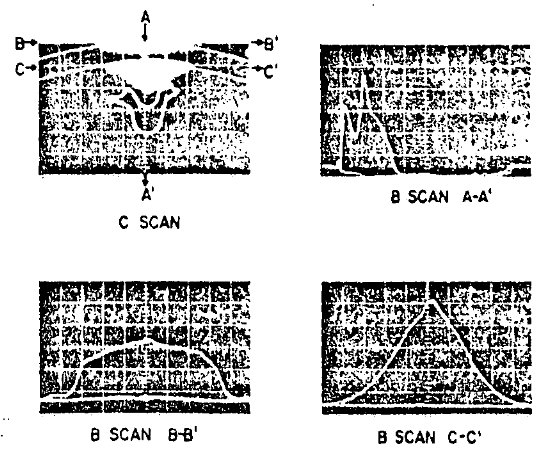

Fig. 34. Thermogram recordings of phantom back exposed to 2450-MHz *C* director. (Peak power absorption density 2.39 W/kg per watt input at surface of muscle, scale 1 div = 2 cm.)

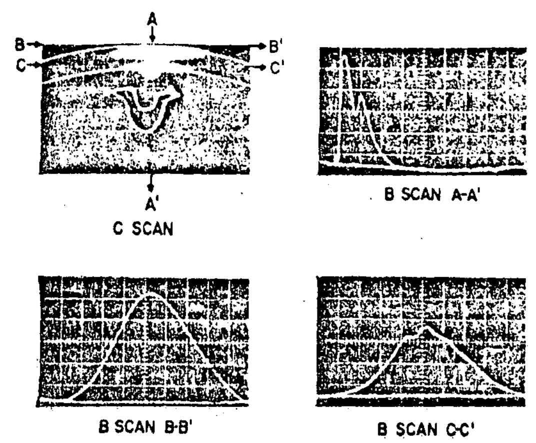

Fig. 35. Thermogram recordings of phantom back exposed to 915-MHz 13-cm by 13-cm aperture source with 1-W input. (Peak power absorption density 3.94 W/kg per watt input at surface of muscle, scale 1 div = 2 cm.)

the temperature at safe levels with the high local and absorbed transmitted power densities.

These data indicate that the 915-MHz modality is superior to the 2450-MHz modality and compares very well with the shortwave inductor modality. In fact, the more uniform pattern of the former would be more desirable for many applications where selective heating of a restricted volume of tissue is desired. Additional comparisons between the absorption patterns produced by various applicators operating at frequencies in the microwave range are given in [23] and [33].

VII. Safety Considerations

The finite direct-contact aperture is not only advantageous to use for the reasons described, but it also allows more efficient application and better control of the energy imparted to the tissues, thereby eliminating unwanted high-level and possibly unsafe side radiation originating from the radiating-type applicators. It should be pointed out that existing commercial diathermy applicators are not high-gain antennas and have broad radiation patterns. The therapeutic effectiveness and leakage radiation of both the 2450-MHz *C* director and the 915 direct-contact applicator were determined while a phantom model of a human back was irradiated with each applicator in an anechoic chamber. The full-scale model shown in Fig. 33 consists of synthetic muscle tissue containing synthetic vertebrae and covered with 2 cm of synthetic subcutaneous fat. Figs. 34 and 35 illustrate the thermographic recordings of the absorbed power distribution in the phantom back for the 2450-MHz *C* director, and the 13-cm by 13-cm 915-MHz direct-contact applicator. Figs. 36 and 37 illustrate the power density in the vicinity of the applicator and the model for the two frequencies as measured with a Narda Model 8100 radiation monitor in the plane parallel to the plane of electric field polarization and perpendicular to the surface of the model. All values for internal power absorption and external power density are normalized for 1-W input to the applicator.

The data indicate that an input of 100 W to the 2450-MHz *C* director would produce a maximum of 239 W/kg absorption at the surface of the muscle and 187 W/kg at the surface of the fat, while at the same time producing a maximum radiation level of 10 mW/cm² at a distance of 25 cm from the end of the applicator at an angle of 45° from the surface of the model. On the other hand, an input of 61 W to the 915-MHz direct-contact aperture source would produce the same maximum 239 W/kg power absorption at the surface of the muscle, but the maximum absorption in the fat would be reduced to 77 W/kg and the outside radiation level would be below 10 mW/cm² at a distance greater than 8 cm from the edge of the applicator. The superiority of the aperture source over the *C* director, both in terms of therapeutic effectiveness and radiation safety, is quite apparent.

VIII. Nonthermal Therapeutic Applications

Therapeutic applications of microwave energy are not limited to the heating of tissues. Microwaves can provide a means of transmission of power into tissues for a variety of therapeutic applications. Once transmitted deeply into the tissues, the power can be converted by standard means to other useful forms of energy by negligibly small transducers. For instance, microwave diodes small enough to be implanted with a hypodermic needle can be used to convert microwave power into a dc current for various therapeutic applications. It has been demonstrated that small dc currents can be used to control the location and the rate of tissue healing or growth [69]–[71]. Implanted microwave diodes could provide a "wireless" method for achieving this. Pulsed micro-

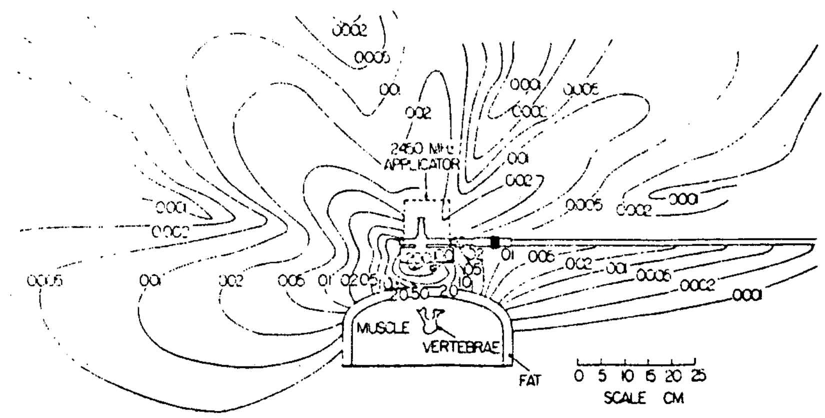

Fig. 36. Leakage radiation (mW/cm²) from phantom model of human back exposed to 2450-MHz C director (1-W input).

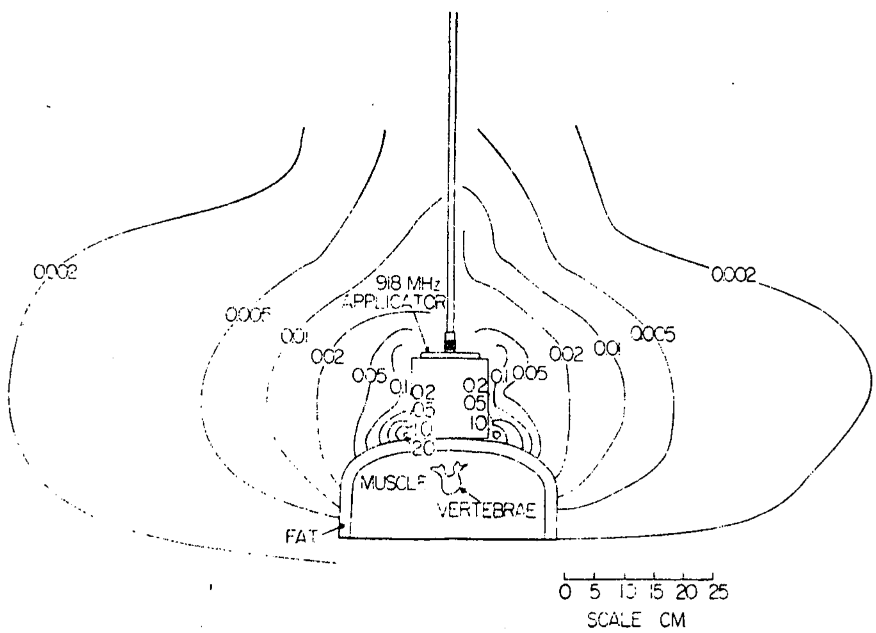

Fig. 37. Leakage radiation (mW/cm²) from phantom model of human back exposed to 918-MHz 13-cm by 13-cm square aperture TE_{10} mode source (1-W input).

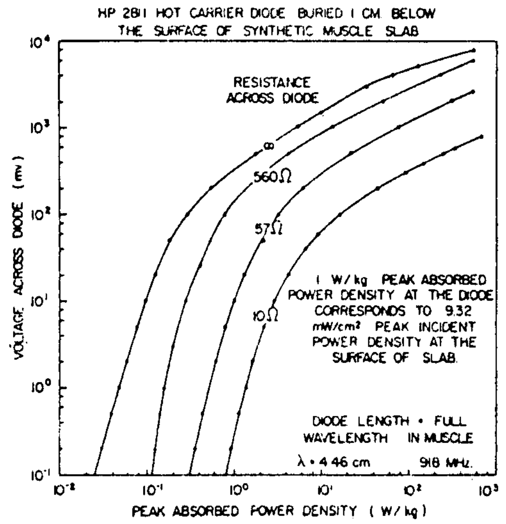

Fig. 38. Voltage output characteristics of diode implanted in muscle tissue irradiated with 915-MHz power.

wave power could also be used in the same fashion to stimulate nerves for control of pain or neuromuscular function, or for stimulating cardiac muscle in pacemaker applications. Transcutaneous low-frequency RF power transmission has been used for applications such as these in the past [72], but it appears that a greater degree of miniaturization is possible with microwaves. Fig. 38 illustrates the characteristics of a microwave diode implanted in the same synthetic muscle tissue that was used for diathermy studies. The leads of the diode can serve the dual purpose of acting as a receiving antenna and also as electrodes for the application of dc or pulsed voltage to the tissue. Additional electrodes may be attached to provide better contact with either local or more remote tissue. The curves illustrate the relationship between absorbed microwave power in the tissue surrounding the diode to the dc voltage delivered across various resistive loads. The required stimulating voltage for eliciting a threshold response is a function of the pulsewidth as shown in Fig. 39 for the sciatic nerve of the frog. The electrode resistance for this case

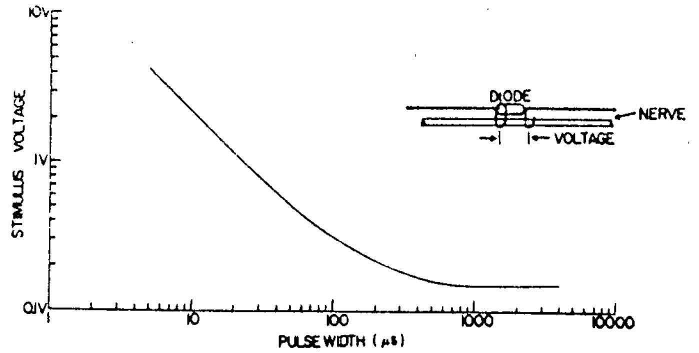

Fig. 39. Threshold voltage as a function of pulsewidth for stimulating nerve tissue with microwave diode. Pulsewidth is plotted versus minimum stimulus voltage across frog sciatic nerve.

was approximately 50 kΩ. Minimum energy is required for a pulse approximately 200 μs in width and 0.2 V in magnitude. Thus, from Fig. 38, a peak absorbed power density greater than 0.5 W/kg would be needed to stimulate the nerve. Thus a typical stimulation rate of 30 pulses per second would require an average absorbed power density of only 0.0003 W/kg. It can be seen (Fig. 34) that this could be delivered by a 13-cm by 13-cm surface aperture through a 2-cm layer of fat, 2 cm deep into the muscle, with a peak input power of only 0.26 W or an average power of only 1.56 mW. The average surface-power flux density corresponding to this would only be 9 μW/cm². It would be no problem at all to produce power levels far in excess of this with small portable solid-state microwave generators. Printed-circuit antenna array concepts would be used to produce a wide range of possible low-profile surface antenna configurations to satisfy various clinical requirements.

IX. Conclusions

Effective therapeutic heating of tissues below the skin and subcutaneous fat layer of patients with EM fields and currents requires a choice of frequency, applicator, and input power so that the temperature of the deeper tissue can be raised to the maximum level of 44° to 45°C within a 5- to 15-min period. The duration to maximum temperature can be controlled by setting the input power level. Just before, or when the temperature reaches this maximum range, vasodilatation will produce a marked increase in blood flow which will limit the rise in temperature in tissues with good vascularity and produce a decrease in temperature by several degrees. A total exposure period of 20 to 30 min is generally required to produce the optimum therapeutic benefits, and during this time the temperature of the superficial tissues must be kept below that of the deeper tissues being heated. It can be shown theoretically that shortwave applicators designed to couple energy to the tissue capacitively cannot satisfy the above requirements for tissues covered with subcutaneous fat (an exception may be pelvic diathermy). Both theoretical studies and experimental measurements show that the shortwave induction coil type of applicator of proper design will provide the maximum deep heating with acceptable surface and subcutaneous fat heating. The toroidal heating pattern produced by these applicators may present some problems, however, in producing localized heating and may produce uneven heating in unwanted areas unless the therapist is familiar with heating patterns produced by this applicator.

It can be shown both theoretically and experimentally that microwave applicators operating at a frequency of 918 MHz are far superior to those operating at a frequency of 2450 MHz. If they are compared with the shortwave induction-type applicators, they are capable of heating better and more uniformly in the musculature with minimal surface and subcutaneous fat heating, whereas the shortwave induction applicator would produce the highest temperature in the subcutaneous tissue and superficial musculature, thus allowing the therapist a choice of the area he wants to heat. The more uniform power distribution of the 918-MHz microwave applicator is clinically more desirable than the toroidal pattern of the shortwave applicator. Experimental studies show that an absorbed EM power density of 50 to 170 W/kg is required to produce the necessary therapeutic heating levels.

EM energy may also be used to stimulate nerves in deep tissues by small implanted diode rectifiers with absorbed power density levels as small as 0.0003 W/kg, thus reducing the size of the surgical receiver implant, provided a suitable outside power source can be developed.

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