Infrared Thermographic Measurement of the SAR Patterns of Interstitial Hyperthermia Applicators

by

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A thesis submitted in conformity with the requirements for the degree of Master of Applied Science
Graduate department of Electrical and Computer Engineering
The University of Toronto

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In the design and quality assurance of hyperthermia applicators, much work has been done in the development of accurate systems for the measurement of specific absorption rate (SAR). The need for higher spatial resolution in SAR measurement systems designed for interstitial applicators has resulted in growing interest in infrared (IR) thermographic measurement. Although IR thermographic measurement has been argued against because of the presence of thermal conduction and convection artifacts in measured SAR patterns, such effects are typically assumed to be negligible, and IR measurement systems continue to see routine use.

In this thesis, the major sources of error and variability in an interstitial IR thermographic SAR measurement system were investigated to determine the accuracy of measured SAR patterns. Conduction and convection artifacts were found to be a considerable source of error in the measured SAR patterns of interstitial microwave and ultrasound applicators for heating durations in excess of 10 to 20 seconds.
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Chapter 1

Background

1.1 Hyperthermia: History

The history of heat as a therapeutic agent is long and varied [21][46]. Hippocrates recorded the use of red-hot irons to treat small, non-ulcerating cancers in 400 B.C. Modern interest in hyperthermia (from the Greek hyper, meaning excess, and therme, meaning heat) began with W.C. Coley, a New York surgeon, who used bacterial pyrogens to treat cancer patients in 1893 [12]. In 1909, W.E. Schmidt first suggested the use of hyperthermia as a radiosensitising agent [43]. In 1941, Mittlemann et al. identified the need for careful dosimetry in measuring temperatures and relating them to energy absorption [35]. Interest in hyperthermia, which had waned with the advancement of X-ray treatments, was revived in 1967, when it was first suggested that cancerous cells were more sensitive to heat than normal cells [7]. Although the specificity of tumour cells to heat has since been questioned, hyperthermia continues to generate considerable interest as an experimental cancer therapy.

1.1.1 Biological Rationale

At temperatures above 42–43°C, cells exhibit a time-temperature dependent cytotoxicity such that for each increase of 1°C, exposure time can be reduced by approximately one-half to induce equivalent cytotoxicity [40]. Although the kinetics of the killing process have been well established, the precise molecular mechanism of thermal toxicity is not well understood, and may be due to a number of factors including membrane damage and nuclear protein denaturation [48]. There are several environmental factors that enhance the effect of hyperthermic cytotoxicity, such as lowering pH or reducing certain nutrient
levels [24]. Hyperthermia has also been shown to be a sensitiser of cells to radiation, especially when the two agents are administered simultaneously [14].

1.1.2 Clinical Rationale

No large-scale clinical studies have been able to validate the efficacy of hyperthermia as an independent therapeutic agent, probably due to incomplete heating of the tumour volume [49]. Clinical trials of hyperthermia combined with radiotherapy, or thermoradiotherapy, have produced mixed results. A Phase III study conducted by the North American Radiation Therapy Oncology Group (RTOG) in the late 1980s found no statistically significant difference in the rate of complete response to the treatment of superficial tumours by radiotherapy and thermoradiotherapy (30% for radiotherapy alone, compared to 32% for thermoradiotherapy) [38]. A more recent study by the same group, comparing radiotherapy and thermoradiotherapy treatment of recurrent, deep-seated tumour sites, produced similar results (53% for radiotherapy alone compared to 55% for thermoradiotherapy) [19]. In both studies, however, the authors concluded that insufficient temperature levels had been reached in the tumour volume due to inadequate heating applicators.

Phase III clinical trials by other groups have been more favourable. The European Society of Hyperthermic Oncology performed a study of 70 patients with recurrent melanomas, reporting complete response in 35% of patients treated by radiotherapy alone, compared to 62% for patients treated with thermoradiotherapy [37]. A collaborative study between the Dutch Hyperthermia Group, the Medical Research Council (England), the European Society of Hyperthermic Oncology, and the Princess Margaret Hospital (Canada) of patients with advanced primary or recurrent breast cancer, produced similarly positive results (41% complete response ratio for radiotherapy treatment, compared to 59% for thermoradiotherapy) [51].

Several of the above clinical studies reported difficulties in heating the entire tumour volume using current heating applicator technology, which establishes the need for additional research into applicator design and performance evaluation.

1.2 Thermal Delivery

There are many ways to heat tissue and the choice of a particular delivery modality is dependent on many factors, including tumour shape and size, and the geometry of the surrounding tissue. Hyperthermia in tissue is induced by the deposition of energy, which can
be delivered to the region of the tumour optically, using a laser, acoustically, using an ultrasound transducer, or electromagnetically, using a variety of different techniques including ferromagnetic induction, resistive radiofrequency heating, and microwave radiation.

Hyperthermia applicators are traditionally classified as either a) external, applied from outside the body, b) intracavitary, applied from within a body cavity such as the rectum or oesophagus, or c) interstitial, embedded directly into the tumour mass. External applicators are typically used in the treatment of superficial tumours, while intracavitary and interstitial applicators allow for hyperthermic treatment of deep-seated tumours which are usually inoperable and otherwise difficult to treat.

The goal of hyperthermic treatment is the necrosis of the entire tumour volume. Failure to do so could result in recurrence, due to the proliferation of cells that received a sub-lethal thermal dose. The ability of a heat source to deliver energy that will penetrate deeply into tissue is, therefore, of prime importance in hyperthermia. Due to their superior penetrative abilities, the two most commonly used types of hyperthermia systems employ either ultrasonic or microwave heating devices.

1.2.1 Ultrasonic Hyperthermia Applicators

Ultrasonic devices produce mechanical waves in tissue which cause heating. Ultrasound applicators for hyperthermia operate in a frequency range from 0.5 to 10 MHz, yielding penetration depths of up to 15 cm in tissue at the low end of the frequency spectrum. Penetration depth in tissue is frequency dependent because of the frequency dependence of the ultrasonic absorption properties of tissue. Ultrasound is of particular use in the treatment of deep-seated, inoperable tumours because the ultrasonic beam can be focused at a specific depth below the surface of the tissue, so that energy can be delivered to deep-seated tumours with minimal absorption outside the tumour volume.

One of the primary disadvantages of focused ultrasound is the beam deflection at tissue-bone and soft tissue-gas interfaces. Beam deflection results in an increased temperature elevation at bone and tissue surfaces, which can be four to six times greater than that observed in resting muscle tissue, resulting in patient discomfort and, possibly, burning [28] [27]. This poses a considerable limitation on where ultrasound can be used and requires special care in temperature monitoring during treatment, to avoid the presence of these 'hot spots'.

The recent advent of interstitial ultrasound applicators has generated new interest in the use of ultrasound for minimally-invasive hyperthermia. Interstitial ultrasound ap-
Applicators can be inserted directly into the tumour mass and deliver their heat locally. Since the acoustic waves propagate only in a small, localised region, the danger of deflection at soft tissue interfaces is significantly reduced. Several such applicators are currently under development, but have yet to be tested in clinical trials, although initial experiments have yielded favourable results [15].

1.2.2 Microwave Hyperthermia Applicators

Microwave hyperthermia devices produce electromagnetic (EM) waves which carry energy into tissue. The rate of energy deposition is quite high, which is particularly important in hyperthermia treatments, because if there is significant delay in energy propagation the energy may not reach the desired site before it is conducted away by the blood.

At microwave frequencies, the electromagnetic properties of tissue are determined primarily by the behaviour of water molecules within the tissue. Generally, the higher the water content of a tissue, the higher the dissipation of energy. Hence, fat tissue, which has a low water content, is heated poorly compared to muscle tissue, which has a higher water content. This is an important consideration in hyperthermic treatment because, unlike capacitive heaters, microwave radiators can deliver energy to a tumour situated below a subcutaneous layer of fat without significantly heating the fat layer.

International regulations established by the Federal Communications Commission restrict the range of microwave frequencies for medical use to the Industrial Scientific Medical (ISM) band, limiting operational frequencies internationally to 433 MHz, 915 MHz and 2450 MHz. Microwave radiation can be delivered externally, using either a single radiator or an array of radiators, or internally, using short, radiating antennas in either single mode or array configuration.

External Applicators

External microwave applicators have been constructed using many different designs. The most common types of applicators for superficial treatment are waveguide, microstrip, and ring applicators, the operational principles of which are described in detail in the literature [22]. Treatment of deep-seated tumours requires some method of focusing the EM field in order to increase the penetration depth of the microwaves in tissue. Phased arrays, which create limited focusing by constructive and destructive interference, have performed favourably in pre-clinical trials [16].
Interstitial Microwave Antennas

Interstitial microwave applicators are made from small antennas much like those used in communication. The most elementary type of interstitial microwave antenna is the dipole antenna, which consists of a flexible miniature coaxial cable with an extension of the inner conductor at the antenna junction. Electromagnetic energy from the antenna is absorbed by the tissue, resulting in heat generation. The highest absorption occurs at the antenna surface near the junction and decreases radially and axially outward, resulting in an elliptically shaped heating pattern.

A major drawback of the dipole antenna is that the maximum temperature rise occurs at the junction, so that it may be necessary to insert the antenna beyond the tumour mass in order to ensure sufficient heating of the entire tumour volume. This not only increases the invasiveness of the procedure, but is not always possible, depending on the location of the tumour. Another drawback of the dipole design is the dependence of the antenna heating pattern on the insertion depth of the antenna into the tissue [50], due to the extreme differences between the electrical properties of air and tissue. The location of maximum heating, referred to here as the antenna hot spot, shifts with respect to the antenna junction as the insertion depth is varied. This is an undesirable property because it complicates treatment planning and limits the positioning of the applicator.

Modified dipole and other antenna designs have been proposed to eliminate some of these drawbacks. A variation on the dipole antenna, called the hot-tip antenna, biases the antenna junction towards the applicator tip, resulting in improved heating near the antenna tip. Other modified dipole designs include multi-node antennas, which bias heating away from the antenna tip, and multisection antennas, which produce more heating near the antenna tip.

A more recent design, called the helical coil antenna, consists of a basic dipole design with a bare wire winding around the insulation of the inner conductor. Helical coil antennas have two primary advantages over dipole designs. 1) The antenna hot spot is located closer to the tip of the antenna, rather than at the junction, so over-insertion of the applicator past the treatment volume is less likely to be necessary, and (2) helical coil antenna heating patterns are independent of insertion depth [41].

A general drawback of interstitial microwave applicators is the small size of the heating pattern, limiting the size of the treatment volume. Interstitial antenna arrays, which employ several applicators simultaneously\(^1\), can be used to treat larger volumes.

\(^1\)typically four applicators in a rectangular array
although the invasiveness of the procedure is increased.

1.3 Heat Transfer

When a power source is applied to an absorbing medium, the rate of energy transfer to the medium can be described by a spatially-distributed function $W(x, y, z)$. Usually the energy transferred to the medium will be converted to heat, the exact method of energy transfer being dependent on the modality of the power source. For a power source in an infinite homogeneous medium of heat capacity $c$ and density $\rho$, the initial temperature rise due to the conversion of deposited energy to heat is

$$dT = \frac{1}{\rho c} W dt$$  \hspace{1cm} (1.1)

1.3.1 Thermal Conduction

When a thermal gradient exists within a medium, heat is transferred by physical interaction between particles with different temperatures, i.e. different kinetic energies. In a solid, thermal conduction results from the movement of free electrons and vibrational energy in the material. The rate of heat transfer is related to the thermal gradient by the thermal conductivity, $k$. In an isotropic medium, the temperature rise associated with deposition of energy, described by equation 1.1, results in a heat flow of $-k \nabla T$, for $k$ independent of temperature. The temperature rate of increase is then described by

$$\rho c \frac{\partial T}{\partial t} = W + k \nabla^2 T$$  \hspace{1cm} (1.2)

which is the familiar Heat Conduction Equation.

1.3.2 Thermal Convection

When the surface of a medium of uniform temperature $T_S$ is exposed to a moving fluid of temperature $T_F$, heat is exchanged between the surface and the fluid. Since the mechanism of heat transfer is not dependent on the interaction between adjacent particles, the rate of convective heat transfer is proportional to the temperature differential, rather than to the temperature gradient. For a surface of uniform temperature $T_S$ and a fluid of temperature $T_F$, the resulting heat flow per unit area can be expressed as

$$q_c = \hat{h}(T_S - T_F)$$  \hspace{1cm} (1.3)
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where \( \bar{h} \) is the mean coefficient of heat transfer, which is dependent on the both the flow rate and physical properties of the fluid. Equation 1.3 is commonly referred to as the *Newton law of cooling*.

Although, forced convection of heat by blood flow plays a considerable role in heat transfer within perfused biological tissue, it is often useful to design simple, non-perfused models in order to study basic heat transfer principles within tissue. For such a non-perfused medium, convective heat flow occurs only at boundary interfaces that are exposed to fluids such as air or water, and can be ignored for a material of near-infinite dimension.

1.4 Specific Absorption Rate

1.4.1 Definition

Specific absorption rate, or SAR, is defined as the rate of energy \( E \) absorbed or dissipated in an incremental mass contained in an incremental volume \( V \) of a given density \( \rho \). Mathematically, SAR can be expressed as

\[
SAR = \frac{d}{dt} \left( \frac{dE}{\rho dV} \right)
\]

where the units of SAR are given in \( \text{W/kg} \). Scaling equation 1.4 by a factor of \( \rho \) results in a more convenient unit of measurement (\( \text{W/cm}^3 \)) and provides consistency with the energy deposition term \( W \) of equation 1.2. This definition of SAR, in terms of absorbed power per unit volume, is usually referred to as the *Absorption Rate Density* (ARD).

The spatially distributed SAR pattern, \( W(x, y, z) \), also called the heating pattern or power deposition pattern, is a useful descriptor in hyperthermia because it identifies both the magnitude and distribution of energy deposited in the tissue, which can be used to predict the resulting temperature rise within tissue. Determination of the SAR pattern is also an important measurement in the design of new applicators and in the quality assurance of hyperthermia devices and treatments.

1.4.2 Clinical Significance

It is important to distinguish between the power deposition pattern of a hyperthermia applicator, i.e. the spatial distribution of applied power, and the temperature distribution in the tissue that results from that power deposition. The heating pattern, which is synonymous with the SAR pattern, represents the spatial distribution of the heat initially
generated in the tissue by energy deposition, which is linearly related to the energy absorbed by the tissue, according to equation 1.1. Thermal diffusion and forced convection due to blood flow can significantly alter the shape of the heating pattern over time, resulting in a temperature distribution in the tissue that may differ significantly from the SAR pattern. Despite this, SAR remains an important quantity in hyperthermia quality assurance and treatment planning for several reasons. SAR is used as a criterion for the characterisation and evaluation of applicator design, and as an input for thermal models of complex tissue geometries.

Applicator Characterisation

The selection of an appropriate applicator for the treatment of a specific tumour is dependent on many factors including the shape, size, and location of the tumour mass. Characterisation of hyperthermia applicators allows one to determine whether the applicator will be sufficient to treat the tumour volume. Evaluation of new applicator designs requires a criterion for comparison that can be applied equally to different designs. Since the goal of hyperthermia is the destruction of tumour tissue with maximal preservation of normal tissue, the ideal measure of applicator performance would be the therapeutic volume, the volume of tissue raised to a temperature sufficient to induce hyperthermic necrosis (42–45°C). SAR is a more convenient parameter because it is easily measured, unlike the therapeutic volume, which must be estimated using an appropriate thermal model.

Applicator SAR patterns are used not only for design evaluation, but also for performance evaluation. For quality assurance purposes, it is necessary to evaluate the performance of hyperthermia applicators on a regular basis, because they can be easily damaged during routine use [52]. If, for example, an interstitial microwave antenna is slightly bent, its heating pattern may differ significantly from the expected pattern, resulting in the potential delivery of heat to regions outside the treatment area. The North American Radiation Therapy Oncology Group (RTOG) quality assurance guidelines for interstitial hyperthermia require that the SAR pattern of microwave antennas be checked at the intended insertion depth prior to each hyperthermia treatment [20].

Thermal Modeling

In the planning of hyperthermia treatments, it is desirable to predict in advance how heat will be deposited within the patient, both for safety considerations (e.g. to avoid excessive
heating), and to increase the likelihood of successful treatment by devising patient-specific treatment protocols. The application of powerful computational methods such as finite difference and finite element analysis has resulted in considerable advancement in the thermal modeling of complex tissue geometries [31].

The two primary models of heat flow in tissue are based on the bioheat transfer equation (BHTE), which incorporates a convective term into the Heat Conduction Equation to model blood flow as a heat sink that removes heat from one location and deposits it in another, and the effective thermal conductivity equation (ETCE), which models both conduction and convective blood flow as local variations in the thermal conductivity of the tissue. For both models, it is necessary to have an accurate measure of the initial energy deposited in the tissue by the heat source in order to model the resulting temperature distribution.

1.5 Phantom Studies

The specific absorption rate pattern of an applicator is highly dependent on the properties and geometry of the absorbing medium. Consequently, the measurement of SAR should ideally be conducted in a medium that exactly reproduces the properties and geometry of biological tissue. Using actual tissue may not be feasible, depending on the SAR measurement system. In addition, the SAR patterns produced would be relevant only for that particular sample of tissue. Also, reproducibility between experiments may be poor with actual tissue, because of the difficulty in obtaining identical tissue samples. Consequently, tissue-equivalent phantom materials are used. Although the SAR patterns produced in such materials do not necessarily reflect the actual power deposition in tissue, they do provide characteristic information about a hyperthermia applicator. Hence, experimental measurement of specific absorption rate using a phantom material is an important first step in the characterisation of hyperthermia applicators and in the evaluation of theoretical hyperthermia treatment models.

A phantom material for a particular hyperthermia modality need only reproduce the physical characteristics of tissue associated with that modality. An ultrasound phantom, for example, should have the same acoustic properties as the target tissue, while an electromagnetic phantom should have the same electrical properties. In both cases, however, the thermal properties of the tissue must be reproduced as well. Since these properties vary in different tissue types [23], phantom materials are generally both modality-specific and tissue-specific, ie. they are designed to mimic the properties of one type of tissue when
heated by a specific modality.

The choice of an appropriate phantom material for the measurement of SAR is dependent not only on the modality of the applicator, but also on the method of measurement. The physical state of the phantom, i.e. liquid, solid, or semi-liquid, depends primarily on the measurement technique. Infrared thermographic SAR measurement, for example, requires a multilayer phantom configuration, so a solid phantom material is preferable. Other desirable phantom properties vary with the mode of energy deposition.

Electromagnetic Phantoms

The electric field pattern of an electromagnetic radiator is dependent on the electrical properties of the medium. Thus, an ideal EM hyperthermia phantom will reproduce both the electrical and thermal properties\(^1\) of tissue. Since the electrical properties of tissues vary, it is necessary to account for tissue inhomogeneity as well. An external microwave hyperthermia applicator, for example, must deposit energy through layers of skin, fat, and muscle tissue, so it is important to be able to simulate all of these types of tissue.

It is possible to use \textit{ex vivo} tissue as a phantom material for the qualitative assessment of microwave hyperthermia applicators. However, they are not necessarily representative of living tissues because the permittivity and electrical conductivity of excised tissue and \textit{in vivo} tissue are not necessarily the same [17]. It is, therefore, preferable to use a tissue-equivalent material which has the equivalent electrical properties of living tissue. Early radiofrequency tissue-equivalent phantom materials used to simulate high water-content tissues such as muscle or brain were composed of saline, polyethylene, and a commercial gelling agent. Materials composed of polyester resin, acetylene black, and aluminum powder were used to simulate the dielectric properties of low water-content tissues such as fat and bone [25]. By varying the amounts of the constituents, the properties of various types of tissues at different frequencies could be simulated [11].

Polyacrylamide gel (PAG), which was originally developed for DNA electrophoresis, has been shown to be a good muscle-equivalent phantom material. A base solution of liquid-phase polyacrylamide is doped with sodium chloride to obtain the desired electrical conductivity. After the addition of a catalyst, the solution undergoes an exothermic reaction and solidifies, resulting in a pliable, yet mechanically supportive gel. By varying the salt content, PAG can be used to simulate the dielectric properties of high water content.

\(^1\)In an ideal infrared-thermographic SAR measurement the only relevant phantom thermal property is its heat capacity. In a non-ideal system, however, the high thermal conductivity of the phantom may result in the formation of thermal conduction artifacts.
tissues at low radiofrequencies (14–40 MHz) [5] and at microwave frequencies (750 MHz–5.5 GHz) [1]. A single PAG phantom covering the range of the three international ISM bands (433 MHz, 915 MHz, 2450 MHz) has also been proposed [47].

Ultrasound Phantoms

As a coherent ultrasonic wave propagates through biological tissue it is attenuated due to absorption and scattering. Absorption results from the irreversible conversion of acoustic energy to local heat and is the primary mode of attenuation in tissue. An ideal ultrasonic phantom material would have the same speed of sound, absorption and scattering properties as tissue, in addition to the same thermal properties. To date, most ultrasound phantoms have been designed for ultrasonic imaging applications.

Early phantom materials were composed of a water-based gel derived from animal hides and containing concentrations of n-propanol and powdered graphite [33]. Unfortunately, the melting point of the gel is quite low (32.5°C), making it a poor choice for thermal dosimetry phantom studies. A polysaccharide gel (agar) has recently been suggested as an alternative to the animal-derived gel [6]. Agar has a much higher melting point (78°C) and is, therefore, unlikely to be damaged during heating. Several variations on the agar phantom have been produced, one of which involves the use of closely packed agar spheres of varying diameter with n-propanol filling the interstitial spaces [10].

SAR patterns are usually determined in a phantom by measuring the rate of temperature rise immediately after an applicator is switched on (see section 1.7). Characterisation of an ultrasonic applicator in this fashion is not practical because most ultrasound phantoms, including those described above, are either liquid or semi-liquid. Thermographic SAR measurement techniques typically require a solid state phantom. The development of a standardised solid state ultrasound phantom would allow for a more direct comparison of the SAR patterns of ultrasound and microwave applicators.

1.6 Direct Measurement of SAR

SAR patterns can be determined either directly, by measuring the energy deposition in a medium, or indirectly, by measuring the temperature rise that results from energy deposition into the medium. Direct measurement of SAR can be either empirical or theoretical. Theoretical analysis of electromagnetic fields in biological tissues is complicated because of the complex geometries involved. There is also no guarantee that the applicator SAR pattern will match the one theoretically calculated, because of non-ideal applicator con-
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struction and performance. Although theoretical analysis is a useful tool in applicator
design, measurement of SAR by empirical methods remains essential to quality assurance.

Measurement of electromagnetic and acoustic waves requires different equipment,
but direct SAR measurement techniques are modality specific. Consequently, it is diffi-
cult to compare applicators of different modalities, because experimental conditions may
vary considerably.

1.6.1 Electromagnetic Field Strength

Microwave and RF SAR patterns can be determined directly by measuring the field
strength of the applicator in an electrically conductive medium. Since the loading of
the applicator will affect its power deposition pattern, it is desirable to use a medium
that duplicates the electrical properties of tissue as closely as possible. Standard E-field
probes are composed of three orthogonal dipoles, which are used to measure the z, y, and
z components of the electric field strength. The energy absorption rate associated with an
electric field is

\[ W(x, y, z) = \frac{\sigma}{\rho c} |\vec{E}|^2 \] (1.5)

where \( \vec{E} = E(x, y, z) \) is the electric field vector (V/m), and \( \sigma \) is the electrical conductivity
of the medium (S/m).

Electric field measurements of small antennas in an attenuating medium such as
tissue, are complicated by the lack of small commercial, non-perturbing, implantable E-
field probes. Electromagnetic interference between the probe and the applicator can result
in distortion of the applicator field pattern [29]. High resistance leads are used to minimise
this interference.

1.6.2 Acoustic Pressure

The energy absorption rate associated with an acoustic (ultrasonic) wave is proportional to
the acoustic intensity of the wave, which is proportional to the square of the acoustic pres-
sure amplitude [36]. Hence, a determination of SAR can be made by direct measurement
of acoustic pressure using a hydrophone [4], or an elastic sphere radiometer [18]. These
measurements are typically made in water, and the SAR pattern in tissue must then be
calculated theoretically.
CHAPTER 1. BACKGROUND

1.7 Indirect Measurement of SAR

The deposition of energy in tissue results in heat generation as described in section 1.3. Over short heating periods, when thermal conduction effects are minimal, it is possible to determine specific absorption rate by measuring the temperature rise from equilibrium in the tissue following a step input of power [39]. For tissue at thermal equilibrium \((\frac{dT}{dt} = 0)\) with no applied power, no heat flow occurs, so that \(\nabla^2 T = 0\). If a step input of power is applied at time \(t = 0\), the rate of temperature increase at time \(t = 0^+\) becomes

\[
\rho c \left. \frac{\partial T}{\partial t} \right|_{t=0^+} = W - k \nabla^2 T
\]

If it is assumed that thermal equilibrium continues to hold at \(t = 0^+\), the specific absorption rate \(W(z, y, z)\) can be written as

\[
W = \rho c \left. \frac{\partial T}{\partial t} \right|_{t=0^+}
\]

Hence, the specific absorption rate \(W\) is linearly related to the initial temperature rise in the medium that results from a step input of power. Therefore, the SAR pattern of an applicator can be determined by measuring the temperature rise during a brief power pulse at each point in the medium and calculating the initial slope \(\left. \frac{\partial T}{\partial t} \right|_{t=0^+}\) at that point. It is usually convenient to measure SAR in a tissue-equivalent phantom material.

There are many different commercially available thermometry devices. In principle, specific absorption rate is independent of the method of thermometry in that the measured SAR pattern should be the same regardless of the thermometry device. In practice, however, there are incompatibilities between certain applicators and thermometry devices and limitations on the spatial and temporal resolution of thermometry systems that may result in significant measurement error. Consequently, not all thermometry systems are useful for SAR measurement, due to limitations on resolution, accuracy, and cost.

Different thermometry techniques place different limitations on measuring the three-dimensional temperature distribution within a medium. The three most common thermometry systems used in indirect SAR measurement are discrete temperature probes, liquid crystal films, and infrared imaging radiometers.

1.7.1 Discrete Temperature Probe Thermometry

The temperature distribution within a solid block of phantom material can be measured using discrete temperature probes, ie. probes that measure the temperature of a single
point in space. An applicator catheter is embedded in a box containing a tissue-equivalent phantom material. Thermometry catheters are embedded in the phantom, along tracks either parallel or perpendicular to the applicator axis. Thermometry probes are inserted into the catheters, and a brief pulse of power, typically less than 60 seconds in duration, is applied to the applicator. Temperature measurements are recorded by the probes at discrete time intervals during the heating process. The SAR at the probe location can be calculated by fitting the temperature-time curve recorded at the probe in order to determine the instantaneous slope at time $t = 0$.

The choice of thermometry probe is limited, due to the perturbing effect of metallic probes on electromagnetic transmission and their high rate of heat conduction [8]. Fibre-optic thermometry systems have been shown to be non-perturbing and are used almost exclusively. Recent studies have suggested that thermistors may be suitable for RF and microwave SAR studies, provided that the thermal artifact induced on the thermistor is accounted for [34]. The Dartmouth automated SAR measurement system uses fibre-optic probes advanced through a planar array of small Teflon catheters, resulting in a two-dimensional grid of temperature data [54]. A quadratic least-squares fit of the form $T(t) = a_2 t^2 + a_1 t + a_0$ is used to calculate specific absorption rate.

Aside from the expense of fibreoptic thermometry systems, the major drawback of discrete temperature probes is the limitation on spatial resolution. Probe catheters cannot be placed less than several millimetres apart, resulting in sparse data collection. While resolution along the axis of the catheters can be more precisely controlled since the probe's position can be shifted in the catheter, the measurements must be made in separate experiments due to the acquisition time of the probes. Hence, only a sparse grid of temperature data for a single plane of the phantom can be acquired in a single experiment. While the temperature perturbing effects of the fibreoptic probes may be reasonably ignored, the question remains as to the perturbation induced by the probe catheters.

1.7.2 Liquid Crystal Thermography

Liquid crystal films (LCF) are thermochromatic, i.e. they reflect visible light at different wavelengths depending on their temperature. Consequently, they can be used to measure the temperature distribution of a surface by placing the LCF in contact with the surface. LCFs have only recently come into use in quantitative thermographic SAR studies [2][13].

In a typical experiment, a LCF is embedded in a liquid block of phantom material
prior to polymerisation. The applicator is inserted into the phantom and a step input of power is applied. As the surface of the LCF is heated, it changes colour due to local variation in temperature, resulting in a visible heating pattern. This pattern can be recorded using a commercial video camera and captured to a computer for later analysis. By acquiring a series of images during the heating process, a time evolution of the temperature distribution of a single plane of the phantom can be recorded. The SAR value at a point $P(x, y)$ on the plane can then be determined by approximating the initial slope of the temperature-time curve at the point using an appropriate fitting function, such as the quadratic fit described in section 1.7.2.

Liquid crystal thermography (LCT) provides significantly greater resolution than that afforded by discrete thermometry probes, the maximum resolution being determined by the resolution of the video camera and digitiser. An important advantage of LCT is that temperature values for an entire plane can be acquired in a single experiment, without the acquisition delay between successive experiments associated with discrete thermometry systems. In addition, LCFs are quite inexpensive when compared to other thermometry systems.

There are several drawbacks associated with LCT. LCFs experience very fast deterioration of the colour-temperature relationship when placed in contact with many chemical agents, including the ingredients of many common phantom materials. Hence, a new LCF should be provided for each experiment. Liquid crystal films are sensitive to fairly narrow temperature ranges, typically from 1-5°C, restricting the maximum temperature rise during a LCT experiment. Also, extensive colourimetric analysis must be performed on the LCF in order to calibrate the thermogram [13].

1.7.3 Infrared Thermography

Radiometry

All objects at temperatures above absolute zero continuously emit and absorb radiation over a broad band of wavelengths in the infrared spectrum (0.76-1000 μm). Most materials are opaque to this thermal radiation, so that the contributed portion of radiant power from locations less than a fraction of a millimetre below the material's surface is negligible. Since maximum radiative emission occurs at a wavelength dependent on the surface temperature of the object, an infrared radiation detector can be used to measure the thermal distribution on the surface of an object.

Commercial infrared imaging radiometers typically use photon detectors to mea-
sure incident radiation by either a photo-conductive or photo-voltaic effect. Such imaging radiometers, referred to as thermal cameras, contain a scanning system to enable the detector to scan a wide surface area. Signal output from the detector is used to modulate the intensity of a video signal, resulting in a thermal image which shows relative temperature differences in a range of either grey-scale or false colours. Video images can be displayed to a commercial video monitor, or captured to a computer using a commercial video framegrabber. Digital thermal cameras, which provide high resolution, high quality images, can transfer data directly to a computer or digital tape for storage and analysis.

SAR Measurement

Infrared (IR) thermography has been used in SAR measurement for several decades. Early phantom studies used an IR thermal camera to measure the temperature distribution of tissue-equivalent phantom materials heated by various electromagnetic sources [25] [30]. As with liquid crystal thermography, infrared thermography has many advantages over discrete thermometry probes including significantly greater spatial resolution, and the elimination of effects due to the presence of thermometry catheters.

Due to the opacity of biological and biological-equivalent materials to infrared wavelengths, spatial temperature distributions can only be measured on exposed surfaces. Consequently, determining the spatial temperature distribution within a three-dimensional medium requires phantom materials that are either pre-cut or formed in thin layers in order to allow planar thermal images to be acquired at different depths within the medium. Combining the two-dimensional thermal image of each planar surface results in a three-dimensional temperature distribution.

Since the target imaging surface must be exposed in order for thermal imaging to occur, it is not possible to obtain temporal temperature distributions. Exposing the imaging surface during applicator heating is not possible, because of the resulting distortion to the power deposition pattern. Without time-evolved temperature data, $\frac{\partial T}{\partial t} \bigg|_{t=0^+}$ cannot be determined. Therefore, equation 1.7 cannot be used to calculate the SAR pattern. However, if thermal conduction is assumed to have a negligible effect during a short heating period, then $k \nabla^2 T = 0$ in equation 1.2, reducing it to equation 1.1. Integrating equation 1.1 gives

$$W = \rho c \frac{\Delta T}{\Delta t}$$  \hspace{1cm} (1.8)

where $\Delta T = T_f(x, y, z) - T_i(x, y, z)$ is the temperature rise at a point $(x, y, z)$ in the medium over some heating interval $\Delta t$. Hence, the SAR at a point in the medium is
linearly related to the temperature rise at that point that results from a brief application of power. Therefore, the SAR pattern of an applicator can be calculated by acquiring surface thermal images of planar phantom layers immediately before and after power is applied to the applicator.

The spatial resolution on the imaging plane is limited only by the resolution of the thermal camera and the video acquisition hardware. Given current imaging technology, a resolution of less than $1 \text{mm}^2$ is readily attainable. By comparison, it would require twenty optical probes spaced side by side, and twenty separate experiments with the probe depth incremented by less than $1 \text{mm per experiment}$, to produce the same resolution over a $4 \text{cm}^2$ area. Spatial resolution in the direction perpendicular to the imaging plane is limited by the thickness of the phantom sheets, resulting in a practical maximum resolvability of $1-2 \text{millimetres}$, because it is difficult to construct thinner phantom sheets.

One of the arguments against infrared thermographic SAR measurement has been the high cost of thermal cameras as compared to other thermal measurement systems [45]. In addition, calculating SAR values from differential temperature data requires ignoring thermal conduction effects that may be significant for long heating durations. For this reason, quality assurance guidelines for hyperthermia have required that the heating duration of an infrared thermographic SAR measurement experiment be limited to a 60 second duration [45] [26]. We will demonstrate here that even a 60 second duration results in inaccurate measurement of interstitial hyperthermia applicator SAR patterns.

**Thermal Conduction Artifact**

When a hyperthermia applicator is first turned on, the initial rate of temperature rise in the medium will be proportional to the energy deposition rate (equation 1.1). If the medium was not thermally conductive, the temperature at each point would rise linearly with time, because there would be no heat flow in the medium, so that equation 1.1 would apply at any time $t$. However, biological tissue is thermally conductive, the rate of heat conduction at a point in the medium being proportional to the thermal gradient at that point. As more heat is deposited in the medium, the local thermal gradients become larger, so that the conduction term $k \nabla^2 T$ in equation 1.2 increases. The longer the heating period, the greater the effect that results from this non-linear conduction term. Since the rate of heat conduction will be different at different points in the medium, the shape of the SAR pattern will become distorted over time. Such a distortion is conveniently referred to as a thermal conduction artifact. If the heating time is sufficiently short, the conduction artifact can
be considered negligible and the measured temperature distribution will vary linearly with the SAR distribution according to equation 1.8.

**Surface Cooling Artifact**

Since it is necessary to physically manipulate the phantom in order to expose a heated surface, there is a time delay between the end of the applied power pulse and thermal image acquisition. During this delay, which is typically on the order of 5 seconds or more [44] [47] [9], the phantom material is cooled by both thermal conduction within the medium, and convective heat transfer due to air flow above the phantom surface. Because the rate of surface cooling is proportional to the thermal gradient at each point on the phantom surface, the amount of cooling will vary considerably from point to point. Consequently, in addition to the diffusion artifact, the temperature distribution measured from a thermal image is subject to a *surface cooling artifact*.

Technically, the conduction and surface cooling artifacts both result from cooling of the phantom material. However, it is convenient to distinguish between the artifacts that result during and after heating. Due to the combined effect of cooling artifacts, it has been argued that infrared SAR measurement is only valid for qualitative assessment of SAR [13]

**Applicator Coupling**

Infrared thermographic SAR analysis is convenient for external applicators, which can be externally coupled to the phantom material using a bolus or coupling pad. Interstitial applicators pose a difficulty because they must be coupled to the interior of the medium, typically requiring that they be embedded in the middle of a phantom layer. This is usually accomplished by using a thicker phantom layer in the middle of a multi-layered phantom block. Given the thickness of most hyperthermia catheters (approximately 1 mm), it is not practical to embed a catheter in a phantom sheet of thickness less than 3 mm, and such phantom sheets are quite fragile so that extreme care must be taken in their handling. As a result, it is not possible to obtain an image of the plane of the applicator axis, since the surface of that plane cannot be exposed to the thermal camera for an embedded applicator. It is necessary instead to rely on images acquired either above or below the applicator plane to characterise the device. Since the rate of energy deposition drops rapidly with distance from the antenna, the SAR pattern only a few millimetres away from the applicator plane does not provide an accurate estimation of the power deposited in the medium.
CHAPTER 1. BACKGROUND

Inserting the applicator between two adjacent phantom layers eliminates the problems associated with an embedded applicator. However, the effect that this will have on the power deposition pattern is unclear, because of the potentially poor and inhomogeneous coupling between the applicator catheter and the phantom that may result. Differences in applicator coupling may considerably alter the shape of the applicator’s SAR pattern [53].

Phantom Layer Partitions

Phantom structures used in thermographic studies are, of necessity, constructed to separate along planar partitions. Liquid or low-viscosity phantom materials require thin partitions, typically constructed of thin plastic (mylar) films or silk screens, to maintain the shape of the phantom at planar boundaries. Solid phantom materials do not require plastic films or other insulating materials to maintain their rigidity. However, it is quite common to use such films to facilitate phantom separation during the thermographic process. Although plastic films provide some advantage in phantom manipulation and in limiting surface cooling [30], the film may act as an insulating partition between adjacent phantom layers during heating. Thermographic studies of radiofrequency hyperthermia applicators have indicated that the presence of an electrical discontinuity in the phantom due to phantom layering results in considerable distortion of the applicator heating pattern [32]. At microwave frequencies, theoretical analysis has indicated that insulating partitions do not significantly affect the SAR patterns of external microwave applicators, provided the partitions are sufficiently thin [42]. The effect of insulating partitions on the measured SAR patterns of interstitial microwave applicators remains unclear.

In the absence of an insulating partition, the boundary between adjacent phantom layers remains a potential source for SAR pattern distortion because air can become trapped between layers during assembly of a phantom block. Theoretical analysis of such an effect is complicated because the air is trapped in what are essentially randomly distributed pockets.

1.8 Summary and Objectives

The need for higher spatial resolution in SAR measurement systems for interstitial applicators has led to increasing use of thermographic techniques such as infrared thermography. The validity of specific absorption rate as a criterion for applicator characterisation and treatment planning is based on the assumption that SAR is an accurate representation of energy deposition in tissue prior to thermal conduction and forced convection of heat by
blood flow. The accuracy of SAR patterns is particularly important in the modeling of heat transfer in biological tissue, since errors at the input to the model can be magnified considerably at the output. Although the literature recognises thermal conduction and convection as sources of error in infrared thermographic SAR measurement, such errors are typically assumed to be negligible for short measurement durations of less than 60 seconds.

The intent of this work is to demonstrate that conduction and convection effects are a considerable source of error in infrared thermographic measurement of interstitial hyperthermia applicator SAR patterns even for heating durations of less than 60 seconds. It will be further demonstrated that there are other factors, such as applicator coupling and phantom layer partitioning, that may result in considerable variability in the measured SAR patterns of interstitial microwave applicators.

References


REFERENCES


REFERENCES


REFERENCES


REFERENCES


REFERENCES


Chapter 2

Methodology

2.1 Introduction

Experiments were performed to determine the effect of thermal conduction and convection, applicator coupling, and phantom layering on the SAR patterns of interstitial applicators measured using a multilayer phantom infrared thermographic technique. For comparative purposes, SAR patterns were also measured using thermocouples embedded in a block of phantom material. Two microwave and one ultrasound applicator were evaluated, in order to demonstrate that the presence of thermal artifacts is independent of the modality of power delivery. The OCI/PMH Thermal Imaging System [4] was used for all infrared thermographic SAR measurements, and was specifically modified for use in the present project. The multisection dipole and ultrasound experiments were performed using an earlier design of the imaging system [4].

2.2 OCI/PMH Thermal Imaging System

The thermal imaging system consisted of an in-house test platform, Inframetrics 522L infrared imaging radiometer\textsuperscript{1}, ComputerEyes/1024 video framegrabber\textsuperscript{2}, PC computer, and a set of Matlab\textsuperscript{3} routines specifically designed for this project. A schematic of the system is shown in figure 2.1. Images were captured to the PC using DEYES image acquisition software provided by the framegrabber manufacturer. A low gain (approx. unity) video amplifier\textsuperscript{4} was used to stabilise the video input presented to the framegrabber.

\begin{footnotesize}
\textsuperscript{1}Inframetrics Corporation, N. Billerica, MA
\textsuperscript{2}Digital Vision Incorporated, Dedham, MA
\textsuperscript{3}The Mathworks Incorporated, Natick, MA
\textsuperscript{4}LM6181, National Semiconductor Corporation, Santa Clara, CA
\end{footnotesize}
CHAPTER 2. METHODOLOGY

In order to ensure accurate, high resolution temperature measurements, the thermal camera required a temperature reference, which was provided by a supply of liquid nitrogen stored in a well fed through the top of the camera. Consequently, it was necessary to keep the camera in an upright position at all times in order to prevent leakage. An infrared reflective mirror was used to bend the path of the infrared radiation by 90°, so that the phantom sheets could be stacked vertically. Horizontal stacking of the sheets would have eliminated the need for a reflecting mirror, but preliminary experiments indicated that the sheets exhibit very poor adhesion when stacked in a horizontal configuration.

Since the heating patterns of interstitial applicators are small, it was essential to be able to accurately identify the location of the applicator within the thermal image. To allow for accurate positioning of the applicator, an 18 cm × 25 cm alignment grid with a 1 mm² resolution was mounted to a plexiglass sheet on the test platform, within the thermal camera target area. The grid was aligned to the imaging plane of the thermal camera by rotating the grid until the thermal image of a heated target positioned on the horizontal origin of the grid was aligned with the thermal image horizontal axis as indicated in figure.
2.2. The target was repositioned on the vertical origin of the grid and the grid was rotated until the target was aligned to the thermal image vertical axis. The target was a 3 cm wide plastic ruler, heated by immersion in water at 100°C.

2.2.1 Calibration Procedure

When a target is placed at a distance \( d \) in front of the thermal camera, the dimensions of the thermal image of the target depend on the field of view (FOV) of the camera, the angle of the reflecting mirror, and the capture resolution of the video framegrabber. In order to correctly calibrate the target \( z \) and \( y \) axes, it was necessary to determine the horizontal and vertical pixel to size ratios of the acquired thermal image. The infrared mirror was maintained at an angle of 45° to both the plane of the test platform and the camera imaging plane in all experiments in order to maximise the camera field of view.

The image axes were calibrated using the alignment grid to correlate a point \( P(x,y) \) on the phantom surface to a point \( Q(h,v) \) in the thermal image, where \( x \) and \( y \) are the distances from the alignment grid origin in millimetres, and \( h \) and \( v \) are the coordinates of a pixel in the thermal image. A near-blackbody target was horizontally aligned to the origin of the alignment grid and a thermal image was acquired. The blackbody target was re-aligned with successive horizontal grid lines at locations of 1–4 cm in 1 cm intervals, and additional thermal images were acquired. The images were ported to the Matlab 4.2 environment using the BMPREAD function and displayed using the Matlab IMAGE function. The alignment edge of the blackbody target in each image was identified visually and its \( y \) coordinate was determined using the GINPUT function. The process was repeated for vertical grid lines from 0–18 cm in 2 cm intervals. Horizontal and vertical calibration constants were determined by a linear least-squares fit of the horizontal and vertical target grid data to the horizontal and vertical image data respectively.

Image axes were re-calibrated after any adjustment of the camera FOV in order to calculate the correct pixel to size ratio. It was not necessary to re-calibrate the axes when the image resolution was changed, since a linear scaling factor could be applied to the calibration constants (e.g. doubling the image resolution halves the size of each pixel).
Figure 2.2: Calibration of the Alignment Grid. (a) Although the target object is aligned to the horizontal zero of the alignment grid, the thermal camera image indicates a mis-alignment with the horizontal image axis. (b) Rotation of the alignment grid corrects the thermal image alignment. The camera target area is indicated by dashed lines.
2.2.2 Imaging Procedure

A phantom block was assembled from a set of thin phantom sheets. The edge of the phantom block was aligned to the $z$-axis of the alignment grid, with the catheter well aligned to the $y$-axis as indicated in figure 2.3(a), to enable the location of the applicator to be accurately determined\(^1\). Prior to catheter and antenna insertion, an initial background thermal image of the imaging plane was acquired\(^2\). The applicator was embedded in the phantom well and power was applied for a heating time $t_h$. After heating, all phantom layers above the desired imaging plane were manually removed. The interstitial applicator and catheter (if applicable) were also removed to prevent the applicator from acting as a source of heat after power was turned off. A thermal image of the surface of the phantom sheet was captured using commercial image acquisition software\(^3\). All images were captured at a resolution of $640 \times 480$ pixels unless otherwise noted. Images were captured in 8-bit grey-scale with brightness and contrast levels of 16% and 8% respectively\(^4\). The images were stored on a PC for later analysis.

2.2.3 Thermal Image Processing

Stored thermal images were ported to the Matlab 4.2 environment for temperature conversion and processing using a set of Matlab routines that were designed for the project. The thermal image axes were converted from pixels to millimetres using the calibration factors measured during the calibration procedure (section 2.2.1). Images were then down-sampled to a $1 \text{ mm} \times 1 \text{ mm}$ spatial resolution, because the resolution of the video framegrabber was higher than the resolution of the thermal camera. Grey-scale values were converted to temperature values using data provided by the thermal camera manufacturer \(^3\). SAR values at each point in the image were calculated using equation 1.8.

Images acquired in earlier experiments required additional processing because a background thermal image was not acquired. These images were first filtered to remove noise contributed by the video framegrabber. Noise pixels were identified by applying a first order difference function to the image pattern, using the Matlab DIFF function. Differences greater than 20 grey-scale levels were considered 'noisy'. Noise pixels were removed by linear interpolation of adjacent 'noiseless' pixels in both vertical directions, or

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\(^1\) Multisection dipole and ultrasound applicators were aligned as indicated in figure 2.3(b).

\(^2\) Background thermal images were not acquired in the multisection dipole and ultrasound experiments.

\(^3\) Digital Vision Incorporated, Dedham, MA

\(^4\) The brightness and contrast levels control the offset and range of the grey-scale respectively. Percentage values are relative to the maximum number of grey-scale levels (256 levels for 8-bit acquisition).
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Figure 2.3: Applicator and Phantom Sheet Orientation. Orientation (top view) of the antenna and phantom sheets during the infrared imaging experiments for (a) the helical coil antenna, and (b) the multisection dipole and ultrasound applicator. No alignment grid was used in the dipole and ultrasound experiments. Image axes orientation is indicated in the images to the right, with the applicator diagram superimposed to show the antenna position. Note that although the thermal images are vertically inverted by the infrared mirror, the orientation of the grey-scale indicator on the bottom of the image remains the same.
in a single direction when noise appeared at the upper or lower edge of the image. Bilinear interpolation was not used because noise occurred preferentially in horizontal bands. Average initial temperature was determined by construction of a histogram of the background region of the image using the Matlab IMHIST function. The average background grey-scale level was then set to the grey-scale level corresponding to the maximum histogram count, which corresponds to the mode of the background distribution. This value was subtracted from the filtered image to remove the background temperature offset.

2.3 Hyperthermia Equipment

2.3.1 Microwave Power Delivery

A commercial BSD-500 Hyperthermia System\(^1\) was used for microwave power delivery in the thermocouple experiments. Although the duration of power delivery was automatically controlled and user adjustable, the system had a minimum treatment duration setting of 60 seconds. Therefore, heating intervals of less than 1 minute could be achieved only by manual switching. Manual switching, however, resulted in an increased manipulation delay because the power had to be manually shut off (due to safety considerations) before the phantom layers could be removed and imaging could occur.

To minimise this delay, an electronic cutoff circuit was designed to automatically limit the duration of power application to a desired time interval during IR imaging experiments. The cutoff switch could not be incorporated into the BSD-500 system, so it was necessary to use an in-house microwave delivery system for the infrared imaging experiments. The in-house system consisted of an HP8656A signal generator\(^2\) coupled to an in-house microwave amplifier with built-in pre-amplifier. An impedance mismatch between the pre-amplifier and the signal generator resulted in a minimum applied power level of approximately 5 W, with a minimum increment of approximately 1 W. The cutoff circuit consisted of an in-house IC count-down timer, adjustable for count-down times of up to 600 seconds, coupled to an SPDT electronic switch\(^3\). The circuit was inserted between the signal generator and the pre-amplifier. The in-house microwave delivery system was used for all infrared imaging experiments unless otherwise noted.

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1. BSD Medical Corporation, Salt Lake City, UT
2. Hewlett Packard Canada Ltd., Mississauga, ON
3. ZFSWA-2-46, Mini-Circuits, Brooklyn, NY
CHAPTER 2. METHODOLOGY

2.3.2 Microwave Applicators

Helical Coil Antenna

The helical coil antenna\(^1\) consisted of a rigid 45 cm long (0.9 mm diameter) coaxial feedline with a 26 mm extension of inner conductor. A 26 mm long helical tin coil emitter (1.05 mm diameter) with a coil turn density of 30 turns/cm was soldered to the tip of the inner conductor. A diagram of the antenna is shown in figure 2.4(a). The antenna was designed to be driven at a frequency of 915 MHz and had a maximum input power level of 10 W.

![Diagram of a helical coil antenna](image)

Figure 2.4: Interstitial Hyperthermia Applicators. Interstitial applicators used in the experiments. (a) Dornier helical coil antenna. (b) BSD multisection dipole antenna. (c) UCSF interstitial ultrasound applicator. All units are in millimetres.

Multisection Dipole Antenna

The multisection dipole antenna was a BSD MA-250 interstitial applicator\(^2\). The antenna consisted of a rigid 32 cm coaxial feedline with a 45 mm extension of inner conductor. A series of enlarged collars were asymmetrically spaced about the junction. The exposed dielectric at the gap was 10 mm long. A diagram of the antenna is shown in figure 2.4(b).

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\(^1\) courtesy of Dornier Medical Systems Incorporated, Kennesaw, GA

\(^2\) BSD Medical Corporation
The applicator was designed to be driven at a frequency of 915 MHz and had a maximum input power level of 15 W.

2.3.3 Interstitial Ultrasound Applicator

The interstitial ultrasound applicator\(^1\) consisted of two 1.0 mm diameter PZT tubular radiators of 1 cm length mounted on a flexible segmented array, shown in figure 2.4(c). The radiators had an operating frequency range of 5–7 MHz, with a maximum input power level of approximately 10 W. The applicator was designed for direct implantation, ie. the transducers are designed to couple directly to a medium, without the need for an insulating catheter.

2.4 Phantom Materials

2.4.1 Phantom Solution

Polyacrylamide gel (PAG) was used to simulate the thermal and dielectric properties of muscle tissue at 915 MHz. A polyacrylamide stock solution was prepared according to the method described by Surowiec et al. [5]. The composition of the stock is shown in table 2.1. Polyacrylamide gel was formed by reacting the stock solution with the catalysts (ammonium persulphate and TEMED). The mixture was poured into a mould, where gel polymerisation occurred following an exothermic reaction.

<table>
<thead>
<tr>
<th>Component</th>
<th>Gel (% by weight)</th>
<th>Stock solution (% by weight)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Acrylamide</td>
<td>26.0</td>
<td>26.0</td>
</tr>
<tr>
<td>N,N'-methylenebisacrylamide</td>
<td>0.196</td>
<td>0.196</td>
</tr>
<tr>
<td>Ammonium persulphate (10%)</td>
<td>1.0</td>
<td>—</td>
</tr>
<tr>
<td>Sodium chloride</td>
<td>1.05</td>
<td>1.05</td>
</tr>
<tr>
<td>De-ionised water</td>
<td>71.7</td>
<td>71.7</td>
</tr>
<tr>
<td>TEMED(^a)</td>
<td>0.05</td>
<td>—</td>
</tr>
</tbody>
</table>

\(^a\)N,N,N',N-tetramethylethylenediamine

A summary of the thermal and dielectric properties of the polyacrylamide gel is provided in table 2.2. Values were taken from the literature [5] [1] [2].

\(^1\)courtesy of C.J. Diederich, University of California, San Francisco, CA
CHAPTER 2. METHODOLOGY

2.4.2 Phantom Block for Thermocouple Experiments

A 22 cm length of 1.19 x 1.70 mm diameter teflon tubing was suspended in a 1 L beaker. Three type K thermocouples (junction ~ 0.1–0.2 mm) were attached with thread to the exterior of the tubing at depths of 7.8 cm, 8.05 cm, and 8.3 cm relative to the top of the beaker. Four additional thermocouples were attached to a small wooden dowel at 0.25 cm intervals. The dowel was bonded to the teflon tubing with a commercial epoxy\(^1\) in a position perpendicular to the axis of the tubing. A schematic of the thermocouple positioning is shown in figure 2.5. The beaker was filled with PAG solution and allowed to set.

\[
\begin{array}{|c|c|c|}
\hline
\text{Property} & \text{Muscle} & \text{PAG} \\
\hline
\text{Electrical conductivity } \sigma \ (\text{S/m}) & 1.37 & 1.39 \\
\text{Rel. Permittivity } \varepsilon & 55 & 55.5 \\
\text{Wavelength } \lambda \ (\text{cm}) & 4.3 & 4.3 \\
\text{Density } \rho \ (\text{kg/m}^3) & 1070 & 1070 \\
\text{Specific heat } c \ (\text{J/kg°C}) & 3470 & 3810 \\
\text{Thermal conductivity } k \ (\text{W/m°C}) & 0.535 & 0.395 \\
\hline
\end{array}
\]

Figure 2.5: Thermocouple Phantom Block. Phantom block configuration used in the thermocouple experiments. Seven type K thermocouples were placed in 0.25 cm intervals in a plane passing through the catheter axis. Numbered circles indicate thermocouple positions.

\(^1\)Lepage's Ltd., Brampton, ON
2.4.3 Phantom Sheets for IR Imaging Experiments

**Standard Sheets**

A 0.25 cm thick plexiglass spacer was inserted between two 22 cm x 24 cm x 0.16 cm sheets of glass. The edges of the spacer and glass were sealed with vacuum grease\(^1\) and clamped to prevent leakage. 100 mL of polyacrylamide solution were poured into the mould and allowed to set for twenty minutes. The resultant 20 cm x 20 cm x 0.25 cm phantom sheets were stored at 4°C until needed. Additional spacers were used to create phantom sheet thicknesses of 0.5 and 1.0 cm. A diagram of the mould is shown in figure 2.6.

![Diagram of the mould](image)

**Figure 2.6: Standard Phantom Sheet Mould.** Basic mould used in the formation of phantom sheets. Polyacrylamide solution was poured into the mould through an opening in the top. Gelling occurred within a ten to twenty minute period, following an exothermic reaction.

**Welled Sheets**

Welled phantom sheets were constructed using the procedure described in section 2.4.3 with the following modifications. A hemicylindrical 1.1 mm diameter steel rod was bonded to one sheet of glass using vacuum grease, as shown in figure 2.7. After the acrylamide

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\(^1\)Dow Corning Corp., Midland, MI
solution had reacted, the steel rod was removed, leaving a hemicylindrical well along one
side of the phantom sheet. Phantom sheets were stored at 4°C.

![Figure 2.7: Welled Phantom Sheet Mould. A helicylindrical steel rod was used to form a well on
the surface of the phantom sheet to fit the applicator catheter.](image)

**Embedded Sheets**

A 21 cm length of \(1.57 \times 2.08\) mm diameter polyethylene tubing was inserted through
two holes drilled in a 0.5 cm thick plexiglass spacer, as indicated in figure 2.8. A 1.1 mm
diameter steel rod was inserted into the polyethylene tubing to maintain the rigidity of the
tubing prior to phantom polymerisation. The spacer was inserted between two 22 cm \(\times\)
23 cm \(\times\) 0.16 cm sheets of glass and sealed as in section 2.4.3. 200 mL of polyacrylamide
solution was poured into the mould and allowed to react for twenty minutes. The gelled
phantom sheet containing the embedded tubing was removed from the mould and stored
at 4°C.

**2.5 Thermocouple Experiments**

**2.5.1 Helical Coil Antenna**

The helical coil antenna was inserted into the teflon tubing (figure 2.5) to a depth of 7.8 cm
and high microwave power (10 W) at 915 MHz was applied for 20 seconds using the BSD-
500 Hyperthermia System. A multi-function 16-channel data acquisition system\(^1\) recorded
and converted thermocouple measurements. Temperature measurements were stored on a

\(^1\)Labmate Series 7000, Labmate Sciometrics, Manotick, ON
Figure 2.8: Embedded Catheter Sheet Mould. Frame used in the formation of an embedded phantom sheet. The catheter was embedded in the middle of the 0.5 cm mould prior to PAG polymerisation.

PC for later analysis. Temperatures at thermocouples 1–7 were recorded at approximately 1.5 second intervals during heating and for an additional 30 seconds following heating. After heating, the phantom block was allowed to equilibrate to room temperature (21°C). The experiment was repeated for antenna insertion depths from 8.05 to 10.8 cm in 0.25 cm intervals. The experiment was also repeated for applied power levels of 5, 7, and 8 W, at a single insertion depth of 9.3 cm.

2.5.2 Multisection Dipole Antenna

The multisection dipole antenna was inserted into the teflon tubing to a depth of 7.8 cm and driven with 15 W at 915 MHz for 30 seconds using the BSD-500 Hyperthermia System. Temperatures at thermocouples 1–7 were recorded at approximately 1.5 second intervals during heating and for an additional 30 seconds following heating. After heating, the phantom materials were allowed to equilibrate to room temperature. The experiment was repeated for insertion depths from 8.05 to 10.8 cm in 0.25 cm intervals.
2.6 Infrared Imaging Experiments

2.6.1 Thermal Diffusion Experiments

Helical Coil Antenna

Four 1 cm and two well 1 cm phantom sheets were equilibrated to room temperature (21–23°C) after removal from storage at 4°C. The phantom materials were stacked into a layered block in the configuration shown in figure 2.9(a) and oriented on the alignment grid, with the phantom well parallel to the horizontal image axis, as indicated in figure 2.3(a). The phantom block was manually compressed to remove air trapped between individual layers. A surface thermal image of the applicator plane was acquired prior to heating, using the modified OCI/PMH Thermal Imaging System. A 20 cm length of 1.57 × 2.08 mm diameter polyethylene tubing was placed in the well formed between the two middle layers of the phantom block. The helical coil antenna was inserted into the tubing to an insertion depth of 12 cm. The antenna was driven with 9.2 W for 10 seconds using the in-house 915 MHz microwave power system. Immediately after heating, a thermographic image of the surface of the applicator plane was obtained. The maximum delay from the end of heating to acquisition of the thermal image was 5 seconds. The phantom materials were then allowed to equilibrate to room temperature for 40 minutes. The procedure was repeated for heating durations of 20, 30, 40, 60, and 90 seconds. Five repeated experiments were performed for each heating duration. The procedure was then repeated for applied power levels of 7.6 W and 10.4 W for heating intervals of 10, 20, 30, 40, and 60 seconds, with three repeated experiments per heating time.

Multisection Dipole Antenna

Two 1 cm, six 0.25 cm, and two well 0.25 cm phantom sheets were removed from storage and equilibrated to room temperature. The phantom materials were stacked in a layered block in the configuration shown in figure 2.9(b), and oriented in the camera target area with the phantom well parallel to the vertical image axis, as indicated in figure 2.3(b). A 16 Ga × 9 in. hyperthermia sensor catheter\(^1\) was placed in the phantom well and coupled to the phantom material using de-ionised water. The segmented dipole antenna was inserted into the catheter to a depth of 9 cm. Low microwave power (5 W) at 915 MHz was applied

\(^1\) Catalog #1609, Valley Medical Research Incorporated, Fruit Heights, UT
CHAPTER 2. METHODOLOGY

Figure 2.9: Phantom Block Configurations. Configuration of phantom sheets in the imaging experiments using (a) 1 cm sheets, (b) 0.25 cm sheets, (c) 0.5 cm sheets, and (d) embedded applicator.
for 30 seconds using the BSD-500 Hyperthermia System. The upper phantom layers, applicator, and catheter were removed immediately after heating. A thermographic image of the surface heating pattern of the applicator plane was obtained with the unmodified OCI/PMH Thermal Imaging System. The maximum delay from the end of heating to acquisition of the thermal image was 10 seconds. The phantom materials were allowed to equilibrate to room temperature for 20 minutes. The procedure was repeated for applied power levels of 5, 8, and 10 W, and for heating times of 30, 45, and 60 seconds.

**Ultrasound Applicator**

Two 1 cm, six 0.25 cm, and two welled 0.25 cm polyacrylamide phantom sheets were removed from storage and equilibrated to room temperature. The phantom materials were stacked in a layered block in the configuration shown in figure 2.9(a). The ultrasound applicator was placed in the phantom well at an insertion depth of 9 cm and coupled to the phantom sheets with de-ionised water. A single element of the applicator was driven with 6.5 W at a frequency of 6.8 MHz for 45 seconds using a BSD-250 Ultrasound Hyperthermia System. The upper phantom layers and applicator were removed within 10 seconds of the end of heating and a thermographic image of the surface heating pattern of the applicator was obtained with the unmodified OCI/PMH Thermal Imaging System. Phantom materials were allowed to equilibrate to room temperature for forty minutes. The procedure was repeated for a heating times of 10 and 20 seconds.

**2.6.2 Surface Cooling Experiments**

Four 1 cm, six 0.25 cm, and two welled 0.25 cm thick phantom sheets were assembled in the configuration of figure 2.9(a). The procedure of section 2.6.1 was used, with the following modifications. The antenna was driven with 10.4 W for 10 seconds using the in-house microwave delivery system. Thermographic images of the applicator plane were acquired at 2.5 second intervals over a 60 second period following the end of heating, the first image being acquired 5 seconds after heating. The experiment was repeated for heating times of 20, 30, 40, 50, and 60 seconds, and for imaging on the +0.25 cm plane, and +0.5 cm plane (for a 30 second heating time).

\(^1\text{BSD Medical Corporation}\)
2.6.3 Phantom Layering Experiments

0.25 cm Phantom Layers

Four 1 cm and eight 0.25 cm phantom sheets were removed from storage and equilibrated to room temperature. A phantom block was assembled using the configuration of figure 2.9(b). The procedure of section 2.6.1 was used, with the following modifications. The antenna was driven with 10.4 W for 10 seconds using the in-house microwave delivery system. After the phantom block equilibrated to room temperature, the experiment was repeated for heating times of 20 and 40 seconds, with imaging occurring on both the applicator and +0.5 cm planes for the 40 second heating duration.

0.5 cm Phantom Layers

The procedure of section 2.6.3 was used with the following modifications. The phantom block was reassembled to the configuration of figure 2.9(c) and equilibrated to room temperature. Thermographic images were acquired for heating times of 10, 20, and 40 seconds on both the applicator and +0.5 cm planes, with an applied power level of 10.4 W.

1.0 cm Phantom Layers

The phantom block was reassembled to the configuration of figure 2.9(a). Thermographic images were acquired for heating times of 10, 20, and 40 seconds on the applicator plane, using the procedure of section 2.6.3.

2.6.4 Applicator Coupling Experiments

Embedded Applicator

The procedure of section 2.6.3 was used with the following modifications. A phantom block was assembled using the configuration shown in figure 2.9(d) and allowed to equilibrate to room temperature. After heating, the applicator and upper phantom sheets were removed. The tubing remained embedded within the 0.5 cm phantom sheet. Thermal images were acquired on the +0.25 cm plane for heating times of 10, 20, 30, 40, 50 and 60 seconds with three repeated experiments for each heating time.

Welled Applicator

The procedure of section 2.6.3 was used with the following modifications. A phantom block was assembled using the configuration shown in figure 2.9(b) and allowed to equilibrate
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2.7 Data Analysis

2.7.1 Calculation of SAR

Specific absorption rate for the thermocouple experiments was calculated using equation 1.7 and temperature data supplied by the discrete thermocouples. $\frac{\partial T}{\partial t} \bigg|_{t=0^+}$ was calculated by applying a linear fit to the first 3 seconds of temperature data (which is only two data points). Such a 3 second fit is the closest approximation to $\frac{\partial T}{\partial t}$ that can be made given the sampling speed of the thermometry system. For comparison, a 5 second fit was calculated using a linear least-squares fit to the first 5 seconds of temperature of the form $T(t) = a_1 t + a_0$, so that $\frac{\partial T}{\partial t}$ at $t = 0$ is just $a_1$.

SAR values for all other heating durations were calculated using equation 1.8, where $\Delta T = T_{final} - T_{initial}$ is the temperature rise at a point $P(x, y)$ in the medium for a heating duration $\Delta t$. Since it is possible to continue to heat during temperature measurement, SAR values were calculated for heating times less than the total heating duration $t_h$ by writing $\Delta T = T_t - T_i$ where $T_t$ is the temperature at a point $P(x, y)$ after a heating time $t$ and $T_i$ is the temperature at the point prior to heating. In such a fashion, it is possible to estimate the contribution of thermal diffusion during heating without including convection effects, as would be necessary in an IR imaging experiment. SAR values calculated using equation 1.8 are referred to as $t$ second fitted values, where $t$ is the heating duration.

SAR values in the infrared imaging experiments were calculated as described in section 1.7.3 using equation 1.8.

2.7.2 IsoSAR Contours

IsoSAR contour plots were calculated using the Matlab CONTOUR function, which uses bilinear interpolation to create lines of constant SAR from a two-dimensional SAR data set. Since individual thermocouple experiments yielded SAR values along a single plane only (perpendicular to the applicator axis), it was necessary to combine the results of successive thermocouple experiments to form a single two-dimensional SAR pattern. No correction was made for insertion depth effects due to the potential for small changes in applicator position in successive multisection dipole antenna experiments. Interaction between the
thermocouple probe and the microwave applicators was also not accounted for. Contour plots were normalised to the maximum SAR value in the plane.

2.7.3 Characterisation of Applicators

Hyperthermia applicators are typically characterised by isoSAR contour measurements. The applicator hot spot was defined as the location of the maximum SAR value relative to the tip of the antenna. The hot spot location is an important descriptor in hyperthermia because it will usually, but not always, identify the location of maximum heating in tissue (the hot spot location may shift due to forced convection of heat by blood flow). The 50% Heating Length was defined as the length of the SAR pattern with SAR greater than or equal to 50% of the maximum value at the hot spot. Similarly, the 50% Heating Width was defined as the radial width of a plane perpendicular to the antenna axis and passing through the applicator hot spot with SAR greater than or equal to 50% of maximum. Heating lengths and widths were determined from the interpolated isoSAR contour plots. The Effective Field Area was defined as the area of the 50% isoSAR contour, i.e. the total area of the heating pattern that has SAR greater than or equal to 50% of the maximum value at the hot spot, and will be denoted EFA$_{50}$. It was convenient to define effective field areas for isoSAR contour levels other than 50%, which were denoted EFA$_{z}$, where $z$ is the contour level. Effective field areas were calculated using the Matlab TRAPZ function.

2.7.4 Surface Cooling

The effect of diffusive cooling on the thermocouple measurements was measured by calculating SAR values using equation 1.8, where $\Delta T = T_f - T_i$, with $T_f$ equal to the temperature recorded at a thermocouple 5 seconds after the end of the heating duration (i.e. after 5 seconds of diffusive cooling), for a $\Delta t$ equal to the heating duration. SAR values calculated in this fashion were referred to as a 'cooled' fit. Since the cooling effect on the thermal images cannot be directly measured, it was necessary to estimate the rate of cooling using temperature data gathered after the cooling period. A linear least-squares fit was applied to the three temperature values corresponding to times of 5, 7.5, and 10 seconds after the end of the heating duration. The fitted line was projected back to the end of heating to estimate the temperature prior to the cooling period (figure 2.10). By applying a separate fit at each point on the plane, a spatial temperature distribution of the imaging plane was calculated which is a first order approximation to the temperature distribution prior to diffusive and convective cooling.
Figure 2.10: Estimation of Temperature Prior to Surface Convection. The temperature at a point in the phantom prior to convective surface cooling was estimated by a first order fit (solid black line) to the three temperature values recorded at times of 5, 7.5, and 10 seconds (black circles) after removal of the phantom layer. The corrected ('uncooled') temperature value is indicated by the intersection of the solid and dashed lines. The theoretical temperature distribution is shown in grey.

References


Chapter 3

Results

3.1 Thermal Diffusion

The effects of thermal conduction and surface convection artifacts on the SAR pattern of a helical coil and multisection dipole microwave hyperthermia applicator were measured by comparing the SAR patterns for different heating durations using both an infrared thermographic and a thermocouple thermometry system. Infrared thermographic experiments were also performed on an interstitial ultrasound applicator. The results of both the thermographic and thermocouple experiments indicated considerable smearing of the SAR pattern during the first 30 seconds of heating, suggesting that a short heating duration is required for accurate infrared thermographic SAR measurement of interstitial applicators.

3.1.1 Thermocouple Experiments

Helical Coil Antenna

Axial and radial SAR profiles along the default profile planes\(^1\) are shown in figure 3.1 for a heating duration of 20 seconds calculated using a 3 second fit, a 20 second fit, and a ‘cooled’ 20 second fit. In all three cases, the antenna hot spot, i.e. the point of maximum SAR, was located on the antenna axis at a distance of 5 mm from the tip of the antenna. The magnitude of the calculated SAR profiles was found to decrease with increasing heating duration. The SAR magnitude at the hot spot was 73% lower than the 3 second fit value after 10 seconds of heating, and 85% lower after 20 seconds of heating. After an additional 5 second cooling period, the value was further reduced so that the maximum SAR was

\(^1\)Axial plane along the antenna axis; radial plane perpendicular to the antenna axis and passing through the applicator hot spot.
95% lower than that calculated by the 3 second fit.

The percentage error in SAR magnitude relative to the 3 second fit value was found to decrease with increasing radial distance from the antenna hot spot, as shown in figure 3.2. Note that at radial distances greater than 5 mm, the initial temperature rise approaches the minimum accuracy of the thermometry system (0.04°C), so that the 3 second fit is extremely poor because of quantisation error and noise. A similar reduction with increasing distance from the hot spot was seen in the axial direction. Hence, the error in estimation of SAR relative to the 3 second fit was greatest at the antenna hot spot. The SAR amplitude error varied by only a few percent over an applied power range from 5 to 10 W (figure 3.3).

![Figure 3.1: Dornier Thermocouple SAR Profiles. Radial and axial SAR profiles of the helical coil antenna for a 10 W/20 s power pulse calculated using a 3 s fit (black), a 20 s fit (dark grey), and a 'cooled' 20 s fit (light grey). Thermocouple measurement points were spaced at 2.5 mm intervals in both the radial and axial directions.](image)

Figure 3.2 also identifies the difference between using a 3 second fit and a 5 second fit. SAR values calculated using a linear least squares fit to the first 5 seconds of temperature data were 45–50% lower than those calculated using the 3 second linear fit. If the 5 second fit values are taken as the 'correct' SAR values, then the error associated with thermal diffusion and cooling artifact for longer heating durations is significantly reduced for shorter heating times (45%, 70% and 90% for the 10, 20, and 'cooled' 20 second fits respectively).

Normalisation of the SAR profiles of figure 3.1 resulted in the profiles shown in
Figure 3.2: Variation of Absolute SAR With Radial Distance From Antenna Axis. The percentage decrease in SAR relative to the 3 second fit value at a point 5 mm from the antenna tip axially (towards the feedline) for a 10 W power pulse applied to the helical antenna with duration indicated by the legend, where 'cooled' corresponds to a 20 s pulse followed by a 5 s cooling period.

Figure 3.3: Variation of Absolute SAR With Applied Power Level. The percentage decrease in SAR relative to the 3 s fit value at a point 5 mm from the antenna tip axially (toward the feedline) for a 'cooled' 20 s 10 W power pulse applied to the helical coil antenna. Radial distance from the antenna axis is indicated by the legend.
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Figure 3.4: Normalised Dornier Thermocouple Temperature Profiles. Normalised radial and axial temperature profiles of the helical coil antenna for a 10 W/20 s power pulse calculated using a 3 s fit (black line), a 20 s fit (dark grey), and a 'cooled' 20 s fit (light grey). Thermocouple measurement points are spaced at 2.5 mm intervals in both the radial and axial directions.

Figure 3.5: Dornier Thermocouple IsoSAR Contour Plots. SAR contour plots of the applicator plane calculated (a) by a 3 s fit, and (b) after a 20 s heating duration followed by 5 s of cooling. Temperature measurement locations correspond to the intersection points of the axial and radial grid lines.
At a radial distance of 2.5 mm from the antenna axis, the magnitude of the SAR was reduced to less than 20% of its value for the 3 second fit. By contrast, the 20 second fit had a magnitude of almost 30% of maximum at this point, and the 'cooled' fit had a magnitude of 60%, indicating a widening of the temperature profile. The axial profile showed a similar widening, with relative reductions in SAR amplitude of 46%, 77%, and 54% for the 3 second, 20 second and 'cooled' 20 second fits respectively, at a distance of 15 mm away from the applicator (towards the feedline). The widening of the SAR contours is even more apparent in the normalised isoSAR contour plots (figure 3.5), where the applicator effective field area was found to increase with heating duration for all contour levels. EFA\textsubscript{50} (the area of the 50% SAR contour) increased by a factor of 1.4 over a 20 second heating period, relative to the EFA\textsubscript{50} of the 3 second fit, and increased an overall factor of 4.3 following an additional 5 seconds of cooling.

**IsoSAR contour plots of the BSD multisection antenna for a 10 W power pulse of 30 second duration are shown in figure 3.6 for a 3 second and 'cooled' 30 second fit. Based on the 3 second fit data, the antenna hot spot was located on the antenna axis 17.5 mm axially from the antenna tip. The location of the hot spot based on the 'cooled' data was only 12.5 mm axially from the antenna tip. Results of the thermocouple diffusion experiments for the BSD multisection applicator were similar to those of the helical coil antenna. The magnitude of the SAR at the hot spot was significantly reduced even after relatively short heating durations, falling by 78% for a 10 second fit, 84% for a 20 second fit, and 92% after a 30 second heating period. After an additional 5 second cooling period, the value was further reduced so that the maximum SAR was 98% lower than that calculated by the 3 second fit, which is of the same order of magnitude as the reduction seen for the helical antenna for a 20 second heating period followed by 5 seconds of cooling.**

As with the helical antenna, the percentage loss of SAR magnitude was found to decrease with increasing radial distance from the applicator axis (figure 3.7). The values in the figure were calculated at an axial distance corresponding to the 3 second fit hot spot location (17.5 mm from the antenna tip). It was not possible to accurately measure the axial variation of SAR percentage loss because of the shift in the location of the hot spot.
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Figure 3.6: BSD Thermocouple IsoSAR Contour Plots. SAR contour plots of the applicator plane of the BSD applicator calculated by (a) a 3 s fit, and (b) a 'cooled' 30 s fit. Temperature measurement locations correspond to the intersection points of the axial and radial grid lines.

Figure 3.7: Variation of Absolute SAR With Radial Distance From Dipole Antenna Axis. The percentage decrease in SAR relative to the 3 s fit value at a point 5 mm from the antenna tip axially (towards the feedline) for a 10W power pulse with duration indicated by the legend, where 'cooled' corresponds to a 20 s pulse followed by a 5 s cooling period.
Figure 3.8: Sample Thermograph and Surface Temperature Plot. (a) A thermograph of the helical coil antenna recorded from the IR camera for a heating duration of 20 s at an applied power level of 10W. (b) Surface temperature plot calculated from the thermograph using the calibration data for the imaging system.


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3.1.2 Infrared Imaging Experiments

Helical Coil Antenna

A typical thermographic image of the applicator plane of the helical coil antenna is shown in figure 3.8(a) for a heating duration of 20 seconds. The grey-scale levels were calibrated such that 160 colours represented the maximum rise of the temperature scale, in this case 5 degrees. The maximum grey-scale level in figure 3.8(a) after subtraction of the background image was 147, which represents a temperature rise of 4.6°C. The surface image shown in figure 3.8(b) results from temperature and scale calibration, and down-sampling\(^1\) to a 1 mm\(^2\) grid spacing. The antenna hot spot was located on the antenna axis 15 mm axially from the antenna tip.

Figure 3.9: Dornier IsoSAR Contour Plots. IsoSAR contour plots of the applicator plane of the Dornier helical coil antenna for a 10 W power pulse of duration (a) 20 s, and (b) 60 s, normalised to the maximum SAR value on the plane.

IsoSAR contour plots of the applicator plane of the Dornier antenna are shown in figure 3.9 for a 10 W power level and heating durations of 20 and 60 seconds. The expansion of the isoSAR contour areas of the 60 second plot relative to the 20 second plot is immediately apparent, the greatest difference occurring for the lower (less than 60%) contours. Effective field area was found to increase significantly with heating duration.

\(^1\)Down-sampling was accomplished by bilinear interpolation using the Matlab INTERP2 function.
plot of the EFA$_{50}$ as a function of heating duration for a 9 W power pulse (figure 3.10(b)) showed an increase of $175\pm 50\%$ for a 20 second heating duration, $150\pm 70\%$ for a 30 second duration and $230\pm 70\%$ after 60 seconds of heating, relative to the EFA$_{50}$ of the 10 second heating duration. For the same heating intervals, the EFA$_{70}$ increased $100\pm 40\%$, $150\pm 65\%$ and $220\pm 80\%$ respectively, relative to the EFA$_{70}$ of the 10 second duration.

The peak amplitude of the SAR pattern at the antenna hot spot was found to decrease significantly with increasing heating duration (figure 3.10(a)). After a 20 second heating duration, the SAR value at the hot spot decreased by $18\pm 10\%$ relative to the 10 second value, dropping to $22\pm 18\%$ after 30 s, and $37\pm 13\%$ after a 60 second heating period.

Figure 3.10: Thermal Diffusion Effect on Dornier SAR Pattern (9W). The effect of thermal diffusion on (a) SAR pattern peak amplitude, and (b) EFA$_{50}$ (light grey), EFA$_{70}$ (dark grey), and EFA$_{90}$ (black).
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Multisection Dipole Antenna

IsoSAR contour plots of the applicator plane of the BSD multisection dipole antenna are shown in figure 3.11 for heating durations of 30 and 60 seconds. The applicator hot spot was located on the antenna axis, 37 mm axially from the tip of the antenna. No increase in effective field area was observed for the two SAR patterns. EFAs was actually 18% smaller than the 30 second value after a 60 second heating period. As with the helical coil experiments, SAR amplitude at the hot spot was found to decrease with increasing heating duration, with a 22% drop observed after a 60 s heating duration, relative to the 30 second SAR value.

![Figure 3.11: BSD IsoSAR Contour Plots](image)

Figure 3.11: BSD IsoSAR Contour Plots. IsoSAR contour plots of the applicator plane of the BSD multisection dipole antenna for a 10 W power pulse of duration (a) 30 s, and (b) 60 s, normalised to the maximum SAR value on the plane.

Ultrasound Applicator

IsoSAR contour plots of the applicator plane of the interstitial ultrasound applicator are shown in figure 3.12 for heating durations of 10 and 45 seconds, for a 7 W power application to a single transducer element. The applicator hot spot was 23 mm from the antenna tip, which corresponds to the centre of the transducer element. The isoSAR contour plots in figure 3.12 indicate considerable expansion of the isoSAR contour areas of the applicator.
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Figure 3.12: Ultrasound Applicator IsoSAR Contour Plots. IsoSAR contour plots for a 7 W power pulse of duration (a) 10 s, and (b) 45 s, normalised to the maximum SAR value on the plane. Only a single element of the ultrasound applicator was activated.

After a 20 second heating period, the EFA50 was 39% greater than the EFA50 measured after 10 second of heating duration, and 110% greater after a 45 second heating interval, again relative to the 10 second heating duration. As with the microwave applicators, the magnitude of the SAR value at the antenna hot spot was found to decrease with increasing heating duration. The maximum SAR value was reduced by 53% (relative to the value calculated for a 10 second heating duration) after a 20 second heating duration, and by 71% after a 45 second heating period.

3.2 Surface Cooling

The effect of the surface cooling artifact on thermographically acquired SAR patterns was estimated using a first order least squares fit to the cooling curve at each point in the thermal image (figure 2.10). The temperature distribution prior to convective surface cooling was approximated using data acquired during the cooling period (section 2.7.4). The surface cooling artifact contributed to an overall reduction in SAR magnitude, but did not result in an increase in applicator effective field area. Sample radial and axial temper-
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Figure 3.13: Surface Cooling Effect on Temperature Profile Amplitude. Temperature profiles before (black line) and after (gray line) surface cooling of the imaging plane, for a 10 W power pulse of 60 s duration. The difference in amplitude is shown by the dashed line.

Figure 3.14: Surface Cooling Effect on Maximum Temperature Rise. Maximum temperature rise before (gray line) and after (black line) surface cooling of the imaging plane as a function of heating duration. The difference between the two is indicated by the dashed line.
ature profiles for both the cooled and corrected ('uncooled') temperature data are shown in figure 3.13 for a 10 W power pulse of 60 second duration. A decrease in temperature was observed along both the axial and radial profiles. The magnitude of the decrease was found to increase with heating duration, as can be seen in figure 3.14. The temperature decrease was approximately 20% of the uncooled value for all heating durations. The shape of the normalised axial and radial SAR profiles remained essentially unchanged despite the change in profile magnitude, so that the normalised SAR patterns were only minimally affected by the surface cooling artifact.

3.3 Applicator Coupling

The effect of the method of coupling of the helical coil applicator to the phantom material was investigated for two types of coupling: 1) the applicator catheter directly embedded in the middle of a phantom sheet, and 2) the applicator catheter placed in a well formed between two adjacent phantom sheets. A comparison of the SAR patterns acquired for the two types of coupling indicated that the embedded helical applicator had considerably more relative heating along the applicator feedline than the welled applicator for heating durations greater than 30 seconds.

Figure 3.15: Effect of Applicator Coupling on SAR Pattern Peak Amplitude. Peak SAR amplitude as a function of power pulse duration for an embedded (black line) and welled (grey line) applicator, for an applied power level of 10 W.
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Applicator coupling was found to have a considerable effect on both the magnitude and shape of the radial and axial temperature profiles. The peak SAR value at the antenna hot spot for the embedded applicator was substantially lower than that of the welled applicator for all heating durations (figure 3.15). After a 10 second heating interval, the embedded applicator hot spot SAR value was 54±7% lower than the hot spot SAR value for the welled applicator. The location of the applicator hot spot remained fixed at approximately 11 mm from the welled applicator tip for varying heating duration, but shifted considerably for the embedded applicator, ranging from 1 to 15 mm from the tip, as indicated in figure 3.16.

![Figure 3.16: Location of Hot Spot Relative to Antenna Tip. The location of the hot spot of the embedded (black line) and welled (grey line) applicator as a function of heating duration.](image)

Normalised axial temperature profiles of both the welled and embedded applicators are shown in figure 3.17 for a heating duration of 60 seconds. After 10 seconds of heating, the shapes of the profiles were essentially the same from the hot spot back towards the feedline3.17(a)). After a 30 second heating interval, there was a significant difference in the shape of the profile back towards the feedline, where the embedded applicator had approximately 20% more heating relative to the hot spot than the welled applicator (figure 3.17(b)). The divergence of the axial profiles was even more evident after 60 seconds (figure 3.17(c)), where the embedded applicator had an average 20% more relative heating as close as 10 mm axially from the antenna hot spot. This suggests that coupling has a considerable effect on SAR patterns acquired for heating durations greater than 30 seconds. No significant difference was observed between the normalised radial profiles of the welled
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Figure 3.17: Effect of Applicator Coupling on Axial Temperature Profile. Axial temperature profiles for an embedded (black line) and welled (gray line) applicator, for a 10 W applied power pulse of duration (a) 10 s, (b) 30 s, and (c) 60 s.
and embedded applicators for any heating duration.

### 3.4 Phantom Layering

The effect of the phantom layer thickness on the SAR patterns of interstitial applicators was investigated for phantom layer thicknesses of 2.5, 5, and 10 mm. It was hypothesised that thinner phantom layers would distort the SAR pattern because of the air trapped between the layers. Little difference was found between the SAR patterns of all three phantom layer thicknesses.

![Figure 3.18: Effect of Layer Thickness on Axial and Radial Temperature Profiles. (a) Radial, and (b) axial temperature profiles of the default profile planes of the helical coil antenna for a 10 W/40 s power pulse for phantom layer thicknesses of 2.5 mm (black line), 5 mm (dark grey), and 10 mm (light grey).](image)

No significant difference was found between either the axial or radial temperature profiles measured for the three phantom layer thicknesses. Axial and radial temperature profiles for the helical coil antenna are shown in figure 3.18 for a 10 W power pulse of 40 second duration. It is evident from these profiles that there was no significant difference in either profile shape or magnitude after a 40 second heating duration. The differences between the profiles of the three thicknesses were well within the repeatability limits of the experiment.

The effective field areas of the three phantom layer thicknesses did not change
substantially with layer thickness. The average coefficient of variation between the EFA$_{50}$ of the three layer thicknesses was 11% for heating durations of 10 and 40 seconds, and 16% for the 20 second heating duration. Mean variation at lower contour levels was higher for the shorter heating times, reaching a 25% mean variation for the 10 second and 20 second heating durations, but only a 15% mean variation for the 40 second duration.
Chapter 4

Discussion

4.1 Thermal Diffusion and Cooling Artifacts

The expansion of the isoSAR contour areas and the decrease in maximum SAR value observed in the thermal images for increasing heating duration are explainable by heat transfer theory. Consider the temperature distribution over time at a point \( P(z, y) \) near the hot spot on the surface of the phantom \( (z = 0) \). An idealised temperature distribution is shown in figure 4.1. When the heat source is on (Region I), heat is conducted down the thermal gradient because of thermal diffusion, resulting in a non-linear temperature distribution. If the heat source is on for a sufficient period of time, the temperature will eventually reach a steady-state value when the rates of heat generation and heat loss at the point are equal (Region II). The instantaneous slope of the temperature distribution at that time would be zero. When the heat source is turned off (Region III), heat continues to diffuse down the thermal gradient, but no additional heat is generated to replace the lost heat. The result is rapid cooling.

In an infrared thermography experiment, the surface of the phantom is exposed at some time after the end of heating in order to image the desired plane. Once the phantom surface is exposed, heat is transported from the surface by convective air flow, in addition to the conductive heat transfer that continues to occur within the phantom. Depending on the rate of heat transfer to the ambient air, the rate of cooling may increase significantly during this period (Region IV). Thermal images acquired in an infrared thermographic SAR phantom study provide the spatial temperature distribution of a phantom surface at a time after the phantom surface has been exposed, corresponding to a time in Region IV of figure 4.1. As a result, SAR values calculated from this data are subject to error due to
both thermal conduction and convection.

![Idealised Temperature Distribution At a Point Near the Applicator Hot Spot.](image)

Figure 4.1: Idealised Temperature Distribution At a Point Near the Applicator Hot Spot. As the duration of heating increases, the slope $\frac{\Delta T}{\Delta t}$ (lines 2-4) deviates increasingly from the slope $\frac{dT}{dt}_{t=0^+}$ (line 1). When the phantom surface is exposed (Region IV), thermal convection results in additional cooling and a further decrease in $\frac{\Delta T}{\Delta t}$ (line 5).

The error in SAR measurement associated with infrared thermography results from the non-linear shape of the temperature distribution of figure 4.1. Since the only information available from an IR experiment is the initial and final spatial temperature distribution, the SAR at a point $P(x,y)$ on a phantom surface must be calculated using equation 1.8. If the temporal temperature distribution is linear, then $\frac{\Delta T}{\Delta t} = \frac{dT}{dt}_{t=0^+}$ for any heating time $t_h$, so that equation 1.8 is identically equal to equation 1.7. Since the temporal temperature distribution is not linear, the slope $\frac{\Delta T}{\Delta t}$ will deviate from the instantaneous slope at $t = 0$ as shown in figure 4.1, points 1 and 2. As the heating duration is increased, $\frac{\Delta T}{\Delta t}$ becomes a poorer estimate of $\frac{dT}{dt}_{t=0^+}$ and the error in the measured SAR increases (point 3). If the duration of the applied power pulse is sufficiently long, the temperature will reach a steady state value. The error in measured SAR will continue
to increase because even though $\Delta T$ is constant, $\Delta t$ continues to increase as the heating duration is increased. Consequently, as heating time is increased, the magnitude of the SAR value will decrease.

When the applicator is turned off, the cooling that follows further increases the error associated with the measured SAR value because $\Delta T$ decreases, reducing the slope $\frac{\Delta T}{\Delta t}$, and, hence, the SAR magnitude still further (Region III of figure 4.1). The error associated with SAR values calculated in this region can be reduced somewhat by using the slope $\frac{\Delta T}{t_h}$, where $t_h$ is the duration of the applied power pulse. When the phantom surface is exposed (Region IV), the error is likely to increase even further, because the rate of cooling is increased due to rapid heat removal by convection. Since the error is dependent only on the temperature distribution in the medium, one would expect the error to be present regardless of the modality of heat delivery.

It is useful to distinguish between the conduction artifact associated with thermal conduction while the applicator is on, and the cooling artifact associated with both conduction and convection of heat once the applicator has been turned off, because they are present for different reasons. The conduction artifact is present in IR thermographic SAR patterns of interstitial applicators because power must typically be applied for a duration of up to 60 seconds in order to generate a good quality thermal image, ie. one with relatively little noise and high temperature resolution. The cooling artifact is present in the SAR pattern because of the manipulation delay between the end of the applied power pulse and acquisition of the thermal image, which results from the time needed to physically separate the phantom layers and remove the applicator to prevent it from continuing to act as a source of heat. Conductive and convective cooling take place during this delay, which is typically on the order of 5-8 seconds [9] [10].

At distances far from the applicator hot spot, the thermal gradient will be quite small and the temperature rise will be approximately linear with time while the applicator is on. When the applicator is turned off, the rate of cooling will be very slow because little heat diffuses across the shallow gradient. In fact, the temperature at these outlying points may continue to rise as heat is diffused outward from the applicator hot spot (figure 4.2). Since the effect of thermal diffusion varies with location, the shape of the SAR pattern becomes distorted. The area of the normalised isoSAR contours increases with time because heat is removed from regions near the hot spot and deposited in outlying regions, resulting in a flattening of the temperature profile. When the phantom surface is exposed, the temperature differential of the surface relative to the ambient air at a point far removed from the hot spot will be quite small. Consequently, the rate of convective
cooling at such points will be much slower than the cooling rate at the hot spot. Hence, the distortion of the SAR pattern will increase still further.

In analysing an IR thermographic SAR pattern, it is not possible to distinguish between the errors contributed by the diffusion and cooling artifacts respectively, because spatial temperature distributions are available for only two discrete points in time, $T(z, y, t_i)$, the initial background temperature distribution prior to heating, and $T(z, y, t_f)$, the temperature distribution after both diffusive and convective cooling. While the repeated IR thermographic experiments with varied heating durations demonstrated the presence of a significant combined diffusion and cooling artifact for heating times of less than 30 seconds, it is not possible to identify directly which is the greater source of error. It is possible, therefore, that there is significant error even in images acquired for a heating duration of 10 seconds because even though the diffusional artifact may be minimal, the cooling artifact may still be significant. Such a theory is supported by the findings of the thermocouple experiments, which indicated a substantial difference between the $\frac{dT}{dt}$ value after a 10 second heating interval and the approximated $\frac{dT}{dt}$ value of the initial 3 second fit at the applicator hot spot.
4.1.1 Thermocouple Experiments

Experiments were performed to investigate the thermal diffusion and cooling artifacts present in infrared thermographic SAR patterns for applicator heating durations of less than 60 seconds. Thermocouple experiments were designed to predict the error in the SAR measurement that resulted from calculating $\frac{\Delta T}{\Delta t}$ from initial and final temperature values rather than determining $\frac{dT}{dt}$ by a fit to the temporal temperature distribution. The observed decrease of the maximum SAR value as heating time was increased results from the poorer fit to the slope $\frac{dT}{dt}|_{t=0^+}$ as time progresses, as was shown in figure 4.1.

It is interesting to compare the maximum SAR values calculated using the 3 second linear least-squares fit of thermocouple data to the maximum SAR values provided in the literature for similar applicator designs. The 3 second fit SAR values for both the helical coil and multisection antennas are between one and two orders of magnitude greater than values reported by Ryan for similar antenna designs [7]. For example, for the BSD antenna, Ryan reports a maximum SAR value of 124 W/kg/W in a loose-fitting catheter, compared to a value of 3970 W/kg/W in our thermocouple experiments using the 3 second fit. However, the Dartmouth automated SAR measurement system [12] used by Ryan employs a second order polynomial least-squares fit of the form $T(t) = a_2t^2 + a_1t + a_0$, resulting in an instantaneous slope of $\frac{dT}{dt}|_{t_0^+} = 2t_0^+a_2 + a_1$, where $t_0^+$ is 10 seconds [7]. SAR is then calculated using equation 1.6. The differences between a 3 second fit and second order fit can be readily seen by examination of figure 4.3, which shows the temperature-time curves at the hot spot of the two antennas recorded during the thermocouple experiments for an applied power level of 10 W, and a second order polynomial least-squares fit to the data points. It is evident that the slope of the second order fit is substantially less steep than the slope of the actual time-temperature curve at $t = 0$, resulting in the reduced SAR values reported by Ryan [7]. Recalculating the maximum SAR value from the dipole thermocouple experiments using a quadratic fit results in a value of 300 W/kg/W, which is still twice that reported by Ryan, but of the same order of magnitude [7]. The difference is likely due to the fact that Ryan used fibreoptic probes seated in catheters [7], while our thermocouples were embedded directly into the phantom material, resulting in higher recorded temperatures and, consequently, higher SAR values.

Satoh et al. [8] report peak SAR values for helical coil antennas which are an order of magnitude lower than those reported by Ryan [7]. The SAR values were calculated using data recorded at 10 second intervals over a 30 second heating period and (presumably) a linear fit. One possible explanation for the different magnitudes is that thermal diffusion
has resulted in a significant reduction in measured SAR during the 30 second heating period. Such an explanation is supported by the results of the present study. In any event, it is evident that the choice of fitting function can have a substantial effect on the magnitude of calculated SAR values. A linear least-squares fit to the initial 5 seconds of the temporal temperature distribution produced SAR values that were, on average, approximately 50% lower than values calculated from a 3 second least-squares fit, although the normalised SAR pattern remained essentially unchanged. This suggests that it would be inadvisable to compare SAR patterns calculated using different fitting functions for anything other than a qualitative determination of SAR pattern shape.

4.1.2 Infrared Imaging Experiments

Experiments were performed to measure the effect of thermal conduction and surface cooling on the SAR patterns measured by an infrared thermographic technique using an applied power pulse duration of 60 seconds or less. Expansion of the normalised isoSAR contour areas and a reduction in maximum SAR indicated the presence of conductive and convective cooling artifacts within the first 30 seconds of heating. Artifacts were found in
the measured SAR patterns of both interstitial microwave and ultrasound applicators.

SAR patterns reported by Ryan for helical coil and multisection antennas compare favourably with our results for a 10 second heating duration, but poorly for heating durations of 30 seconds or greater [7]. For a BSD multisection antenna, with a 'snug fit' catheter of approximately the same dimensions as the one used in the infrared imaging experiments, Ryan reports an applicator hot spot location of 37 mm from the applicator tip, and a 20% isoSAR radial contour width of approximately 15 mm at the hot spot [7]. After a 30 second heating duration, the hot spot of the thermographic SAR pattern was also located at a distance of 37 mm from the applicator tip. However, the radial width of the IR thermographic 20% isoSAR contour was approximately 40 mm. For the helical coil applicator, Ryan reports a 20% radial contour width of only 9 mm, while the thermographic pattern had a radial width of 14 mm for a 30 second heating time, and 13 mm for a 10 second heating time [7]. However, the 80% isoSAR contour width reported by Ryan is approximately 1.5 mm, while the 10 second thermographic SAR pattern had an 80% heating width of 1.75 mm, and the 30 second image had a width of 5.75 mm. It should be noted that the helical applicator used by Ryan had a 1 cm longer coil length and one half the coil turn density of the applicator used in the present study, so that a direct comparison of the findings of the two studies is inadvisable. However, Satoh et al. report little variation in hot spot location and maximum SAR for helical applicators with coil lengths of 2.5 and 3.5 cm and different coil turn densities [8].

Surowiec et al. report a maximum SAR value of 13.8 W/kg/W for the BSD MA-251 multisection applicator\(^1\) after a 60 second heating period using an infrared thermographic SAR measurement technique [10]. Results of the present study indicate a maximum of 26 W/kg/W for the BSD MA-250 applicator for the same heating duration, and a maximum of 33.3 W/kg/W for a 30 second heating interval. However, Surowiec et al. are unclear on the method of catheter coupling to the medium, which may account for the difference between the two findings.

A comparison of the results of the thermocouple and infrared imaging experiments indicates differences between the SAR patterns calculated from the thermocouple data, and those measured from the infrared images. Since the data in the radial direction is very sparse, it would be inappropriate to compare isoSAR contours. In fact, the only legitimate comparison that can be made between the experiments is for 'cooled' thermocouple data, since there is always a 5 second period of cooling in the thermal images. For the helical

\(^1\)The MA-251 is 0.08 mm wider than the MA-250 because it contains integrated thermometry, but is otherwise identical to the MA-250
coil applicator, the location of the applicator hot spot was 10 mm closer to the applicator tip for the fitted thermocouple data when compared with the infrared image data. For the multisection dipole antenna, the hot spot location was 19–25 mm farther away from the applicator tip for the thermocouple experiments. This difference in hot spot position can be accounted for by the different catheter materials used in the experiments. Ryan has shown that the hot spot location of a helical coil applicator shifts away from the antenna tip for larger diameter catheters, while the shift is towards the antenna tip for multisection antennas [7]. Since the catheters used in the helical coil thermocouple experiments were narrower than those used in the IR experiments, the shift in hot spot location of the helical coil antenna agrees with the literature. Although the interior diameters of the catheters used with the multisection dipole antenna were the same for both IR and thermocouple experiments, both the exterior diameter and the catheter composition were not. Given the shift in hot spot location, it appears likely that the SAR pattern is affected not only by the dimensions of the catheter, but by other catheter properties as well. Array studies conducted by James et al. found that higher peak SAR values were measured with thicker walled catheters [3]. Hence it is clear that both inner and outer catheter diameter have an effect on the SAR pattern.

SAR values calculated for the thermocouple experiments were significantly higher than those measured in the IR experiments even after accounting for thermal diffusion effects before and after heating. For the helical coil applicator, for a 20 second heating duration followed by a 5 second cooling period, the maximum SAR for the thermocouple experiments was 307 W/kg/W, compared to a value of 84±16 W/kg/W for the IR experiments. A similar difference was found for the multisection dipole applicator. The disparity is likely again explainable by the difference in catheter diameter. Ryan report higher maximum SAR values for helical coil applicators and lower maximum SAR values for multisection dipole applicators in thin, 'snug-fitting' catheters, such as those used in the thermocouple experiments [7], when compared with applicators in wider catheters, such as those used in the imaging experiments. Another consideration is that convective cooling may play a greater role than diffusion in the cooling artifact. Since no convective cooling took place in the phantom interior during the thermocouple experiments, one would expect correspondingly higher SAR values at the hot spot, because of the reduced effect of the manipulation delay.
4.1.3 Surface Cooling Artifact

There are two possible ways to reduce the artifacts present in the IR thermographic SAR patterns: 1) Reduce the heating duration, which will bring \( \frac{\Delta T}{\Delta t} \) and \( \frac{\Delta T}{\Delta t} \bigg|_{t=0^+} \) closer together, thereby reducing the error in the SAR approximation, and 2) Reduce the delay between the end of heating and image acquisition, thereby reducing the length of the conductive and convective cooling period after the heat source is turned off. The latter alternative would likely require some form of automated manipulation device, as it was not found possible to reduce the manipulation delay to less than approximately 5 seconds, which still results in substantial cooling. Hence, a cooling artifact will always be present in infrared thermographic SAR images. The former alternative is accomplished by reducing the length of the applied power pulse. However, there is a practical limit on how short the applied power pulse can be because the thermal images that result from short heating times are very noisy due to the local fluctuations in background temperature, which are on the order of 1–2°C and may be considerably higher than the temperature rises induced by a short heating pulse. Increasing the applied power level would likely overcome the noise problem. However, many commercial interstitial microwave applicators have maximum input power levels below 15 W, and varying the power level between 5–10 W was found to have little effect on the SAR error.

Calculation of the temperature distribution prior to surface cooling using a first order fit to the temperature distribution during the 10 second period following phantom surface exposure indicated no distortion of the SAR pattern, and a decrease in maximum SAR of approximately 20% for all measured heating durations. These results suggest that most of the distortion of the SAR pattern occurs during heating, and not during cooling. Consequently, it could be argued that, provided that the heating duration is short enough so that the conduction artifact is kept minimal, the normalised infrared thermographic SAR pattern is an accurate representation of the applicator power deposition pattern. It is important to recall, however, that the 'uncooled' temperature distribution was calculated based on a first order approximation. Analysis of the thermocouple results indicates that the rate of cooling within a 10 second period following the end of heating is not constant, which was an assumption of the first order approximation. In fact, the expansion of the radial SAR profile during the 5 second cooling period is evident in the result of the helical antenna thermocouple experiments (figure 3.4). Hence, the only conclusion that can be drawn from the approximation is a lower limit on the effect of surface cooling, i.e. cooling certainly affects the SAR amplitude within a 5 second period, and is likely to have an
CHAPTER 4. DISCUSSION

effect on the shape of the pattern as well.

4.2 Applicator Coupling

The properties of an antenna embedded in a material medium differ from the properties of an antenna embedded in air because they are dependent on the electrical characteristics of the medium [4]. To reduce the sensitivity of the current distribution along the antenna to the electrical properties of the medium, hyperthermia applicators are typically inserted into catheters. The catheter acts as an insulating sheath, reducing current leakage from the antenna into the medium. The sheath also acts as a pathological barrier, which is of considerable clinical importance due to the invasiveness of an interstitial hyperthermia treatment.

The characteristic impedance of a microwave hyperthermia applicator may vary considerably depending on the properties of the insulating sheath. Ryan reports significant differences in the magnitude and shape of SAR patterns of applicators situated in loose and snug-fitting catheters [7]. James et al. report changes in maximum SAR for thin and thick walled catheters [3]. For this reason, hyperthermia quality assurance guidelines suggest that measurement of applicator SAR patterns take place in the same type of catheter intended for clinical treatment [2].

The insulating sheath is only one consideration in applicator coupling. In an interstitial microwave hyperthermia treatment, the catheter and antenna are embedded directly into a tumour to produce localised heating. To reproduce this coupling in a thermographic phantom study using a solid phantom material, the catheter is usually embedded into a phantom sheet prior to polymerisation. Consequently, thermographic analysis must rely on images acquired either above or below the applicator plane, which are not an accurate representation of the power deposition on the applicator plane, as explained in section 1.7.3.

Splitting the phantom at the applicator plane, and coupling the antenna catheter to the phantom by inserting it into a small well, resulted in differences between the welled SAR pattern and that observed for the embedded applicator (section 3.3). The differences result from the change in the characteristic impedance of the applicator between the two types of coupling. One particular point of interest is the greater heating along the applicator feedline observed for the embedded applicator, which may be due to resistive heating of the feedline. If the helical antenna was both resistively and radiatively heating the medium (due to imperfect design), one would expect to see greater heating along the embedded applicator, since conductive heat transfer through the phantom is greater than through
the air gap around the welled applicator. The applicator feedline was, in fact, hot to the touch during power delivery, which does indicate some resistive heating.

The shift in location of the applicator hot spot is more difficult to account for. The fact that there is a difference in the hot spot location for the welled and embedded applicators is not surprising, considering the difference in coupling. The case of an embedded and welled applicator can be compared to the case of a snug and loose-fitting catheter, in that the one type of coupling involves a greater air gap around the applicator than the other. The only difference between the two cases is the location of the catheter wall, which is at the outside of the air gap for the loose-fit catheter, and somewhere in the middle of the air gap for the welled applicator. For the snug and loose-fit catheters, Ryan reports a shift of the applicator hot spot away from the antenna tip for the loose-fitting catheter [7]. Correspondingly, one would expect to see a shift of the hot spot away from the antenna tip for the welled applicator, which was seen in figure 3.16 for heating times greater than 30 seconds. However, for heating times below 30 seconds, a shift in the opposite direction is observed.

The most surprising aspect of figure 3.16 is the shift of the location of the embedded applicator hot spot towards the antenna tip for increasing heating duration. One possibility is that the physical properties of the phantom are altered over time because of repeated use. Another possibility is that the phantom properties are temperature dependent. However, these theories do not account for the fact that no shift is observed for the welled applicator. The observed hot spot location shift may be due to the fact that the imaging plane was 2.5 mm above the applicator, rather than on the applicator plane. Hence, the observed shift in hot spot location may be due to conduction and cooling artifacts, rather than an actual shift in the hot spot location on the applicator plane. In any event, it is evident that the choice of coupling mechanism has a substantial effect on the measured SAR pattern.

4.3 Phantom Layering

The power deposition pattern of an insulated microwave antenna in a material medium is affected not only by the electrical characteristics of the medium, but by the shape and size of the medium as well [4]. For an infinite homogeneous isotropic medium, such as a large, solid block of phantom material, the effects of the shape and size of the medium on the antenna properties can be reasonably ignored, because boundaries are too far away to be of consequence. However, infrared thermographic SAR measurement requires phantom materials which are composed of thin layers in order to allow planar surfaces to be exposed
CHAPTER 4. DISCUSSION

for thermal imaging. Provided that the dimensions of the phantom sheet are much greater than the penetration depth of microwaves in the medium, the phantom can be considered infinite. However, because the sheets are very thin (less than 10 mm), they cannot be considered infinite on a plane perpendicular to the phantom surface. Consequently, electrical discontinuities at the interfaces between adjacent phantom layers can potentially lead to distortion of the shape and magnitude of the SAR pattern [6]. Such distortions are unpredictable because air may become trapped between adjacent phantom layers.

As the thickness of each layer within a block of phantom material is increased, fewer layers are required to form the block, resulting in fewer boundary interfaces and, hence, fewer electrical discontinuities. For example, a 2 cm thick block formed from eight 2.5 mm thick sheets would have seven boundary interfaces between adjacent layers, while one formed with 10 mm thick sheets would have only one interface. Since the additional of each new layer increases the likelihood of an electrical discontinuity, one would expect poorer experimental repeatability using thinner phantom layers.

The only advantage to using thin layers is to allow imaging of several different phantom planes without having to construct a separate phantom block for each experiment. For external hyperthermia applicators, which have large field areas, phantom sheets are typically at least 1 cm in thickness [9]. To characterise an interstitial applicator, which has a much smaller field pattern, higher spatial resolution is required so significantly thinner phantom sheets are required. Characterisation of a single applicator would not ordinarily require more than an image of the applicator plane, since symmetry about the antenna axis can usually be assumed. For such cases, the phantom block need only be split on the applicator plane, and the only effect of the electrical discontinuity will be on the coupling of the applicator to the medium. In measuring the SAR patterns of interstitial applicator arrays, symmetry cannot always be assumed, hence the ability to image on more than one plane is still desirable.

It is evident from the results of the layering experiments that layer thickness does not have a significant effect on the SAR patterns of the applicator plane. On planes above or below the applicator, the magnitude of the SAR pattern is very small, because of the rapid drop-off in energy deposition with radial distance from the antenna. As a result, it is difficult to make a meaningful comparison between SAR patterns on such planes. Consequently, it remains unclear whether the SAR pattern is distorted on planes other than that of the applicator.
4.4 Significance of the Results

It is evident that the SAR patterns of interstitial hyperthermia applicators measured using an infrared thermographic technique do not accurately reflect the power deposition pattern of the applicator because of thermal conduction and convection artifacts present when heating durations exceed 10–20 seconds. The heating duration should, therefore, be restricted to this time frame for the characterisation of interstitial applicators. In addition, the SAR patterns of the applicator plane acquired using a welled catheter technique may underestimate the 50% heating length along a microwave applicator feedline by as much as 35%.

The results of the present study have implications beyond infrared thermographic analysis. Liquid crystal thermography, which was described in section 1.7.2, has been proposed as an alternative to both infrared thermography and discrete thermometry systems because it combines high spatial resolution with accurate temperature measurement. The quantitative colourimetric analysis of LCT described by Cristoforetti et al. would not be expected to exhibit any of the sources of variation described in the present infrared thermographic study, because imaging is performed through a transparent phantom [1]. Consequently, the technique requires no phantom layering and thermal conduction effects can be accounted for using time-evolved temperature data. For LCT techniques that do not use a transparent phantom and, therefore, require surface imaging of the liquid crystal plate, such as the technique described by Labonté et al. [5], thermal conduction and cooling artifacts may occur.

4.5 Summary and Conclusions

The OCI/PMH Thermal Imaging System was modified for use with interstitial hyperthermia applicators. The modified system was used to study the effect of thermal conduction and convection artifacts, applicator coupling, and phantom layering on the measurement of SAR patterns of interstitial microwave applicators. Cooling effects on interstitial ultrasound applicators were also investigated. Thermal conduction and cooling artifacts were found to decrease the magnitude and expand the isoSAR contours of the SAR pattern of both microwave and ultrasound applicators, for heating durations greater than 10–20 seconds. The type of coupling (welled or embedded) of the applicator to the phantom material was also found to affect the SAR pattern. Variation of phantom layer thicknesses between 2.5 mm and 10 mm produced no effect on the SAR pattern. It was concluded
that infrared thermographic SAR measurement of interstitial microwave and ultrasound applicators does not produce accurate SAR patterns if the duration of the heating time exceeds 10–20 seconds, and may not be accurate even for heating durations of less than 10–20 seconds due to the presence of surface cooling artifacts.

4.6 Suggestions for Future Work

Improving the accuracy of the infrared thermographic SAR measurement system described in this thesis would be difficult because of the limitations posed by the measurement technique itself. The use of a digital thermal camera with higher temperature resolution may allow for a reduced applicator heating duration and, consequently, a reduced thermal conduction artifact. However, reduction of the surface cooling artifact is not possible without automating the imaging procedure, including phantom manipulation. The design of such a device would be complicated by the necessity to both expose the imaging surface and remove the applicators without damaging the fragile phantom layers.

If the manipulation delay could be significantly reduced, the possibility exists of calculating SAR from the falling edge, rather than the rising edge of the power pulse. Equation 1.7 is valid provided that the phantom is in a state of thermal equilibrium prior to measurement of $\frac{\partial T}{\partial t}$. Such a state exists not only prior to power application, but also at some time $t$ after a sufficiently long heating duration. Consequently, SAR can also be calculated by waiting for the temperature distribution within the phantom to reach steady state, turning off the applicator, and measuring the initial rate of cooling in the phantom. The primary difference between such a technique and the technique described in this thesis, is that there is no restriction on imaging once the phantom surface has been exposed. Hence, it would be possible to acquire several thermal images each second following the applied power pulse and measure $\frac{\partial T}{\partial t} \bigg|_{t=0}$ by an appropriate fit to the time-temperature curve at each point in the phantom. The question remains, however, as to what effect surface convection would have on SAR patterns measured in this fashion.

Given the limitations presently posed by infrared thermography, other thermographic SAR measurement techniques seem to be more practical and more appropriate for interstitial applicator characterisation. Magnetic resonance imaging, for example, has been used successfully to monitor temperature changes in phantom materials [11]. Such a technique could be applied to SAR measurement, but poses limitations on applicator materials and is not cost-effective when compared with liquid crystal films or fibreoptic thermometry systems. It is evident that a comprehensive comparison of available SAR
measurement systems would be quite useful in ensuring that the measured SAR patterns are an accurate representation of power deposition.

References


REFERENCES


IMAGE EVALUATION
TEST TARGET (QA-3)

150mm

6"

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