Thermal simulation of breast tumors

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It is well known that differences in energy consumption exist for normal and cancerous tissue. These differences lead to small but detectable local temperature changes, which is why infrared imaging has been used in the detection of different types of cancer; however, the early instrumentation was not sensitive enough to detect the subtle changes in temperature needed to accurately diagnose and monitor the disease. In recent years the sensitivity of infrared instruments has greatly improved. In this paper the bioheat transfer equation is solved for a simplified model of a female breast and a cancerous tumor in order to quantify the minimum size of a tumor or the maximum depth of a certain sized tumor that a modern state-of-the-art infrared imaging system can detect. Finite Element simulations showed that current state-of-the-art imagers are capable of detecting 3 cm tumors located deeper than 7 cm from the skin surface, and tumors smaller than 0.5 cm can be detected if they are located close to the surface of the skin.

Keywords: Cancer simulation; thermopathology; bioheat equation; thermal simulation.

Es bien conocido que existen diferencias en consumo de energía entre tejido normal y tejido canceroso. Estas diferencias generan pequeños cambios en la temperatura que pueden ser detectables; es por eso que se ha tratado de utilizar la termografía infrarroja para detectar diferentes tipos de cáncer. Los primeros trabajos al respecto utilizaban cámaras termográficas que no eran lo suficientemente sensitivas para detectar los pequeños cambios en la temperatura necesarios para monitorear y diagnosticar efectivamente esta enfermedad. En años recientes la sensitividad de las cámaras termográficas ha aumentado notablemente; en este trabajo se resolvió la ecuación de transferencia de calor en tejidos biológicos para un modelo simplificado de un seno femenino y un tumor canceroso con el objetivo de cuantificar el tamaño mínimo y la profundidad máxima de un tumor que puede ser detectado con un sistema moderno de termografía infrarroja. Las simulaciones de elemento finito demostraron que los sistemas termográficos modernos pueden detectar tumores de 3 cm de diámetro localizados a una profundidad mayor a 7 cm y tumores menores a 0.5 cm si se encuentran cerca de la superficie de la piel.

Descriptores: Cáncer de seno; simulación térmica; método del elemento finito.

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1. Introduction

All objects with a temperature above absolute zero emit infrared radiation from their surface. The Stefan-Boltzmann Law defines the relation between radiated energy and temperature by stating that the total radiation emitted by an object is directly proportional to the object's area, its emissivity, and to the fourth power of its absolute temperature. Since the emissivity of the human skin is extremely high, within 1% of that of a black body, measurements of infrared radiation emitted by the skin can be converted directly into accurate temperature values; this makes infrared imaging an ideal procedure to evaluate surface temperatures of the body [1].

Temperature changes within the human body in relation to disease have been recognized for many centuries. In particular, elevated body temperature has been used as an index of illness and often as an indicator of the progression of a disease [2].

Infrared imaging was introduced into medicine in the late 1950s. Early studies suggested there were applications of the technology in areas as diverse as detection of breast cancer and malfunctions of the nervous system. However, the early instrumentation was not sensitive enough to detect the subtle changes in temperature needed to accurately detect and monitor disease [1, 3].

In recent years, the sensitivity of infrared instruments has greatly improved so that it now approaches 0.025°C, a level at which 0.05 to 0.1°C differences or changes in temperature can now be reliably measured. The development of focal plane technology has permitted very high-quality imaging as well as the export of digital temperature measurements to computer-assisted image analysis and algorithm development [3].

One of the areas that has been explored using infrared imaging is cancer detection. This is due to the fact that there are differences in energy consumption of normal and cancerous tissue that lead to small but detectable local temperature changes [1]. For example, a typical infrared image of a breast tumor reveals a 1-2°C elevation in skin surface temperature at the periphery of the tumor, with the tumor mass often being associated with a corresponding reduction in skin surface temperature [4].

Cancer is a major public health problem in the United States and other developed countries. Currently, one in four deaths in the United States is due to cancer. Breast cancer will be the most commonly diagnosed type of cancer among women in 2007;, it is expected to account for 26% (178,480) of all new cancer cases among women [5]. Since a cure rate higher than 95% is possible if breast cancer is treated in the

earliest stages [6], early detection strategies are of vital importance in order to reduce the mortality due to this illness.

Initial studies of the application of IR imaging to breast cancer concentrated on using breast IR imaging (contact or tele-thermometry) as a stand-alone technology for the detection of breast cancer in a screening environment. The early Breast Cancer Detection and Demonstration Projects, which were carried out between 1973 and 1981 by the American Cancer Society and National Cancer Institute of the United States, showed that IR imaging was able to tell the surgeon that the patient was very likely to have breast cancer, but not able to tell accurately the location of the possible lesion. Therefore IR imaging was considered unacceptable as a stand-alone detection device. However, the FDA approved in 1982 breast thermography as an adjunctive diagnostic breast cancer screening procedure [7].

Michel Gautherie from the National Institute for Health and Medical Research in Paris, France, analyzed 147 breast cancer patients and measured in-vivo temperature distributions and thermal conductivities using fine-needle thermoelectric probes; by using these parameters, he was able to determine the metabolic heat production of cancerous tissue and compare it with the metabolic heat production of normal tissue [8].

In the present work, the bioheat transfer equation is solved for a simplified model of a female breast and a cancerous tumor using the metabolic heat rates and thermal conductivities for cancerous and normal tissue reported by Gautherie. These simulations were performed in order to obtain the sensitivity required by an infrared imaging system that will effectively detect a breast tumor of a certain size located at a certain depth.

2. Numerical Modelling of Heat Transfer in Biological Tissue

The transport of thermal energy in living tissue is a complex process involving multiple phenomenological mechanisms including conduction, convection, radiation, metabolism, evaporation, and phase change.

Bioheat transfer processes in living tissues are affected by the blood perfusion through the vascular network on the local temperature distribution. When there is a significant difference between the temperature of blood and the tissue through which it flows, convective heat transport will occur, altering the temperatures of both the blood and the tissue. Perfusion based heat transfer interaction is critical to a number of physiological processes such as thermoregulation and inflammation [1].

Pennes published the seminal work on developing a quantitative basis for describing the thermal interaction between tissue and perfused blood [9]. His work consisted of a series of experiments to measure temperature distribution as a function of radial position in the forearms of nine human subjects. Pennes proposed a model to describe the effects of metabolism and blood perfusion on the energy balance within tissue. These two effects were incorporated into the standard thermal diffusion equation, which is written in its simplified form as:

$$\rho \cdot c \frac{\partial T}{\partial t} = \nabla(k \cdot \nabla T) + \omega_b \cdot c_b \cdot \rho_b (T_a - T) + q_m, \quad (1)$$

where k is the thermal conductivity of tissue, ρ_b and c_b are the density and the specific heat of the blood, ω_b is the blood perfusion rate (ml/s/ml), q_m is the metabolic heat generation rate (W/m^3), T_a is the arterial blood temperature, and T is the local temperature of the breast tissue. The temperature of the arterial blood is approximated as the core temperature of the body.

The breast model used in this study was approximated using a hemisphere with a 9 cm radius to match the average patient in Gautherie's study [8]; also, a layer 1.3 cm thick was placed beneath the hemisphere to simulate the chest wall, and the tumor was considered a perfect sphere (Fig. 1).

A density of 920 Kg/m^3 and a heat capacity of 3000 J/Kg°C was used for both normal and cancerous tissue. The metabolic heat generation and blood perfusion rate for normal tissue were considered to be 450 W/m^3 and 0.00018 ml/s/ml respectively; in the case of cancerous tissue, values of 29,000 W/m^3 and 0.009 ml/s/ml were used to account for the higher blood perfusion rates and metabolic heat generation found in cancerous tissue [1].

An effective thermal conductivity of 0.42 W/m, as estimated by Gautherie [8], was used for both normal and cancerous tissue.

Finite Element Method simulations were performed using COMSOL Multiphysics, a convective boundary condition with a heat transfer coefficient of 5 W/m^2 K was used at the skin surface to account for natural convection [1].

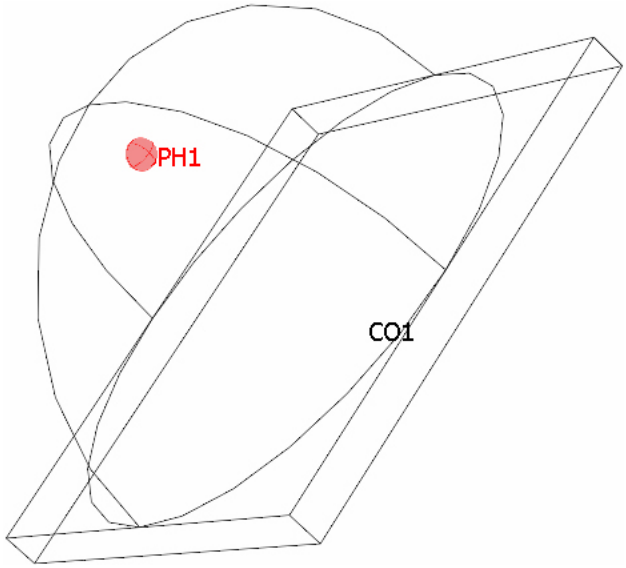

FIGURE 1. Computational model used to replicate the typical patient in Gautherie's study. The model consisted of a 9 cm radius hemisphere attached to a 1.3 cm layer that replicates the chest wall; tumors were considered as perfect spheres.

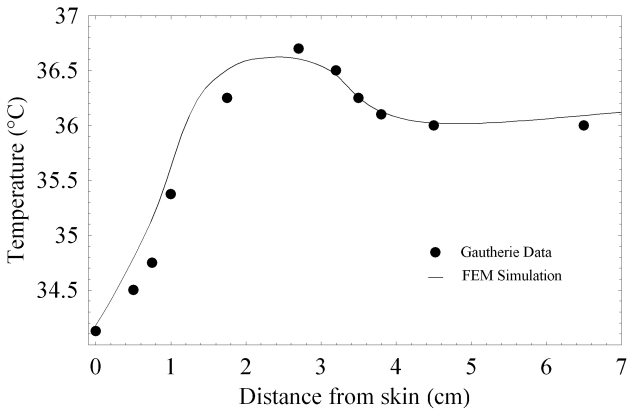

FIGURE 2. Temperature distribution obtained using a finite element simulation of a 2.3 cm tumor with its center located 2 cm deep from the skin surface compared to experimental data.

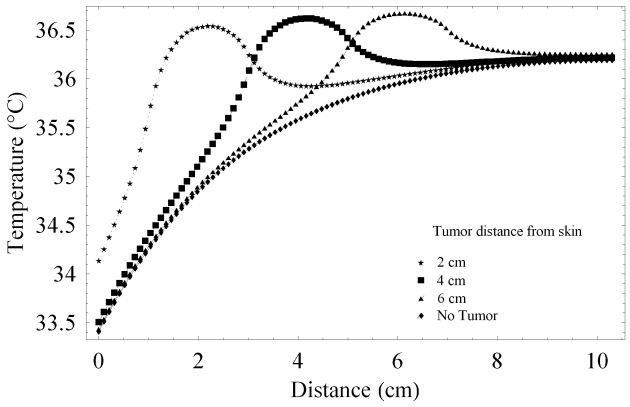

FIGURE 3. Temperature profiles for a 2 cm tumor located at different depths compared to a healthy breast.

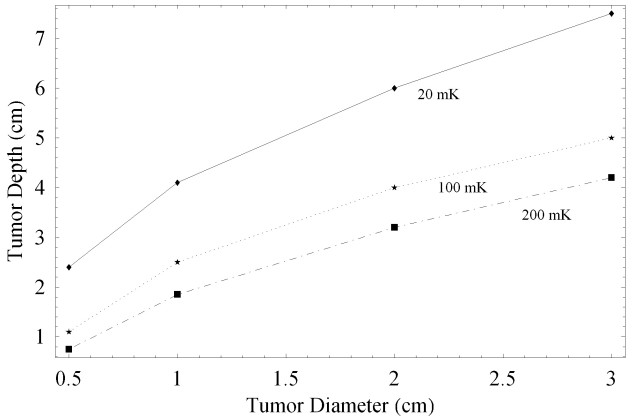

FIGURE 4. Minimum size and maximum depth of tumors that infrared imagers with 20, 100 and 200 mK sensitivities can resolve.

As a first step toward validating the proposed model, FEM simulations were compared to the experimental data reported by Gautherie [8]. Figure 2 shows good agreement between the numerical model and Gautherie's experimental data.

3. Results

The effectiveness of using an infrared imager to detect breast tumors depends on the capability of the imager to resolve the difference in temperature at the skin surface generated by the tumor. Figure 3 shows the temperature profiles generated by a 2 cm tumor located at different depths. Tumors near the skin surface generate differences in temperature larger than 0.7°C at the skin surface; these differences in temperatures can be easily resolved by a state-of-the-art infrared imager.

In order to quantify the minimum size of a tumor or the maximum depth of a certain sized tumor that a state-of-the-art imager can detect, a parametric study was performed by varying the size and depth of tumors and matching the difference in temperatures at the skin surface with the sensitivity of state-of-the-art imagers.

Figure 4 shows the maximum depth and minimum size of a tumor in order to be resolved by infrared imagers with 20, 100 and 200 mK sensitivities.

From Fig. 4 it can be seen that current state-of-the-art imagers are capable of detecting 3 cm tumors located at distances farther than 7 cm away from the skin surface; these imagers can also detect tumors smaller than 0.5 cm located close to the surface of the skin.

4. Conclusions

In this work, a thermal model of a female breast and a cancerous tumor was developed using experimental and previously published data. Finite element simulations performed using this thermal model, showed good agreement with published experimental results. Using this thermal model a parametric study was performed by varying the size and depth of tumors and matching the difference in temperatures at the skin surface with the sensitivity of state-of-the-art imagers. This parametric study showed that current state-of-the-art imagers can detect 3 cm tumors located at distances farther than 7 cm away from the skin surface, and tumors smaller than 0.5 cm located close to the surface of the skin.

Acknowledgments

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