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Optimization of skin cooling by computational modeling for early thermographic detection of breast cancer

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ABSTRACT

The purpose of this study is to enhance the early detection of breast cancer using dynamic infrared (IR) imaging by optimizing thermostimulation with cooling stress to improve thermal contrast. A 2D hemispherical breast model was built to compute steady-state and transient surface temperature profiles for tumors of different size (10–30 mm), depth (6.6–26.6 mm) and location (15°–90°). Larger tumors and tumors closer to the skin surface leave sufficiently large thermal signatures (~0.6 °C) to be detected by steady state IR imaging. Smaller and deeper tumors in the middle and bottom portion of the gland, with thermal contrasts below 0.1 °C, require dynamic imaging with thermostimulation (cooling) to achieve satisfactory thermal contrast for IR detection. In this paper, we consider cooling times of 15–25 min and cooling temperatures of 5–15 °C to optimize thermal contrast. Cooling penetration depths during the cooling phase for constant temperature cooling at 5 °C, 10 °C and 15 °C were analyzed. To achieve the maximum thermal contrast for deeper and smaller tumors, the tissue should be cooled 5–15 min, and in the maximum thermal contrast of the thermal recovery phase appears after 20–45 min. Effects of tumor size and depth on maximum thermal contrast were analyzed systematically to provide recommendations and guidelines for clinical applications. Thermal signatures computed in this study provide valuable data for inverse reconstruction algorithms that allow the measurement of tumor properties, such as the metabolic heat generation rate.

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1. Introduction

Breast cancer is the most common cancer in women worldwide, about 40,610 women in the US are expected to die in 2017 from breast cancer. Death rates have been decreasing since 1989 and this decrease is attributed to early detection efforts [1]. Early diagnosis is very important to reduce mortality in the breast tumor patients. Various imaging methods, such as X-ray mammography (first and most common imaging modality), ultrasound, magnetic resonance imaging (MRI), and infrared (IR) thermography are used in the detection.

Cancerous breast tumors, of special interest for early detection, are known to generate more heat and develop increased blood supply compared to the surrounding healthy tissue, which is mirrored in the local increase of skin surface temperature above the tumor. The elevation of skin surface temperature relative to the temperature of the surrounding skin unaffected by the tumor – termed thermal contrast (TC) in this study – can be visualized and accurately measured with infrared (IR) thermography. Thermography is a non-invasive, non-contact and painless detection method for detection of cancerous and non-cancerous lesions, including breast cancer. Breast thermography had achieved an average sensitivity and specificity of 90% under strictly controlled environmental conditions [2]. The research results had showed that 96% of the 47 studied breast carcinoma cases were detected with infrared thermography utilizing a structured methodology [3]. The advantage of thermography over other imaging techniques is its ability to detect elevated metabolic activity levels and increased blood supply, which can be indicators of malignancy. Mammography only detects the presence of the tumor, but lacks the ability to distinguish between heat generating (potentially malignant) and benign lesions.

A large body of literature focused on thermography for breast tumor detection and diagnosis, including image acquisition protocols, image storage, segmentation methods and feature extraction [4–8]. Researchers developed mathematical models to estimate breast tumor thermal properties to support infrared imaging. Shi et al. [9] presented a new heat transfer model suitable for thermal tomography, and proposed the diagnostic criteria of female breast

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Nomenclature

ρ	density (kg/m^3)
c	specific heat (J/kg K)
t	time
T	temperature (°C)
k	thermal conductivity (W/m K)
ω	blood perfusion rate (1/s)
Q	metabolic heat generation (W/m^3)
h	convection heat transfer coefficient (W/m^2 K)
q''	heat flux (W/m^2)

Subscripts

i	skin layer
b	blood
c	cooling
∞	ambient condition
cp	cooling phase
ss	steady state
max	maximum

diseases using the q-r characteristic curve. Koushik and Subhash [10] estimated the size and location of a tumor in the breast based on the measurement of the skin surface temperature, and the steady-state temperature distribution in the tissue-tumor system was obtained by solving the Pennes bioheat equation using the finite volume method. Saniei et al. [11] estimated the depth, size, and metabolic heat generation rate of breast tumors using the dynamic neural network model (DNN) based on breast thermal images. Bezerra et al. [12] presented a methodology to estimate the thermal conductivity and blood perfusion of breast tissue based on infrared images with the FLUENT software used for the numerical simulation.

All these studies focused on the application infrared thermography technology, and used the IR images as starting point to determine tissue properties by computational modeling. To better understand and interpret thermal signatures of tumors developing on the skin surface, and to efficiently apply infrared thermography to achieve the maximum thermal contrast, theoretical thermal models of breast tumors are studied. Chanmugam et al. [13] quantitatively analyzed thermal signatures of breast cancer lesions using a 3D computational model. The effect of tumor size and depth on thermal contrasts (maximum temperature difference) was analyzed for the steady state only. No in-depth analysis of the cooling load effect on deep tumors, such as the effect of cooling time and temperature on thermal contrasts, has been reported in the literature. Hatwar and Herman [14] introduced a computational model of the axisymmetric tumorous breast with six tissue layers. They simultaneously estimated size location and blood perfusion of the cancerous breast lesion from steady state and transient surface temperature data, iteratively, using an inverse algorithm based on Levenberg–Marquardt method. Amri et al. [15] used a simplified three-dimensional two-layer numerical breast model to analyze steady-state and transient temperature distributions. A cold stress was applied in their model with the aim to detect breast tumor depth. The effects of the cooling temperature and cooling duration on the transients were also examined for different tumor diameters and depths. However, in this study the breast model was assumed to be a cube shape, only consisting of a 5 mm thick fat layer and 45 mm thick gland layer. A more realistic model of the female breast is considered to be a hemisphere with different layers of uniform thickness, with tumors at different locations of gland area.

To enhance the sensitivity and effectiveness of dynamic quantitative IR imaging for diagnosis, especially for the early detection and diagnosing deeper lesions in general and breast cancer in particular, thorough and systematic insight into and subsequent optimization of the cooling process and the thermal recovery phase is necessary. A systematic analysis and optimization of skin cooling for IR diagnostic applications was carried out for near-surface lesions by Cheng and Herman [16]. They defined lesions with less than 4 mm invasion as near-surface lesions and their study aimed at the quantification of the thermal features of skin cancer. The key challenge for both near-surface and deep lesions is to apply sufficient cooling to achieve maximum thermal contrast between healthy tissue and regions affected by the lesion, in order to detect it with an IR camera as well as to accurately quantify it. Dimensions (diameter and depth), thermal signatures and cooling requirements of early skin cancer lesions can be dramatically different from those of deep lesions (such as breast cancer), therefore it is necessary to treat deep lesions as a different class of thermostimulation problems. During thermostimulation, the overcooling of the tissue is a particular challenge for near surface lesions, as discussed in [16]. Their results suggest that moderate cooling temperatures (around 20 °C) and short cooling times (<2 min) are optimal for near-surface lesions, corresponding to conditions which are easy to achieve in a clinical setting. Cooling for dynamic IR imaging is much more challenging for deep lesions, since it will take longer for the thermal disturbance to reach the lesion and subsequently manifest itself as a thermal signature on the skin surface. Therefore, minimizing cooling time and patient discomfort is of paramount importance for deep lesions, including breast cancer.

In this research, we consider a 2D cross section of hemispherical breast model with a cancerous lesion within the gland layer and compute surface temperature profiles for a variety of situations to support early detection of breast cancer by IR imaging. The model is created with the software COMSOL Multiphysics v5.1 and it systematically considers thermal responses for tumors of different size, depth and location. While larger tumors and tumors closer to the surface exhibit a strong enough thermal signature for detection under steady state conditions, the detection of deeper and smaller tumors requires additional measures to broaden the applicability of infrared imaging in clinical applications. These cooling and imaging requirements will be addressed in the paper. For improving the sensitivity and effectiveness of deep tumor thermal imaging, a cooling stress is applied to enhance thermal contrast between breast tumor and healthy tissue. The deeper the tumor, the lower cooling temperature and longer cooling time is needed. The aim of this paper is to optimize the cooling time and cooling temperature with tumor size, depth and location, while minimizing patient discomfort. In addition, cooling techniques feasible for implementation in a clinical setting will be evaluated. Therefore, this study complements and expands our prior work on near surface lesions to address specific challenges imposed by the quantitative detection deep lesions. We also discuss the limitations of IR imaging in terms of tumor size and depth, as well as key parameters determining the magnitude of the thermal contrast.

Testing cooling strategies and limitations of the measurement system in a clinical setting would potentially cause discomfort or even injury to the person, the test subject, and access to human subjects and breast cancer patients is extremely limited and

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