

**EVALUATION ON LASER BEAM ASSISTED INFRARED THERMAL DIAGNOSIS OF  
 SUPERFICIAL CIRCULATION DISEASE**

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**INTRODUCTION**

Temperature is a long established indicator of health [1]. Abnormalities such as malignancies, infection, and vascular disease cause localized temperature non-uniformity which shows as hot spots or as asymmetrical patterns in an infrared thermograph. Although the body temperature may be measured locally by a mercury thermometer or thermocouples placed in physical contact, infrared thermal imaging has been favored in medicine for diagnosis purposes for several decades due to its noninvasive and non-contact features. However, with the increasing application of other non-invasive techniques such as CT and MRI, the infrared method for disease diagnosis seems to have increasingly fallen out of favor [2, 3]. The failure of the technique has been attributed to difficulties in interpreting the thermal images with subtle thermal pattern, rather than the approach being theoretically flawed [4]. In fact, the vascular and metabolic changes that occur at sites of superficial diseases would be expected to translate into changes in skin surface temperature. In order to accurately detect diseases especially at early stages or deeply embedded, many attempts have been performed to enhance the skin thermal expression [5, 6].

In this study, a new laser-beam assisted approach for infrared thermal diagnosis was proposed, and its feasibility was evaluated through numerical simulation. Two typical types of vascular change model were considered during numerical analysis, including effects of microcirculation and large vessel.

**METHODS**

The computation domain was taken as a  $0.1\text{m} \times 0.1\text{m} \times 0.2\text{m}$  cube and depicted in Figure 1, in which  $x$  denotes the tissue depth from the skin surface while  $y$  and  $z$  are along the surface. The well-known Pennes equation is used to model heat transfer in human bodies:

$$\rho c \cdot \partial T(\mathbf{X}, t) / \partial t = \nabla \cdot k \nabla [T(\mathbf{X}, t)] - \omega_b \rho_b c_b T(\mathbf{X}, t) + Q(\mathbf{X}, t) \quad (1)$$

where  $Q(\mathbf{X}, t) = Q_m + Q_r + \rho_b c_b \omega_b T_a$ ;  $\rho$ ,  $c$  are the density and the specific heat of tissue, respectively;  $\rho_b$ ,  $c_b$  denote density and specific heat of blood;  $\mathbf{X}$  contains the Cartesian coordinates  $x$ ,  $y$  and  $z$ ;  $k$  is the thermal conductivity, and  $\omega_b$  the blood perfusion;  $T_a$  the arterial temperature which is treated as a constant, and  $T(\mathbf{X}, t)$  the tissue temperature;  $Q_m$  and  $Q_r$  are the metabolic heat generation and spatial heat generation due to external heating.

For vascular change in microcirculation (i.e., change in the blood perfusion and/or metabolic rate), the tissue temperature can be determined by Equation 1. For vascular change in large vessel, the heat transfer in large vessel is governed by the convective heat transfer equation [7]

$$\rho_b c_b \frac{\partial T}{\partial t} = \frac{hP}{S} (T_w - T) - C_b v \frac{\partial T}{\partial z}, \quad \mathbf{X} \in \Omega_1 \quad (2)$$

The spatial heat generation induced by laser beam is given by Beer's law

$$Q_r = \eta P_0 \exp(-\eta x) \quad (3)$$

where  $\eta$  is scattering coefficient of tissue, and  $P_0$  is spatial heating power flux at skin surface.

The generalized boundary condition for the heat transfer occurring at skin surface is generally composed of three parts (i.e., convection, radiation and evaporation), which can be written as [6]

$$-k \left. \frac{\partial T}{\partial n} \right|_{skin} = h_f (T_s - T_f) + \sigma \varepsilon (T_s^4 - T_f^4) + (3.054 + 16.7 h_f W_{rsw}) (0.256 T_s - 3.37 - P_a) \quad (4)$$

where  $h_f$  is the convection heat transfer coefficient;  $T_s$  and  $T_f$  are the skin and surrounding air temperatures respectively;  $\varepsilon$  is the skin emissivity, and  $\sigma$  the Stefan-Boltzmann constant;  $W_{rs_w}$  is the skin humidity,  $0 \leq W_{rs_w} \leq 1$ , and  $W_{rs_w} = 0,1$  respectively mean that the skin is dry and entirely wet;  $P_a$  the vapor pressure in ambient air.

The numerical algorithm used in this study is revised from that developed in our previous work [7]. Readers are referred to [7] for more details.

## RESULTS AND DISCUSSION

Figures 2-4 shows part of results, in which Figures 2 and 3 are for the case of vascular change in large vessel (the central line of the large vessel is at  $[x=0.006 \text{ m}, y=0.05 \text{ m}]$ , and the diameter of large vessel is 1mm), and Figure 4 for the case of vascular change in microcirculation (i.e., change in the blood perfusion and metabolic rate). The heating time of laser beam for all cases is 5 seconds. It is clearly shown in Figure 2 that with the heating of laser beam, the difference between thermal pattern for normal case and that for disease case is obvious. Further results indicate that the laser-beam assisted approach for infrared thermal diagnosis can significantly enhance the sensitivity of temperature mapping on skin surface over the disease region. The results depicted in Figure 3 and Figure 4 suggest that with multiple laser beams' heating, the temperature over the disease region is much different from that over the normal tissue region. It is also indicated that employing multiple laser beams can conveniently detect the superficial diseases, which may offer a new mode for infrared thermal diagnosis. This potential feature is expected to have important clinical implications for the practical use of infrared imaging system in the non-invasive diagnosis of superficial diseases.

More results including some experimental tests and detailed discussion will be presented at the conference.

## ACKNOWLEDGMENTS

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## REFERENCES

- [1] Jones, B. F., 1998, "A reappraisal of the use of infrared thermal image analysis in medicine," *IEEE Transactions on Medical Imaging*, **17**, pp. 1019-1027.
- [2] Head, J. F., Wang, F., Lipari, C. A. and Elliott, R. L., 2000, "The important role of infrared imaging in breast cancer," *IEEE Engineering in Medicine and Biology*, **19**, pp. 52-57.
- [3] Keyserlingk, J. R., Ahlgren, P. D., Yu, E., Belliveau, N. and Yassa, M., 2000, "Functional infrared imaging of the breast," *IEEE Engineering in Medicine and Biology*, **19**, pp.30-41.
- [4] Xie, W., McCahon, P., Jakobsen, K. and Parish, C., 2004, "Evaluation of the ability of digital infrared imaging to detect vascular changes in experimental animal tumors," *International Journal of Cancer*, **108**, pp. 790-794.
- [5] Ohashi, Y. and Uchida, I., 2000, "Applying dynamic thermography in the diagnosis of breast cancer - techniques for improving sensitivity of breast thermography," *IEEE Engineering in Medicine and Biology*, **19**, pp. 42-51.
- [6] Deng, Z. S. and Liu, J., 2004, "Mathematical modeling of temperature mapping over skin surface and its implementation in thermal disease diagnostics," *Computers in Biology and Medicine*, **34**, pp. 495-521.
- [7] Deng, Z. S. and Liu, J., 2004, "Monte Carlo simulation of the effects of large blood vessels during hyperthermia," *Lecture Notes in Computer Science*, **3314**, pp. 437-442.

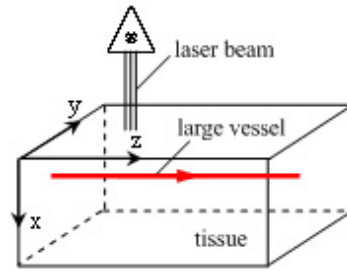
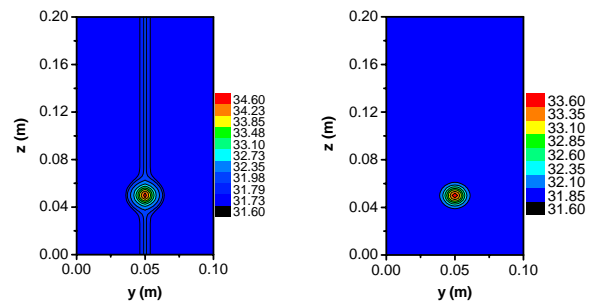


Figure 1. Schematic of computational domain



(a)  $v=0.1\text{m/s}$  (b) thrombus,  $v=0$   
Figure 2. Temperature mapping at skin surface 120 s after heating with single laser beam

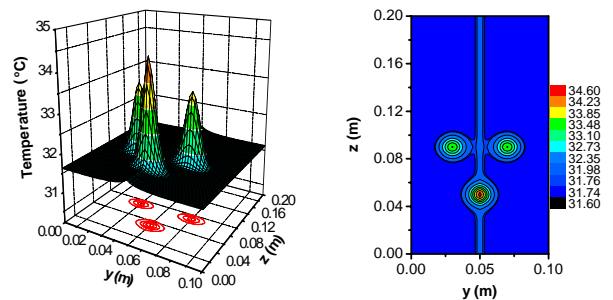
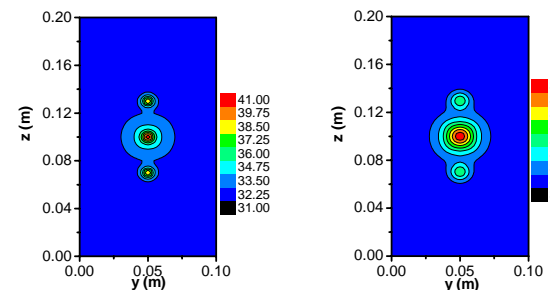


Figure 3. Temperature mapping at skin surface 120 s after heating with 3 laser beams



(a) 60 s after heating (b) 120 s after heating  
Figure 4. Temperature mapping at skin surface after heating with 3 laser beams