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Development of a temperature distribution simulator for lung RFA based on air dependence of thermal and electrical properties

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***Abstract*— Radio frequency ablation (RFA) for lung cancer has increasingly been used over the past few years, because it is a minimally invasive treatment. As a feature of RFA for lung cancer, lung contains air. Air is low thermal and electrical conductivity. Therefore, RFA for this cancer has the advantage that only the cancer is coagulated, because the heated area is confined to the immediate vicinity of the heating point. However, it is difficult for operators to control the precise formation of coagulation zones due to inadequate imaging modalities. We propose a method using finite element method to analyze the temperature distribution of the organ in order to overcome the current deficiencies. Creating an accurate thermal physical model was a challenging problem because of the complexities of the thermal properties of the organ. In this study, we developed a temperature distribution simulator for lung RFA using thermal and electrical properties that were based on the lung's internal air dependence. In addition, we validated the constructed simulator in an in vitro study, and the lung's internal heat transfer during RFA was validated quantitatively.**

I. INTRODUCTION

A. *Radio frequency ablation (RFA) for cancer*

Radio frequency ablation (RFA) is an important modality for treating cancers and has increasingly been used over the past few years [1]-[2]. RFA involves an electrode being percutaneously introduced into the cancer and RF energy being applied, whereupon the temperature of the tissue rises due to the ionic agitation generated by radio waves at 470 kHz. Tissue coagulation occurs as a result of protein denaturation when the tissue around the electrode reaches a temperature of around 60°C. Subsequently, moisture evaporation occurs and the cancer becomes completely necrotic at 70-80°C. This percutaneous procedure, which has proven to be effective and safe, also has the advantage of being minimally invasive, meaning lower-impact operations and shorter hospital stays.

B. *RFA for lung cancer*

In recent years, the effectiveness of RFA has been reported for liver cancer and breast cancer, and as a consequence RFA is now beginning to be applied in lung cancer. The RFA procedure for lung cancer is to percutaneously puncture the cancer with the RF electrode. Subsequently using X-ray CT, the operator checks an ablation region and the relative positions of the cancer and RF electrode. Finally, the radio wave is turned on.

A characteristic of RFA for lungs is that these organs contain abundant air. Air is low in thermal conductivity and electrical conductivity. Therefore, the internal air focuses the RF energy from an electrode needle onto a limited region. Thus, the lung is more suitable for cauterization than the breast, liver or bone.

C. *Practical shortcomings of RFA for lung cancer*

On the other hand, precise cauterization of lung cancers is very difficult because the presence of pneumothorax and air inside the lung greatly affects the heat transfer and temperature distribution near a RF electrode. In fact, excessive ablation or non-ablation of cancers has been reported in lung RFA treatment [3]-[4]. In the case of the lung as compared with other organs such as the liver, the actual region of ablation may not coincide with the original region that was targeted by the operator. Additionally, although the temperature distribution of the ablation region is invisible, it is possible using RFA in combination with X-ray CT to determine protein denaturation during cautery for lung cancers. So, the pattern of heat transfer during lung RFA and increase in tissue temperature near the RF needle is not clear.

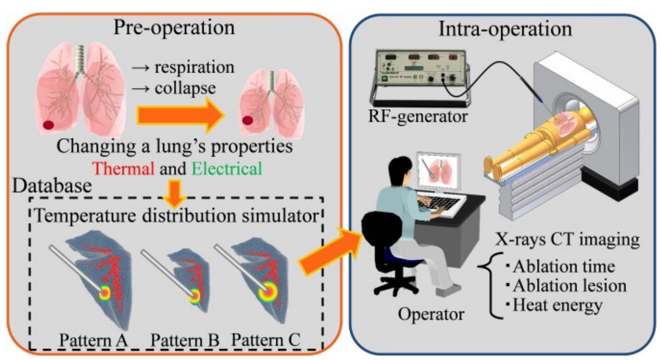

Figure.1. Overview of system concept.

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II. Objectives

The objective of this study was to develop a precise ablation system for the support of RFA treatment of lung cancer, using a computer simulator based on an accurate thermal physical model of lung. An overview of the system concept is shown in Figure 1. The proposed surgical simulation attempts to provide visual information on the condition of the cauterized cancer for the operator. It involves:

1) Optimized RF energy to cauterize the lung cancer
2) Appreciation of the ablation region and ablation time
3) Heat transfer to the normal tissue adjacent to the cancer

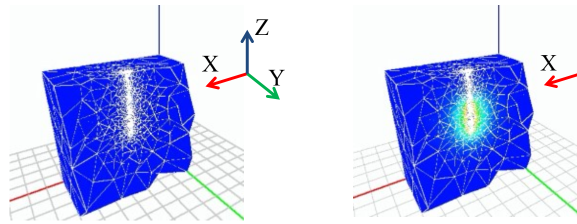
In order to construct an accurate ablation simulator, it is necessary to use a reliable thermophysical model that has an accurate database of the thermophysical properties, and electrical properties of the organ such as thermal conductivity and electrical conductivity. However, creating an accurate thermophysical model was a challenging problem because of the complexity of the thermophysical properties of the organ. One of the complexities is that the lung's thermal properties and electrical properties change with its internal air content due to respiration, or collapse due to RF electrode insertion. The thermal and electrical conductivity are included in the bio heat equation (1) that is required to calculate the tissue temperature around a cancer [5].

$$\rho C \frac{\partial T}{\partial t} = \lambda \nabla^2 T + \sigma |E|^2 - \rho \rho_b c_b F (T - T_b) + Q_m \qquad (1)$$

where ρ is the density of the organ [kg/m^3], c its specific heat [J/kgK], T its temperature in °C, λ its thermal conductivity [W/mK], σ its electrical conductivity [S/m], E its electrical field [V/m], ρ_b the density of blood, c_b the heat capacity of blood, F the blood perfusion coefficient of the organ [m^3/kgs], θ_b the blood temperature and Q_m the metabolic heat source term of the organ [W/m^3]. The left-hand clause 1 is heat capacity. The right-hand clause 1 is heat transfer. Clause 2 is the heat value from the RF electrode and clause 3 is the absorption of heat by organ's blood flow.

From our previous research, the lung's thermal conductivity and electrical conductivity decreased non-linearly with increasing internal air volume in the lung [6-7]. From the equation (1), we suggest that the reason why the extent of ablation of lung lesions using RFA is smaller than in other organs is due to decreases. The novelty of the present study is in measuring and modelling the internal air dependence of the lung's thermal and electrical properties, based on quantitative experiments. We constructed a temperature distribution simulator for lung RFA (Fig. 2). The finite element method (FEM) simulator consisted of 4235 elements and 22152 nodes. Figure 2a shows the lung simulator before RFA. Figure 2b shows the temperature rise near a RF electrode at 600 s after RFA. The constructed simulator was validated using an in vitro experiment, and the lung's internal heat transfer was validated quantitatively from the viewpoint of the RFA heat transfer mechanism.

This report is organized as follows: Section III describes the method of temperature distribution measurement using the simulator and in vitro experimental validation. In Section IV and V the results and discussion are presented. Finally, Section VI details our conclusions and plans for future work.



(a) 0 s (b) 600 s

Figure. 2. Temperature distribution simulator for lung RFA [5-7].

III. Materials and Methods

A. Simulation

Figure 3 shows the lung model used for the simulation. The model was 70 mm in length, 70 mm in width and 50 mm in depth. The RF electrode was inserted from the X-Y plain surface to a depth of 35 mm towards the Z-axis. Figure 4 shows electrical and thermal boundary conditions. In order to determine the heat value from the RF electrode, it was necessary for the electrical field distribution to be close to the needle. The RF electrode that was used in the in vitro study generated radio waves from near the needle tip to the tip. The distance was 20 mm. Therefore, electrical field analysis was input at 60 V in part of the needle tip and 0 V in the return electrode. In other boundaries the insulation condition due to surface contact with air or needle insertion was determined. After the specific heat value was determined, temperature distribution was calculated from thermal analysis using equation (1). It was used to set two types of thermal analysis boundary conditions. One was the heat transfer condition between the needle surface and lung tissue. The other was the heat insulation condition, and the boundary was the same as that of the lung surface that was in contact with air. The reason for the heat insulation condition was that air has a very low thermal conductivity, and a value that is much lower than the organ's thermal conductivity. So, it sets the heat insulation condition on the lung surface. The boundary temperature was 20°C where is experimental room air temperature.

Under the above-mentioned simulation conditions temperature distribution during RFA was calculated by input of the organ's thermal and electrical parameters. There were five input parameters. The thermal and electrical properties are shown in Table I.

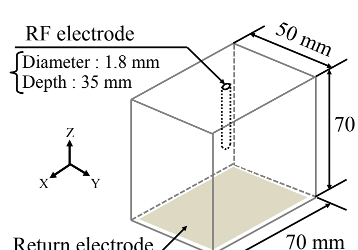
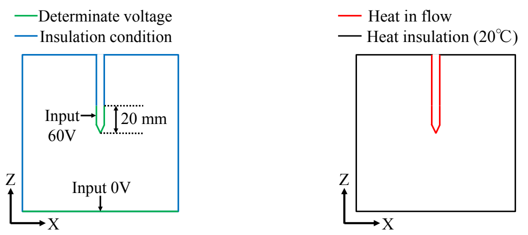


Figure. 3. Simulation shape model.



(a) Electrical field analysis (b) Heat transfer analysis

Figure. 4. Boundary condition.

TABLE I

Input parameters for the temperature distribution simulator

Parameter pattern	*Thermal Conductivity W/mK*	*Electrical Conductivity S/m*	*Density kg/m³*
A	0.200	0.423	480
B	0.125	0.423	480
C	0.200	0.423	240
D	0.125	0.423	240
E	0.200	0.100	480
F	0.125	0.100	480
G	0.200	0.100	240
H	0.125	0.100	240

Firstly, the specific heat value was 2500 [J/kgK] in all of the eight pattern analyses, because the specific heat does not change the lung's internal air as has been previously reported [8]. Secondly, the eight pattern analyses were set at 60 V input. The voltage matched the in vitro experimental condition. Thus, the three lung properties of thermal conductivity, electrical conductivity and density were changed by variations in the lung's internal air volume. Pattern A represents the collapsed lung condition. The collapsed lung did not contain air. Pattern H represents the aerated lung condition. Aerated lung includes abundant air. In lung cancer ablation, the three parameters are mutually related. In order to validate the influence of each single parameter on temperature distribution, we set and operated more that six pattern analyses (Patterns B ~G). The simulation used measured values of thermal and electrical conductivity and density values published in the literature [8].

B. In vitro experiment

We measured the temperature distribution of hog pig lung during RFA in order to validate how the internal air dependence of the thermal conductivity and electrical conductivity affect ablation and the simulator. The temperature distribution measurement method has been reported in one of our previous studies [9]. We measured temperature distribution using the same method as that used in this previous study. Firstly, the RF electrode and three thermo couples were inserted into the lung at 5 mm intervals from the RF electrode. The insertion location of the needle was the distal portion of the lung. The needle insertion location is periphery of a lung and coincides with the RFA target used in clinical practice. Secondly, RFA was started and measured temperature. Finally, carbonization was generated near the needle, and the RF generator automatically turned off radio waves (Roll-off). With using the above method, we measured the temperature until Roll-off.

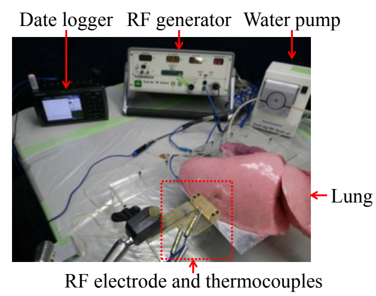


Figure. 5. Experimental overview of in vitro measurement temperature [9].

IV. Results

The results of temperature distribution analysis up to 200 s are shown in Figure 6. The measured temperature data from the collapsed lung and aerated lung up until 200 s are shown in Figure 7. When we cauterized collapsed hog lungs, the RF generator became Roll-off within 120 s. This time was shorter than cases of aerated lungs. So, we analyzed temperature distribution up to until 200 s. Figure 7a shows data from the collapsed lung, which did not include air, and Figure 7b shows the aerated lung's data, which included bundant air.

From the results of FEM analysis and the experimental results, the temperature distributions of the Pattern A and the Pattern H were similar to the measured temperature distributions in the collapsed and aerated lung until Roll-off.

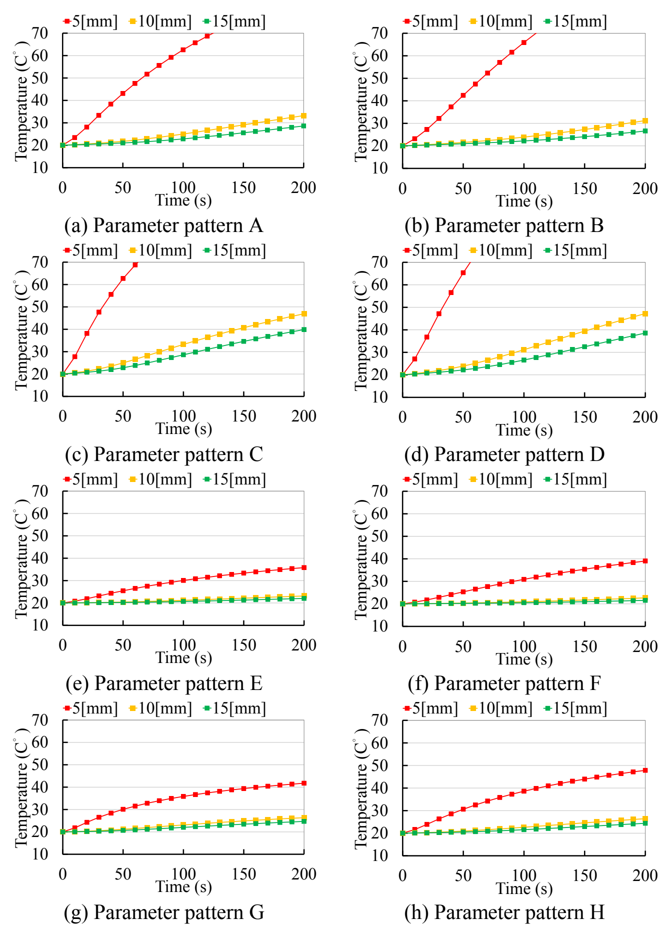
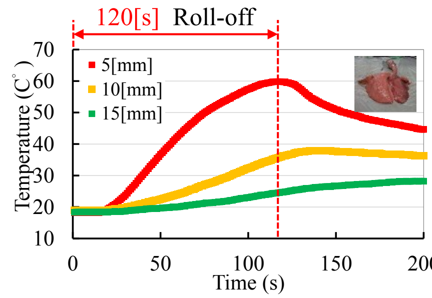
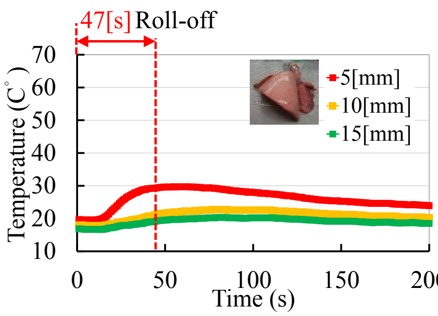


Figure. 6. Results from lung temperature distribution simulation.



(a) Collapsed lung



(b) Aerated lung

Figure. 7. Experimental results from temperature distribution measurement.

Figure 7a shows that the collapsed lung was maintained for 120 s before Roll-off. On the other hand, the ablation times in the aerated lung were shorter than in the collapsed lung and it was maintained for 47 s before Roll-off.

V. Discussion

In the present study, findings from FEM analysis and an in vitro experiment indicated that decreasing the electrical conductivity by means of airflow into lung had a greater effect on ablation than decreasing the thermal conductivity or density. With regard to the mechanism of electromagnetic ablation, as the first step, the heat output from the RF electrode was decreased by the internal air in the lung. This drop in the heat output from the electrode represented a decrease of 76% in comparison with the electrode in the collapsed lung. Therefore, the temperature in the aerated lung was lower than that in collapsed lung at a depth of 5 mm, determined both by analysis and experiment. The next step was the heat transferred from the nearest RF electrode to the lung surface. At this time, due to decreasing thermal conductivity, the heat energy was focused on the tissue nearest to the tip of the RF needle and was hardly scattered to the lung surface. It was supposed that decreasing the thermal conductivity affects the temperature difference at depths between 5 and 10 mm, or 10 and 15 mm. However, in order to inhibit the scattering of heat before ablation is finished, a decrease in thermal conductivity and density were not confirmed as properties required to achieve this. In the final step carbonization was generated in the tissue closest to the needle tip. Thus, we suggest that the reason that RFA in the aerated lung has a reduced heat output focused on tissue nearest the needle tip, and ablation was completed before heat could scatter at distance from the needle was due to tissue carbonization at the needle tip.

VI. Conclusions and Future Work

In this paper, we presented findings from a temperature distribution simulator that we constructed for lung RFA, and from an in vitro experimental validation study. Data obtained using the FEM simulator was a very good match with the actual heat transfer data obtained using thermal and electrical properties based on the lung's internal air dependence. In addition, due to the development of the FEM simulator, the heat transfer properties of the lungs during RFA were evaluated more clearly from the viewpoint of the mechanisms involved in RF ablation over the entire period that it was carried out.

In the future, more detailed validations of the dependence of thermal and electrophysical properties on the levels of air in the lung will be necessary. It is also necessary to measure SAR (Specific Absorption Rate) in order to make clear contribution of lung's thermal and electrical properties ratio for ablation zone and size. We intend to further develop the ablation simulator. The planning and navigation system will also be further developed to enable its use for the delivery of safe and precise clinical RFA treatment for lung cancer.

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