Effect of changes in lung physical properties on microwave ablation zone during respiration

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Abstract

Microwave ablation is a promising minimally invasive treatment for cancer. However, due to the respiratory movement of the lungs, it is very difficult to accurately predict and control the microwave ablation zone. Therefore, the influence of the changes of the physical parameters of the respiratory process on the microwave ablation zone is studied. Firstly, based on the 4D-CT describing the respiratory process of the lungs, all the image data are from 100 non-small cell lung cancer radiotherapy patients (50 males and 50 females, average 58 years, range 55–61 years). According to the theory of porous media, the change of the effective thermal conductivity of the lung tissue during the breathing process is obtained. The effective thermal conductivity of the lung parenchyma during respiration varies from 0.16 to 0.20 W/m °C, with the lowest value at the end of inspiration and the highest at the end of expiration. The transient problems during microwave ablation of pulmonary tissue are analyzed by finite element method. The changes of relative permittivity, conductivity and density changes during the breathing process are also considered. The results show that the microwave ablation zone is significantly larger under dynamic physical parameters. At the end of expiration, when the tissue parameter is set to constant, the ablation lesion area is more concentrated around the tip and slot of the antenna, and the backward heating effect is smaller. Ablation volume was superior in nonventilated lungs. Therefore, single-lung ventilation can be considered during pulmonary ablation to reduce the impact of breathing on the ablation area. These findings can be useful to further our understanding the MWA and hold promise towards achieving successful treatment objective as well as enhanced therapeutic output via improved treatment planning and strategy. This study provides the basis for clinical pulmonary ablation and can also be used as a preoperative plan to provide guidance to physicians.

Keywords Microwave ablation · 4D-CT · Effective thermal conductivity · Porous media · Lung cancer

1 Introduction

In the past few years, more and more hyperthermia methods have been used in clinical practice and become one of the important methods for treating cancer [1]. Currently, clinically applied local hyperthermia methods include Microwave Ablation (MWA), Radiofrequency Ablation (RFA), High-intensity Focused Ultrasound (HIFU), and Laser-induced Interstitial Thermotherapy (LITT) [2]. Among them, MWA is a common clinical hyperthermia method. Due to its effectiveness in the treatment of parenchymal

organ tumors such as liver cancer and kidney cancer. MWA is gradually applied to patients with intolerance surgery to treat primary or metastatic lung tumors [3]. During pulmonary microwave ablation, the microwave radiation probe is percutaneously punctured to the tumor lesion site under CT or ultrasound guidance, and the microwave energy is radiated to the lesion. Due to the pyrolysis effect of microwave on the tissue, the tissue surrounding the ablation probe obtain high temperature. High temperature can cause irreversible damage to the tissue, thereby achieving the purpose of killing malignant tumors.

Unlike other parenchymal organs, the lung is filled with a large amount of air, which greatly affects the temperature distribution in the microwave ablation zone, leading to accurate prediction and control of ablation areas is very challenging. The amount of air in the lungs varies, so the physical parameters of the lungs also change with the breathing

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process. The physical properties of organs are important factors in determining the ablation area. It has been reported that excessive ablation and insufficient ablation during treatment lead to treatment failure [4, 5]. Santos et al. [6] compared the diameter of the ablation area between the ventilated and unventilated lungs, and the results showed that the unventilated lung ablation area was significantly larger than the ablated area of the ventilated lung.

The purpose of this paper is to study the changes in physical parameters during the respiratory process of the lungs and the effects of changes in physical parameters on the microwave ablation zone. The necrotic area formed by microwave ablation is mainly determined by two factors, one is the active heating area formed by the deposition of microwave energy in the tissue, and the other part is due to the existence of temperature difference, the passive heating zone formed by heat conduction. Wherein the formation of the active heating zone is related to the dielectric properties of the tissue; Passive heating zone is related to the thermal conductivity of the tissue. The effective thermal conductivity of the lung tissue is an important physical parameter reflecting the thermal conductivity of the tissue. Firstly, based on the 4D-CT image of the human lung, the three-dimensional model of the lung during the respiration process is reconstructed, and the volume change of the lung during the breathing process is obtained. Then according to the theory of porous medium, the change of effective thermal conductivity of lung parenchyma during respiration is obtained. At the same time, the lungs are accompanied by other physical changes during the breathing process, including relative dielectric constant, conductivity and density, and finally the lung microwave ablation simulation model is established. The effects of changes in effective thermal conductivity, relative dielectric constant, conductivity and density on the microwave ablation temperature field were investigated.

2 Methodology

2.1 Effective thermal conductivity of lung parenchyma

First, study the changes of effective thermal conductivity in the respiratory process of lung tissue. The thermal conductivity of biological tissues is important for further study of the heat transfer characteristics and heat transfer mechanisms of tissue ablation. The heat transfer process of biological tissues is usually described by bioheat transfer models: for example, Pennes bioheat transfer equation, Weinbaum–Jiji model, the Wissler model, the Baish model etc., must take into account the thermal conductivity of the tissue [7].

Due to the complexity of the lung tissue structure, there are many physiological activities that make it difficult to describe the thermal properties of the lung tissue. Khaled and Vafai [8] and Khanafer and Vafai [9] propose that the porous medium theory is most suitable for the heat transfer in biological tissues. The microscopic anatomical structure of the lung parenchyma is quite complicated. Guo [10] regards the lung parenchyma as a porous medium to study the heat transfer process. The effective thermal conductivity of a porous medium depends on various factors such as the characteristics of the solid skeletal structure, the size and spatial distribution of the voids, and the type of fluid in the void space [11]. Due to technical limitations, the thermal conductivity of living lung parenchyma cannot be directly measured by experimental methods. Therefore, it is considered to study the effective thermal conductivity of the lung using a method based on human medical image data.

Traditional CT does not reflect changes in organ morphology during respiration. In recent years, with the advancement of imaging technology, Four Dimensional Computed Tomography (4D-CT) has appeared in order to further reflect the characteristics of changes in the internal organs of the human body with respiratory movements [12]. Based on 4D-CT image analyze the change of the position and volume of lung during respiration, the 3D lung model of dynamic motion was used to describe the change of pulmonary porosity, and then the effective thermal conductivity of the lung during ventilation was obtained.

2.1.1 Three-dimensional reconstruction of lung model

The research materials used in this paper are from The Cancer Imaging Archive (TCIA), an open access database for medical images of cancer research. The data contains clinical data and Computed Tomography (CT) from 100 non-small cell lung cancer (NSCLC) radiotherapy patients (50 males and 50 females, average 58 years, range 55–61 years) who received SBRT were included in a retrospective study. According to the TNM staging system of Union for International Cancer Control, 47 patients were diagnosed at stage I and 53 patients were at stage II, all patients underwent respiratory training before 4D-CT scanning, and the respiratory was controlled. The tomographic image of the lungs consists of 10 phases of CT images in a complete respiration process, all samples containing a total of 1000 image sequences. The 4D-CT images of each sample contain 10 groups, respectively respiratory phase 0, 10%, 20%, 30%, 40%, 50%, 60%, 70%, 80% and 90%. Among them, according to the standard of 4D-CT, 0 is the inhalation end, 50% or 60% is the end of exhalation, and each phase sequence image is reconstructed three-dimensionally. The image acquisition method is spiral scan, the resolution of the image is 512×512 . The entire image sequence of each individual is sorted according to

different respiratory phases. Based on the MIMICS (Materialise) platform for 3D reconstruction of the lung 4D-CT image, and measuring the volume of the reconstructed lung.

Since the lung is filled with a large amount of air, it has a lower density than the surrounding tissue and organs, and there is a significant difference in CT value from the surrounding tissues and organs. Therefore, the lung image is firstly used in the threshold segmentation method. The CT value of the DICOM image ranges from -1024 to 1024 Hu, and the CT value of the air is -1024 Hu. Therefore, the CT threshold of the lung is set to $-1024 \sim -500$ Hu [12]. The segmentation result is shown in Fig. 1.

The method based on threshold segmentation algorithm alone cannot separate the region of interest from the surrounding organs. Next, use the region growing algorithm to segment the image. Select the appropriate seed point to obtain the connected pixels around the seed point, and then obtain a complete segmentation image of the lung parenchyma. The segmented images were observed layer by layer, the lung parenchyma was completely separated from the surrounding organs and the other tissues with similar density were not adhered. The effects of other tissues and organs on the lung volume were removed, and the lungs were completely segmented. The result is shown in Fig. 2.

After segmenting the lung tissue, a three-dimensional model of ten phases in the respiration process was reconstructed based on the MIMICS platform. Using the pixel accumulation method to calculate the lung volume of different respiratory phases, a total of 1000 three-dimensional lung models were obtained. The results of the 3D reconstruction model are shown in Fig. 3. The changes in volume and porosity of different phases during respiration are shown in Fig. 4. The first model is the end of inspiration on a breathing cycle, in this case, it is the largest volume, and the chest pressure is greater than the external atmospheric pressure, then exhaled outward, and the lung volume gradually decreases. The fifth or the sixth phase is the end of exhalation, the inhalation process starts from the fifth phase, and the tenth phase is inhaled, the lungs are restored to their original volume and pressure. In this process, different individuals have different lung volume changes during breathing. The volume of some individuals' lungs were reduced by more than 500 mL, while others are not. In order to eliminate individual differences, the average volume and porosity of the lungs in the same respiratory phase of male and female individuals was taken separately. As can be seen from Fig. 4, the volume of the male lung is larger than that of the female, and the change

Fig. 1 Image of the lung segmentation based on threshold segmentation

Fig. 2 Image of the lung segmentation based on the regional growth algorithm

Fig. 3 Three-dimensional reconstruction model of lungs in different phases during respiration

in the breathing process is also more significant than that of the female, this is mainly due to the difference in breathing patterns.

2.1.2 Porosity of lung tissue

Porosity refers to the ratio of the total volume of micro voids in a porous medium to the total volume of the porous medium. The calculation formula of the porosity ε in each phase of the lung is described as follows [13]:

$$\varepsilon = \frac{V_t - V_0}{V_t} \times 100\% \quad (1)$$

Among them, V_t is the volume after reconstruction of the image sequence of different phases of the lung, and V_0 is the volume when the lung does not contain filling air.

The volume varies with the breathing process. Usually, which consists of four parts: tidal volume, inspiratory volume, supplemental expiratory volume, and residual air volume. They do not overlap each other and all add up to the total amount of gas in the lungs. For a normal adult

Fig. 4 Changes in volume and porosity of different phases during male and female breathing

individual in a calm breathing state, the volume V_0 when the lungs do not contain air is calculated as follows [14]:

$$V_0 = V_t - V_{\text{tidal}} - V_{\text{supplemental}} - V_{\text{residual}} \quad (2)$$

V_{tidal} is obtained according to the volume change of the three-dimensional reconstruction model of the lungs in different phases. The normal adult's supplemental expiratory volume is 900–1200 mL and the residual gas volume is 1000–1500 mL, so this article will supplement the expiratory volume and The remaining gas volume was normal adult mean values of 1050 mL and 1250 mL [14]. The porosity of the different phases of the lungs was calculated according to the formula (1) and (2).

Studies have shown that the lung tissue is porous medium and the effective thermal conductivity of the porous material depends on the complex geometry of the medium. If the heat conduction in solids and fluids occurs in parallel, then the effective thermal conductivity is the weighted arithmetic mean of the solid thermal conductivity k_s and the thermal conductivity k_f [15].

$$K_e = 1 - \epsilon_s + \epsilon_f \quad (3)$$

For the solid thermal conductivity k_s of the lung parenchyma, the literature [11] pointed out that the thermal conductivity in biological tissues is related to content of the protein, fat and water in biological tissues, wherein the protein and fat have the same thermal conductivity, which is about 0.175 W/m °C. Therefore, the thermal conductivity of the solid skeleton of the lung can be calculated from its material composition. [16].

$$k_s = 0.418 \times M_w + 0.175 \quad (4)$$

Among them, M_w is the mass content of water in the biological tissue, which can be expressed by the water content of the lung, and the mass fraction of the lung parenchyma refers to (lung wet weight – lung dry weight)/lung wet weight.

Fig. 5 Changes in effective thermal conductivity of lung parenchyma during respiration

The water mass fraction of the lungs is relatively high at 0.75–0.80 [14], and the average value was used in this study.

For the effective thermal conductivity of lung tissue, where k_f is the thermal conductivity of the air in the lungs, the thermal conductivity of dry air at 0 °C under normal pressure is 0.024 W/m °C, due to air from the mouth, then through the trachea, and at all levels. Under bronchial heating, the air filled in the lung parenchyma contains a certain degree of temperature. According to the literature, the air thermal conductivity of the lung is set to 0.026 W/m °C [17].

2.1.3 Effective thermal conductivity of the lung tissue

According to the formula (3), the effective thermal conductivity of the parenchymal tissue is calculated. When the thermal conductivity of the solid tissue of the lung parenchyma (the extravascular water content of the lung is constant and the thermal conductivity of the air in the lung are fixed, $k_s = 0.509$ W/m °C, $k_f = 0.026$ W/m °C), the effective thermal conductivity of the porous medium increases with the decrease of porosity. The porosity of the lungs is the highest at the end of inhalation and the thermal conductivity is 0.16 W/m °C, because air is a poor thermal conductor, its thermal conductivity is much lower than that of solid biological tissue. At the end of expiration, the pulmonary porosity is the lowest and the effective thermal conductivity of the lung parenchyma is 0.20 W/m °C. In order to eliminate individual differences, the average effective thermal conductivity of the lungs in the same respiratory phase of male and female individuals was taken separately. The 4D-CT images of each sample contained 10 phases, and the effective thermal conductivity of the lungs of males and females. The average value of the changes in different phases is shown in Fig. 5. In general, the effective thermal conductivity of male lung tissue is significantly greater than that of female, and the change in effective thermal conductivity

in a respiratory phase is greater than that in females. In the male and female data, the standard deviation (SD) of the effective thermal conductivity of different individuals was less than 0.005 in the same respiratory phase.

2.2 Pulmonary tissue microwave hyperthermia simulation

In order to study the effect of the change of effective thermal conductivity of lung parenchyma on the temperature field distribution during thermal ablation, finite element method is used to solve the lung ablation simulation model coupled with electromagnetic field and bioheat transfer model. The model also takes into account the changes of other several key physical parameters in the breathing process, include relative dielectric constant, electrical conductivity, effective thermal conductivity and density.

2.2.1 The geometry of the tissue ablation model

An ideal two-dimensional axisymmetric simplified model of pulmonary ablation of is shown in Fig. 6, which idealize the lung tissue to a regular shape for less computation. The ablation probe (antenna) is made of an elongated coaxial cable. To build the antenna, a small ring slot of a specified width is cut on the outer conductor, close to the short-circuited distal tip of the antenna to allow electromagnetic wave propagation into the tissue. This design used a 1 mm wide slot, which can be easily fabricated. Furthermore, the antenna is encased in a PTFE catheter to prevent adhesion of the antenna to desiccated ablated

tissue. The structure and dimensions of the ablation needle are shown in Fig. 5, which operates at 2.45 GHz.

2.2.2 Tissue ablation governing equations

The microwave energy is radiated to the surrounding tissue based on the dielectric material of the antenna, and the temperature of the tissue is increased due to the deposition of the electromagnetic field energy of the microwave in the tissue. The ablation model must take into account biothermal effects, tissue specific absorption rates, and other electromagnetic field effects. According to the Maxwell equation, the magnetic and electric field distributions of microwaves in homogeneous, isotropic media are described below [18].

$$\nabla \times [\epsilon_r^{-1} \nabla \times \mathbf{H}] - k_0^2 u_r \mathbf{H} = 0 \quad (5)$$

The electric field distribution is described as follows:

$$\nabla \times [u_r^{-1} \nabla \times \mathbf{E}] - k_0^2 \epsilon_r \mathbf{E} = 0 \quad (6)$$

where u_r is the relative magnetic permeability of the medium. Generally speaking, the relative magnetic permeability of the biological tissue is 1, k_0 is the wave rate of the electromagnetic wave in vacuum. Described by the following formula [19]:

$$k_0 = \frac{\omega}{c_0} = \omega \sqrt{\epsilon_0 u_0} \quad (7)$$

$$\epsilon_0 = 8.85 \times 10^{-12} \text{ F/m} \quad (8)$$

$$u_0 = 4\pi \times 10^{-7} \text{ H/m} \quad (9)$$

Fig. 6 **a** Antenna structure and size and **b** two-dimensional symmetric simplified model of pulmonary microwave ablation

where ω is the angular frequency of the microwave and c_0 is the propagation speed of the light in the vacuum. In addition, the dielectric constant and electrical conductivity of biological tissues are frequency dependent, and vary with the frequency of the applied electromagnetic field. At the same time, for the lung tissue, the respiratory structure causes changes in the structure and material composition of the lungs. The electrical constant and conductivity are also affected by the breathing process. According to the 4th-order Cole–Cole model, the dielectric constant and electrical conductivity of the lung tissue at the end of expiration and inhalation at 2.45 GHz were calculated. The Cole–Cole model is described as follows [20]:

$$\varepsilon_r(\omega) = \varepsilon'_r - j\varepsilon''_r = \varepsilon_\infty + \sum_{n=1}^4 \frac{\Delta\varepsilon_n}{1 + (j\omega\tau_n)^{1-\alpha_n}} - j \frac{\sigma_s}{\omega\varepsilon_0} \quad (10)$$

where $\Delta\varepsilon_n$ is the dielectric increment at the n th relaxation time in the living tissue, ε_0 and ε_∞ are the dielectric constants when the static dielectric constant and the frequency are infinite, and τ_n is the relaxation time. The effective conductivity σ_{eff} of the tissue can be derived from the relationship between the imaginary part of the complex dielectric parameter and the following formula [21].

$$\sigma_{eff} = \omega\varepsilon_0\varepsilon''_r \quad (11)$$

In order to reduce the amount of calculation, the lung ablation model established in this study is a two-dimensional axisymmetric model. In this case, the wave equation can be simplified into the following form representation.

$$\nabla \times [\varepsilon_r^{-1} \nabla \times \mathbf{H}_\varphi] - k_0^2 u_r \mathbf{H}_\varphi = 0 \quad (12)$$

The magnetic field strength is determined by the total input power of the microwave antenna. The following boundary conditions are applied to the energy input port of the microwave antenna:

$$H_{\varphi 0} = \frac{1}{Zr} \sqrt{\frac{ZP_{in}}{\pi \ln(r_2/r_1)}} \quad (13)$$

where r_2 and r_1 are the inner and outer radii of the coaxial cable, respectively, and Z is the wave impedance. The input power P_{in} has an important influence on the ablation temperature of the targeted tissue. The electric field strength E can be calculated according to Faraday's law.

Due to the deposition of electromagnetic field energy of the microwave in the tissue, the tissue temperature is increased. The heat transfer process of biological tissue under microwave irradiation can be described using the Pennes bioheat transfer equation, which takes into account the cooling effect of blood perfusion on tissue temperature [22].

$$\rho c \frac{\partial T}{\partial t} = \nabla k \nabla T - \rho_b c_b w_b T - T_b + \rho \cdot SAR \quad (14)$$

$$SAR = \frac{\sigma E^2}{\rho} \quad (15)$$

ρ is the tissue density (kg/m^3), c is the specific heat of the tissue, k is the thermal conductivity of the tissue, c_b is the specific heat of the blood (J/m K), w_b is the blood perfusion rate (S^{-1}), and SAR is the specific absorption rate of the tissue to the microwave energy. σ is the electrical conductivity (S/m) of the lung tissue, and E is the electric field strength (V/m).

The degree of ablation damage is closely related to the temperature value it receives. The determination of tissue damage range needs to consider the effect of temperature and time accumulation. According to the thermal dose concept introduced by Arrhenius model, the reference temperature is 43°C , and the integral form is expressed as follows [23]:

$$\Omega(\tau) = \int_0^\tau A e^{\left[\frac{-E_a}{RT(t)}\right]} dt \quad (16)$$

$$\theta_d = 1 - \exp(-\Omega) \quad (17)$$

where Ω represents the degree of damage to the tissue, θ_d is the fraction of necrotic tissue. T represents the reference temperature, t represents the time exceeding the reference time, and the parameter R is the gas constant value of 4, A and E_a represent the frequency factor and reaction energy barrier, respectively. Other physical parameters for the model are shown in Table 1 [23, 24].

Influences of changes in the physical parameters of lung tissue during the breathing process on the temperature distribution of the microwave ablation zone, six sets of parameter

Table 1 Physical properties used in the finite element model (FEM)

Physical properties	Effective thermal conductivity K_f ($\text{W/m } ^\circ\text{C}$)
Relative permittivity of dielectric	2.03
Relative permeability of dielectric	1
Relative permittivity of catheter	2.1
Relative permeability of dielectric	1
Relative permittivity of air slot	1
Relative permeability of dielectric	1
Blood density, ρ_b	1060 kg/m^3
Specific heat capacity of blood, c_b	3600 $\text{J/kg } ^\circ\text{C}$
Perfusion rate, w_b	0.00361 $1/\text{S}$
Blood temperature, T_b	37 $^\circ\text{C}$

Table 2 Input parameters for the temperature distribution simulator

Parameter pattern	Effective thermal conductivity K_f (W/m °C)	Relative permittivity ϵ_r	Conductivity σ (S/m)	Lung density ρ_L (kg/m ³)
A	0.16	20.47	0.80	240
B	0.18	34.42	1.24	360
C	0.20	48.38	1.68	480
D	$ 0.04 \sin(0.5\pi t) + 0.16$	$ 27.9 \sin(0.5\pi t) + 20.47$	$ 0.88 \sin(0.5\pi t) + 0.8$	$ 240 \sin(0.5\pi t) + 240$
E	$ 0.04 \sin(0.6\pi t) + 0.16$	$ 27.9 \sin(0.6\pi t) + 20.4$	$ 0.88 \sin(0.6\pi t) + 0.8$	$ 240 \sin(0.6\pi t) + 240$
F	$ 0.04 \sin(0.7\pi t) + 0.16$	$ 27.9 \sin(0.7\pi t) + 20.47$	$ 0.88 \sin(0.7\pi t) + 0.8$	$ 240 \sin(0.7\pi t) + 240$

patterns were set up in this study, as shown in Table 2, in which the effective thermal conductivity is derived from former part of this article. The relative dielectric constant and conductivity of the lung tissue are calculated according to the 4th-order Cole–Cole model. The density is based on Ref. [25]. In all parameter patterns, the specific heat capacity of the lung tissue was set to 2500 J/kg °C, which is due to studies showing that lung tissue specific heat capacity value are independent of changes in gas content during respiration [16]. The effective thermal conductivity, relative permittivity, electrical conductivity and density all change with the respiratory process of the lungs, and the sinusoidal function is used to represent the physical parameters of the periodic changes of the lung tissue. The temperature distribution of the microwave ablation zone in the lungs under blocked ventilation and normal ventilation was studied separately. First, simulate the microwave ablation process of the lung under static physical parameters. Under the static physical parameter pattern, the physical parameter model of the lung tissue is divided into three states, which respectively reflect the end of expiration, the end of inspiration, and the intermediate state. First block the lungs at the end of inspiration, the lung has the highest air content. The physical parameters at this time are set to pattern A; When the lungs are at the end of expiration, the ventilation is blocked, at this time, the air content in the lungs is the least during the whole breathing process, and the lungs are in a dry state, and the physical parameters is set to parameter pattern C; The parameter pattern B indicates the physical properties of the tissue is between the end of expiration and inhalation, the lungs are semi-filled. During the microwave ablation procedure in the lungs, the lungs of the patient are often normally ventilated. Therefore, the physical parameters of the lungs change periodically. A sinusoidal function is used to describe the periodic variation of the respiratory process. The maximum value of the function is the value of the tissue at the end of expiration, the minimum value is the value at the end of inspiration, and the angular frequency is related to the respiratory rate. In the resting state, the normal adult's respiratory rate is 15–20 times/min. Calculating the angular frequency of the sine function based on the respiratory

rate, The angular frequency of the sinusoidal function is set to 0.5, 0.6 and 0.7 respectively. The effects of respiratory frequency on the temperature field of the ablation zone were studied by comparing D–F with different respiratory frequencies.

2.2.3 Calculation procedure

FEM (Finite Element Method) is an effective tool that can be used in any physical fields, such as the bio-heat transfer problem in MWA [26]. The transient problems during microwave ablation of lung parenchyma are analyzed by FEM using COMSOL™ MULTIPHYSICS (ver5.4a). The procedure in the electromagnetic field and heat transfer problem solving is shown in Fig. 7. The time steps are 2×10^{-12} s for solving electromagnetic wave equation and 0.01 s for solving the bio-heat equation, respectively.

3 Results and discussion

3.1 The treatment region of coaxial antenna at different parameter patterns

The steady-state temperature distribution of the microwave ablation antenna in the lung tissue at 600 s and 10 W is shown in Figs. 8 and 9. Wherein Fig (a) and (b) are the isotherm and damage region clouds of the temperature field radiated around the antenna under the same parameter pattern, respectively. The microwave energy is the strongest near the tip and slot of the antenna, obtaining the highest temperature of the tissue around the slot, and the temperature decrease with the increase of the distance. The shape of the damage field is ellipsoidal, with the long axis (LA) along the direction of the antenna and the short axis (SA) perpendicular to the antenna direction.

The damage area can roughly assume two states. One is that damage fraction of the tissue near the antenna can reach more than 70%, and the tissue presents an irreversible damage state. As the distance from the antenna increases, the complete necrotic area transitions into a reversible damage

Fig. 7 The procedure of calculation in the computational model

area [27]. The irreversible zone accounts for about 30% of the overall ablation area. According to the Arrhenius equation, when the damage fraction of the tissue is less than 0.5, the cell structure of the tissue is in a reversible damage state, and this area has a significant impact on the actual damage area of the final microwave ablation [28]. The reversible damage area gradually transitions to normal tissue, and tissue temperature at the edge gradually approaches the initial value of the model. The cooling effect of blood perfusion and the thermal conductivity of the tissue as well as the penetration distance of the microwave in the tissue limit the extent of the heating zone [29]. When the tissue properties is set according to Table 2. Under the static physical parameters, the tissue property parameters do not change with time, the ablation zone under parameter pattern A B C is compared. Dimensions of the thermally ablated area in terms of long-axis length (LA) and short-axis length (SA) are shown in Fig. 10. In particular, it can be noted that, the long-axis length of the ablated area obtained under the inhalation state (parameter pattern A) increased significantly, but short-axis

length is smaller than under the B C parameter pattern. Other studies can be compared by considering a delivered power of 10 W at 2.45 GHz for 10 min in ex vivo bovine and porcine. In [30], a coaxial antenna was used: a thermally ablated area with LA of 21.8 mm and SA of 18.4 mm, it can be seen that when the lungs are filled with air, the heating area is limited, and the rearward heating effect of the antenna along the long axis is more obvious. Under parameter pattern C, the lung is in the end-expiratory state, at which the effective thermal conductivity, relative dielectric constant, electrical conductivity and density of the parenchymal tissue is the largest value, obtaining an expanded ablation zone. Comparing the damage cloud under the parameter pattern A, B and C, as the four physical properties of the tissue increase during exhalation. Dimensions of the thermally ablated area in terms of long-axis length (LA) and short-axis length (SA) reduced by about 10 mm and 3 mm respectively, as seen from Fig. 10. During microwave ablation process, a safe ablation boundary is required to prevent tumor recurrence, the safety margin is set to 10 mm width [31]. The ablation

Fig. 8 Isothermal surface and damage cloud map of microwave ablation zone under static physical parameters

zone formed under dynamic physical parameter patterns is generally larger than the ablation zone formed under the static physical parameters. Moreover, the LA length of the ablation region increase significantly, almost all the tissue in contact with the antenna is burned. Santos examines the effect of single-lung ventilation on ablation size using a microwave system. In [6], ablation diameters and volume were compared between the ventilated and nonventilated lungs, statistically significant differences were found in the ablated area. Ablation volume was superior in nonventilated lungs. This experiment verifies the correctness of our model. The enhanced back-heating effect of the antenna caused by periodically changing physical properties, which leads to an increased damage to healthy tissue, rather than focusing the antenna radiant energy field near the tip and slotted area. In the normal respiratory frequency range, there is no significant difference in the damage zone under different physical transformation frequencies.

3.2 Temperature increment around the ablation antenna

An ellipsoidal temperature distribution is formed around the microwave ablation antenna, and four temperature sampling

points are arranged perpendicular to the ablation antenna, respectively, 5 mm, 10 mm, 15 mm, and 20 mm from the ablation antenna. The results are shown in Fig. 11, Under the static physical parameters, The temperature at a distance of 5 mm increase sharply first, then the growth rate gradually slows down and finally stabilizes, the highest temperature reached during the ablation process is 88.56 °C, and the standard deviation is 3.44; Under the dynamic physical parameters, the speed of temperature increment at this point is greater than in the static case, the maximum temperature always changes during the entire ablation period, and the highest temperature reached at this point is 122.87 °C, the standard deviation (SD) is 14.31. In [32], the tissue close to the radiating antenna can obtain a very high temperature (up to 120 °C). At 10 mm, under static physical parameters, the highest equilibrium temperature is 54.76 °C, and the standard deviation is 2.41. Under dynamic physical parameters, the maximum temperature at this point is 65.29 °C, the standard deviation is 4.49. At 15 mm, under static physical parameters, the highest equilibrium temperature is 43.47 °C, the standard deviation is 1.07, under the dynamic physical parameters, the maximum temperature at this point is 47.18 °C, also the standard deviation is 1.53. Physical property changes have a greater impact on the point

Fig. 9 Isothermal surface and damage cloud map of microwave ablation zone under dynamic physical parameters

that is close to the antenna. As the distance increases, the rate of temperature rise and the maximum temperature are significantly reduced. Due to the loss of microwave energy propagating in the tissue, the near-field region is mainly due to the active heating region formed by microwave energy deposition. The temperature increment of the tissue further from the antenna is formed by heat conduction. The passive heating zone formed by heat conduction, which is mainly affected by the thermal conductivity of the structure. While

under the dynamic physical parameters, The temperature of the sampling point is not stable throughout the ablation time, especially at 5 mm and 10 mm, this may be due to the fact that in the near-field region of the antenna, the relative dielectric constant and conductivity are two major factors that affect the deposition of electromagnetic energy, which will affect the temperature of the active heating zone.

3.3 The change of fraction of damage tissue with time

Fig. 10 Dimensions of ablation zone under different physical parameters

The fraction of necrotic tissue is shown in Fig. 12. At 5 mm and 10 mm, the formation speed of the necrotic tissue is significantly higher than the other two points, These two points are in the active heating area of the antenna. Protein in tissues is quickly inactivated under high temperature [33]. At the same time, there are slight differences in the rate of damage formation at these two points under different physical parameters, but in the end these two points are completely damaged. At 15 mm, the tissue damage score after 10 min of ablation under dynamic physical parameters is higher than that under static physical parameters, which is in the reversible damage area of the ablation boundary and this area will ultimately affect

Fig. 11 Temperature change of temperature monitoring point in different parameter pattern during ablation

the entire actual ablation zone. At 20 mm, under different physical parameters, the difference in the damage score of the tissue is not very significant, and the state of tissue is in a relatively safe range.

4 Conclusions

This study is based on the 4D-CT image of the lungs to reconstruct three-dimensional models of the lungs during breathing. The change of lung porosity is calculated by measuring the volume of the three-dimensional model of the lung. To study the effect of porosity change on the effective thermal conductivity of lung tissue, and to obtain the change of effective thermal conductivity of lung tissue during respiration. Establishing a lung ablation model, studying the influence of physical parameters on the temperature distribution of the thermal ablation zone leads to the following conclusions:

1. The effective thermal conductivity of the porous medium in the lung increases with the decrease of the porosity, and the porosity of the lungs is the highest at the end of inhalation and the thermal conductivity of the tissue is at least 0.16 W/m °C; the highest value of effective thermal conductivity is 0.20 W/m °C. The results of this study are consistent with the thermal conductivity measured in porcine lungs, which is in the range of 0.125–0.18 W/m °C [34]. The experiment was based on

measurements of isolated porcine lungs, and blood perfusion was not considered during the experiment, and literature suggests that increasing blood perfusion rate will increase the effective thermal conductivity of the tissue.

2. During ablation process, the effects of effective thermal conductivity, relative dielectric constant, electrical conductivity and density on the ablation area during the respiratory process were comprehensively considered. The results showed that at the end of expiration, the electrical conductivity and thermal conductivity of the tissue are at the maximum, and these two values are the key physical parameters affecting the microwave ablation zone. The greater the conductivity of the biological tissue is, the greater the value of the specific absorption rate of the tissue, thus producing a higher temperature; The increased effective thermal conductivity increases the transfer of heat to the tissue surrounding the antenna, thereby expanding the ablation zone. The damage area formed around the microwave antenna is significantly larger than that at the end of inhalation, and the backward heating effect of the antenna get reduced. The ablation area is closer to the tip and the slot area of the antenna, and the SA of the ablation zone get increase. Under dynamic physical parameter patterns, the backward heating effect of the antenna is more significant than under the static parameter pattern as a whole, and LA of damage tissue is among 62–65 mm, while the SA of damage tissue is not effectively expanded. This con-

Fig. 12 Tissue damage fraction of temperature monitoring points in different parameter pattern, These points are located at 5 mm, 10 mm, 15 mm, 20 mm from the antenna

dition will damage more healthy tissue and is not effectively killed for tumors at the center of the puncture site. At different respiratory frequencies, the ablation area does not change much. Therefore, during clinical treatment, it may be considered to block ventilation with single-lung ventilation, so that the antenna obtains a more regular ablation zone under static physical parameters. The results also show that in the ablation parameter pattern C, the ablation zone formed by the antenna is more concentrated at the tip and the slot, and the backward heating effect is significantly weaker than other conditions, and less damage to the healthy tissue around the antenna. Therefore, it can be considered that at the end of expiration, the lungs are ablated when the lungs expel air to obtain the desired ablation result. Those findings can be useful to further our understanding the MWA and

hold promise towards achieving successful treatment in lung tumor ablation.

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Compliance with ethical standards

Conflict of interest The authors declare that there is no conflict of interest regarding the publication of this paper.

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