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An efficient parallel numerical modeling of bioheat transfer in realistic tissue structure

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ABSTRACT

A fast and accurate numerical method of solving bioheat transfer problems is critical for many biomedical applications, such as efficient implementation of thermal therapy planning in a clinical setup, and evaluation of the thermal effects induced by specific energy absorption rate (SAR) from external electromagnetic field. This paper developed a parallel finite-difference method based on alternating direction explicit (ADE) scheme to solve three dimensional transient bioheat equation for the realistic tissue structure. An optimized processing pipeline for accurate presentation of the irregular boundary condition was established to considerably reduce the staircase errors induced by cubic voxel. The voxels order aligned along diagonal direction was designed to allow fast parallel strategy for the lower and upper matrix solution involved in ADE scheme for irregular tissue. The test cases including a real head model have indicated that the developed method can support large time step and cubic voxel approximation of irregular interface and boundary, and also produce significant parallel efficiency.

1. Introduction

The heat transfer in biological tissue is one of the most fundamental physical principles, which significantly affects the biochemical reaction and physiological process. The prediction of the temperature evolution in living tissue is of great importance for modeling of many clinical practices including hypothermia therapy for cancer treatments [1], brain cooling [2,3], or evaluation of the thermal injury caused by external device such as the electromagnetic field of magnetic resonance imaging [4] and cellular phones [5]. Variants of the Pennes bioheat transfer model have been successfully applied to estimate the tissue temperature evolution, which have been validated by experiments [6]. In a simulation framework, however, one of the central questions is to develop a fast and accurate numerical method for solving bioheat transfer equation.

Recently, the simulation approaches for bioheat transfer have been reviewed in [7]. Among the existing simulation methods, the finite difference method (FDM), or denoted by finite-difference time-domain (FDTD), is widely applied for biomedical applications, especially in the context of increasing concern over electromagnetic radiation exposure [4]. One of the reasons is that FDM is easily implemented in terms of coding, where no complex mesh generation process involved in finite element method (FEM) is demanded. In addition, cubic voxel used in FDM can be directly obtained from the medical imaging such as the magnetic resonance imaging (MRI) and the computed tomography (CT), which are also represented by cubic voxel. However, FDM suffers considerably from staircase errors induced by curvilinear boundaries [8], and time-step limitation in explicit scheme [7], which have significant impact on the computational accuracy and efficiency in particular. Samaras et al. [9] proposed a conformal scheme to reduce the staircase effect, which modifies the boundary area to improve the calculation accuracy of heat flux at irregular boundary. The alternating-direction implicit (ADI) approach allow a larger time step to rapidly compute temperature evolution in biological tissue [10]. Our previous work [7] has developed a boundary-source method to efficiently correct the heat transfer of boundary, which was further combined with ADI method to improve the computational efficiency. The immersed boundary method [6] has also been applied to solve complex heat transfer of cryo-freezing tissue. In order to rapidly evaluate temperature increase induced by electromagnetic field, the spatial filter method [11] based on fast Fourier transform has been developed to significantly reduce computational time. However, it cannot capture the accurate thermal effects of irregular boundary condition, which is often important for many applications, such as scalp cooling for thermal regulation

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of brain injury [2]. In addition to developing efficient numerical scheme, parallel method could also speed up computational process. Zeng et al. [12] has developed an efficient parallel ADI method for application in cryoablation. However, the parallel strategy has not been widely used for bioheat transfer simulation because of its complexity, respectively, considering complex structure.

This paper is aimed to develop a simple and efficient parallel alternating direction explicit (ADE) scheme to solve three dimensional transient bioheat equation for the real tissue structure. ADE method has been used for solving heat conduction equation only for the simple cubic geometry [13,14], while it has not been applied for bioheat transfer solution with irregular tissue structure. More importantly, we here developed an efficient parallel strategy for ADE with complex tissue structure based on special voxel alignment and rearrangement method. In addition, the optimized processing pipeline for the irregular boundary condition was established, which could be directly combined with both medical image reconstruction and the developed ADE scheme. The paper is organized as follows: firstly we describe a modified bioheat transfer equation and its ADE discretization considering irregular boundary. Afterwards the parallel method for ADE is presented. The detailed numerical simulation tests are subsequently implemented to validate the numerical scheme developed in current work.

2. Methodology

2.1. Modified bioheat equation considering irregular boundary condition

In our previous work [7], we have constructed a boundary-source $Q_w = T_w – A_w T$ to correct the heat transfer from irregular boundary. Thus a modified Pennes' type bioheat equation considering irregular boundary condition is given by He et al. [7]

$$\rho c \frac{\partial T}{\partial t} = \nabla \cdot [\kappa \nabla T] + \omega_b C_b (T_b - T) + Q_m + Q_e + (T_w - A_W T) \tag{1}$$

where T is the temperature and t denotes the time, ρ is the tissue density, c is the specific heat capacity, k is the thermal conductivity, ω_b is blood perfusion from capillary blood, subscript b denotes blood, Q_m is metabolic heat generation, and Q_e is exterior heat generation such as SAR induced by electromagnetic field exposure. The detailed parameters (T_w and A_w) description involved in boundary heat source will be discussed at next section. The current modified bioheat transfer model requires the adiabatic condition on the tissue boundary, which can be automatically met by enforcing zero thermal conductivity to voxels outside tissue domains. This method allows cubic voxel approximation of the complex tissue structure without explicitly representing the irregular boundary.

2.2. Optimized processing pipeline for irregular boundary condition

For general thermal boundary condition expression $\alpha\partial T/\partial n + \beta T = \gamma$, the parameters T_w and A_w involved in boundary source term can be estimated by He et al. [7]

$$A_W = \frac{S\kappa_w \beta}{V(\beta l + \alpha)}, \quad T_w = \frac{S\kappa_w \gamma}{V(\beta l + \alpha)} \tag{2}$$

where parameters α, β and γ are determined by the detailed boundary conditions, such as $\alpha = 1$ and $\beta = \gamma = 0$ denoting adiabatic condition. The geometric parameters S, V and l denote, respectively, the intersection surface area, volume and distance for the boundary voxel, which are completely determined by the boundary geometry information. Here, we give an optimized processing pipeline for the above three parameters determination, which has four main steps summarized in Fig. 1.

2.2.1. Step 1: representing the boundary with smooth index

The geometric structure of tissue can be obtained from segmentation and reconstruction of medical images such as MRI. Image segmentation provides volumetric quantification of the tissue shape. The binary mask (denoted by index $\phi = -1$ and $\phi = 1$) for each tissue can be determined based on labeled voxels according to tissue type (see Fig. 1A). We can smooth the binary value defined at the voxel vertex to represent the tissue boundary with zero level value [15] (see Fig. 1B). It is noteworthy that the level set based segmentation method can be directly used for boundary representation with zero level set [16].

2.2.2. Step 2: voxel classification

The voxels can be classified as three types: tissue voxel defined by $\phi \leqslant 0$, background voxel for $\phi > 0$, and boundary voxel for other condition (illustrated in Fig. 1C).

2.2.3. Step 3: geometric parameters calculation

For each boundary voxel, the intersection surface can be estimated by isosurface of $\phi = 0$. Firstly, the intersection point (such as x_1 and x_2 in Fig. 1D or 3D case in Fig. A1 of Appendix A) determined by $\phi = 0$ at the edge of boundary voxel is calculated, and the point on the intersection plane $\mathbf{x_c} = M^{-1}\sum_{i=1}^{M}\mathbf{x_i}$ is estimated, where M is the number of intersection points for all the edges of the voxel. $\mathbf{n} = \nabla\phi/|\nabla\phi|$ denotes the normal direction of the plane. Then we can obtain the plane equation by $\mathbf{n} \cdot (\mathbf{X}–\mathbf{X}_c) = 0$. Furthermore we could estimate the above geometric parameters based on the analytical relations connecting with plane equation. We should enforce $l = 0$ for $l < 0$. The detailed calculation of S and V are given in Appendix A.

2.2.4. Step 4: boundary voxel rearrangement

Not all the boundary voxels directly participate in heat transfer calculation. The boundary voxel with small volume (here defined as the smaller than 20% voxel volume) should be discarded to ensure numerical scheme stability. In order to correcting the heat transfer from boundary condition, however, we rearrange the discarded boundary voxel to the adjacent tissue or boundary voxel with large volume (larger than 20% voxel volume), which are both called as active voxel (illustrated in Fig. 1E). For each discarded boundary voxel, the occupied S and V are redistributed to the adjacent active voxel according to the normal direction of the intersection plane. The parameter S would affect the heat flux calculation, and V has impact on the heat generation (such as metabolic heat generation), which would be scaled by the increment volume ratio. A sample of the rearranged increment calculation is given in Fig. 1E.

For voxel with multiple tissue surfaces, we could divide the voxel into many sub-voxels (such as 27 smaller sub-voxels) to calculate the surface and volume of each tissue segment. The geometric parameters S, V and l need to be calculated only once at the boundary voxels. Thus the computational time can be ignored.

2.3. Discrete scheme of diffusion term

The second-order partial derivative of T with respect to x adopts a central-differencing scheme:

$$\frac{\partial}{\partial x}\left(\kappa\frac{\partial T}{\partial x}\right)\Big|_{i,j,k}^{n} = \frac{\kappa_{i+1/2,j,k}\left(T^n_{i+1,j,k} - T^n_{i,j,k}\right) - \kappa_{i-1/2,j,k}\left(T^n_{i,j,k} - T^n_{i-1,j,k}\right)}{\Delta h^2} \tag{3}$$