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Real-time Rendering of Drug Injection and Interactive Simulation of Vessel Deformation Using GPU

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Abstract— **Developing patient specific model for the simulation of chemotherapy drug injection is important in medical application. This paper proposed a two-phase fluidic method to simulate chemotherapy drug injection and an improved lumped element method to simulate deformation of vessel at real-time by using GPU for general computing. Firstly, a three-dimensional (3-D) model of hepatic vessels is reconstructed from clinical CT-images using multi-layer method. A 3-D thinning algorithm based on Valence Driven Spatial Median (VDSM) is applied to generate unit-width skeleton of the vessel tree. The two-phase flow simulation of drug injection is based on Hagen-Poiseuille model by introducing a friction factor using bubbly flow Reynolds number. The improved lumped element method achieves good simulation realism at high computational speed to simulate deformable object. Real-time rendering and interaction of vessel deformation, self collision, and surface tearing has been realized and demonstrated in a virtual experiment.**

I. INTRODUCTION

It has been reported since 1983 that chemoembolization improves survival of patients with unresectable hepatocellular carcinoma (HCC) [1]. This therapeutic method is consist of two procedures: local delivery of chemotherapy, and a minimally invasive procedure called embolization. In the first procedure, the anti-cancer drug is designed to be delivered to the cancerous tumor in appropriate flow rate, which is monitored by MRI fluoroscopy. Therefore, patient-specific simulation and visualization of chemotherapy drug injection is indispensable for the technical success of the treatment. In recent years, several groups have been working on the simulation of one-fluid-phase drug injection based on Hagen-Poiseuille model [2]. On the other hand, an interventional radiologist requires skilled hand-eye coordination to perform complicated surgery such as embolization, and this skill can be effectively enhanced through surgical training in a virtual-reality environment. Physics-based simulation of the hepatic vessel deformation at real time is one of the most difficult part for developing such training system.

This paper introduces a two-phase flow simulation of chemotherapy drug injection in hepatic vessels. The 3-D model of vessel skeleton is reconstructed from clinical CT-images using the method describes in Section II. An improved lumped element method is proposed to simulate vessel deformation using GPU for general computing. An exploratory algorithm based on information theory is also developed to simulate surface tearing of hapetic vessel.

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II. MATERIALS AND METHODS

A. Reconstruction of Hepatic Vessel

Three dimensional reconstruction of hepatic vessel from clinical CT-images is the first step for physics-based simulation. In our implementation, a multi-layer method is applied to extract the raw regions of interest and form the 3-D volumetric model of hepatic vessels. Next, a 3-D thinning algorithm based on Valence Driven Spatial Median (VDSM) is utilized to generate unit-width vessel skeleton which will be used in the drug injection simulation.

1) Reconstruction From CT-images: The shape of hepatic vessel is irregular, but it can be modeled as one or more finite beam elements with circular cross section. The 2-D cross section of hepatic vessel is first segmented from clinical CT-images, and then, Fast Marching Method (FMM) is used to compute and extract the vessel's centerline. The distance transform of the object's boundary is computed by the method, which assumes the boundary evolves in normal direction with constant speed. The centerline points are then determined along discontinuities in the distance transform-points where the moving boundary collapses onto itself. Along this centerline, each layer of the extracted 2-D vessel cross section is assembled into a 3-D volumetric model. The reconstructed 3-D model using multi-layer method is shown in Fig.1(a).

2) 3-D Thinning and Generating Unit-width Skeleton: The refinement of the 3-D model is performed in two rounds. The first round is implemented by using a recursively 3-D Gaussian Filter which shrinks the mesh surface and eliminates the major noise due to inaccurate CT scanning. By varying the number of recursions, different level of refinement could be achieved. In order to obtain a more compact representation of the model, the 3-D thinning algorithm [3] is used to extract its skeleton in a more accurate way. This algorithm has the advantages of preserving the model's topological structure while guaranteeing the obtained skeleton will be close to the medial axis of the region.

However, the limitation of the original 3-D thinning algorithm is that it has problem of generating unit-width curve skeleton, which is desirable for fluidic simulation. Therefore, a post processing procedure based on valence driven spatial median (VSDM) algorithm [4] is applied to eliminate crowed region and generate unit-width curve.

The post processing begins with valance computation, which calculates the *degree* (number of object voxel in its

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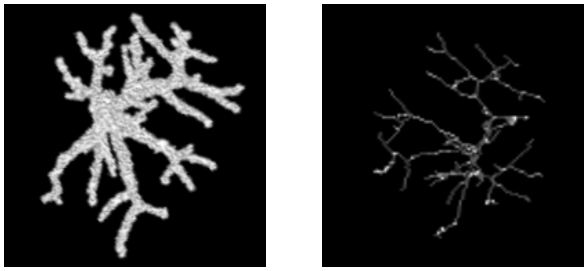

(a) (b)

Fig. 1: Hepatic vessel reconstruction: (a) 3D model from CT images. (b)Unit-width skeleton of vessel tree

26 neighbors) of each voxel on the skeleton. Then, voxels which have degree >2 and at least one of its neighbors has degree >2 are classified as *crowded point*. Next, these crowded points are organized into *crowed regions*, which are sets of 26-connected crowded points. In order to generate a unit-width curve skeleton, all crowded regions must be eliminated. Therefore, a pattern matching is performed to find all object voxels having degree ≤ 2 and 26-adjancent to each crowded region. These voxels are marked with *exit points*. We determine the center of each crowed region using VDSM algorithm. After that, the shortest path between the center and each exit point is calculated using A* searching algorithm. Any crowded point that is not on any of the shortest path is then removed, and the output skeleton is in unit-width curve shown in Fig. 1(b).

B. Flow Simulation

Drug injection through hepatic vessels is assumed to be steady, incompressible and Newtonian. The conventional model treats blood vessels as channels with continuously varying diameters, and chooses a conical pipe model to be the elementary segment in the vessel tree [2]. This geometrical assumption is valid since it resembles the general shape of blood vessel. The problematic part of most drug injection simulations is that they considered the vascular fluid as one-phase flow, whereas the blood in natural human body carries a significant amount of gas bubbles. It has been reported in [5] that the void fraction in human blood varies from 0.386-0.433. The micro-bubbles in laminar blood flow change the near-wall shear stress, and by extension, affect the frictional drag reduction induced by the fluid.

The chemotherapy drug injection is a two-phase porous medium system, and the objective is to simulate its global fluidic behavior. In a two-phase flow, the total pressure drop is due to the variation of kinetic energy and that due to the friction on the walls of vessel. The kinematic pressure drop can be modeled using the Hagen-Poiseuille equation, which is given by

$$\Delta P_k = \frac{8\mu L\dot{m}}{\pi R^4 \rho_h}, \tag{1}$$

where μ is the dynamic viscosity, L is the length of the tube, $\dot{m}$ is the mass flow rate, R is the radius of the tube, and ρ_h is the homogenous density of the two-phase flow, which is calculated in (2),

$$\rho_h = \rho_l(1-\varepsilon) + \rho_g\varepsilon, \tag{2}$$

where ε is the void fraction of the fluid. For a general branch segment that is filled and represented as a conical pipe with radii R_1 and R_2 at its ends, the kinematic pressure drop can be derived as

$$\Delta P_k = \int_0^L (\frac{8\mu\dot{m}}{\pi R^4\rho_h})dx, \tag{3}$$

$$R = \frac{R_2 - R_1}{L}x + R_1, \tag{4}$$

$$\Delta P_k = \frac{8\mu L\dot{m}}{\pi\rho_h}\left[\frac{1}{3}\left(\frac{1}{R_1R_2^3} + \frac{1}{R_1^2R_2^2} + \frac{1}{R_1^3R_2}\right)\right]. \tag{5}$$

Next, the most difficult part of the simulation is the frictional pressure drop. We introduce a *two-phase friction factor* f_{tp} to describe this fluidic phenomenon [6]. For a fully developed steady flow in blood vessel, the frictional pressure drop can be expressed as

$$\Delta P_f = \frac{f_{tp}L\dot{m}^2}{R\rho_h}, \tag{6}$$

where f_{tp} can be expressed in terms of Reynolds number based on the Blasuis equation,

$$f_{tp} = \frac{0.079}{Re^{0.25}}. \tag{7}$$

Different from the conventional way of using the Reynolds number of liquid to represent the corresponding characteristic of the two-phase flow, we calculate the bubbly flow Reynolds number in a more accurate way in (8),

$$Re = \frac{2\dot{m}(1-\varepsilon)R}{\mu}. \tag{8}$$

The quality averaged viscosity is then derived based on the void fraction

$$\mu = \varepsilon\mu_g + (1-\varepsilon)\mu_l, \tag{9}$$

where μ_g and μ_l are the corresponding dynamic viscosity of gas and liquid in the two-phase fluid. Hence, by integrating (6) along the conical tube, the frictional pressure drop can be derived,

$$\Delta P_f = \frac{0.316\mu^{0.25}\dot{m}^{1.75}L}{\rho_h(2-2\varepsilon)^{0.25}}\left(\frac{1}{R_2-R_1}\right)\left(\frac{1}{R_1^{0.25}} - \frac{1}{R_2^{0.25}}\right). \tag{10}$$

The total pressure drop along a conical tube segment $\Delta P = \Delta P_k + \Delta P_f$. The two radius and the length of the segment are obtained in the vascular reconstruction step. Hence, by input the mass flow rate of drug injection, the partial length of each segment that is filled by fluid at each time step can be simulated.

C. Physics-based Interactive Simulation of Hepatic Vessel Deformation

1) Improved Lumped Element Method: High-fidelity simulation of the flexibility and deformability of blood vessel is important in medical simulation. The conventional way to simulate such soft-bodied object is Finite Element (FE) method, which decomposes the object into a number of volume elements, and uses continuum mechanics model to simulate its biomechanical property. FE method is generally computationally intensive and only the simplest variants such as linear shape function have been employed in real-time simulation [7].

Our objective is to simulate complicated, large-scale and user-controlled interactions between input device and the deformable hepatic vessel model. This requires dynamic simulation of soft body, which is computationally intensive, and complex interaction, which is unpredictable and discontinuous in time. Based on these requirements, an improved lumped element approach is proposed. The method focuses on balancing simulation realism and computational speed. Although there is a small trade-off in accuracy, the method is sufficient to capture the underlying physical phenomena of soft tissue in a computationally efficient way by using GPU for general computing.

The lumped element method treats a soft-bodied object as tetrahedral meshes with two principle parameters: damping ratio and stiffness. This allows us to simplify the description of the behavior of the spatially distributed physical systems into a topological structure consisting of discrete entities that approximate the behavior of the distributed system [8].

An implicit time integration approach is utilized to compute the non-linear equation of the system at each time step. This is achieved by using topological Backward Euler to update the velocities and positions of all vertices. Since stiffness and damping ratio are assumed to be the principle parameters of the soft body deformation, a first-order approximation in (11) is performed to simplify the description of interaction between each meshed segment,

$$f(x_{t+\Delta t}, v_{t+\Delta t}) \approx f(x_t, v_t) + \mathbf{B}\cdot(v_{t+\Delta t} - v_t) + \mathbf{K}\cdot(x_{t+\Delta t} - x_t) + \mathbf{E}\cdot(v_{t+\Delta t} - v_t), \tag{11}$$

where $\mathbf{B}$ is the damping ratio, $\mathbf{K}$ is the stiffness, and $\mathbf{E}$ is the velocity dependent error factor. The macroscopic stiffness of the blood vessel is obtained by empirical relationship. The property of damping ratio B and the velocity dependent error factor E can be integrated into a combined damping coefficient $\tilde{\mathbf{B}}$ using linear regression method, and the the final linearization of the system is shown in (12),

$$(\Delta t \cdot a_{t+\Delta t})\cdot(\mathbf{M} - \tilde{\mathbf{B}}\Delta t - \mathbf{K}\Delta t^2) = f(x_t, v_t)\Delta t + \mathbf{K}v_t\Delta t^2. \tag{12}$$

2) Soft Body Tearing Simulation: Realistically simulating surface tearing of soft tissue is important in surgical simulation. In our method, tearing simulation of a soft-bodied vessel is performed by using a tear factor based on the biomechanical property of the respective tissue. The magnitude of the tear factor is utilized to define force limit on the non-linear spring joint between each meshed segment so that break could occur when forces applied on the joint exceed the limit.

Allowing virtual joints in the model to be broken is the first step of tearing simulation, and the next step is defining the threshold to control the moment when the tearing should stop. Different from the conventional way of using the number of vertices in the system as the threshold, we define the system entropy to be the threshold, which is given by

$$S = c\sum_i p_i \ln p_i, \tag{13}$$

where S is the entropy of the system, c is the Boltzmann constant and p_i is the probability that the system is in the *i*-th microstate. After tearing occurred, new vertices will be generated in the model since the steady state is now becoming a disordered system due to its highly irregular shape under twisting, elongation and contraction. This dramatically increase in the randomness of system will increase entropy of the model. When the threshold in the corresponding buffer cannot hold the entropy generated by the new vertices anymore, tearing stops.

3) Parallelization and Implementation on GPU: Task parallelization on GPU is performed to optimize the intensive computation in simulating vessel deformation. The idea is motivated by the recent advancement in GPU architecture. In the past decade, newly developed multiple computation units of multi-cores GPU can execute numerous threads in parallel. This architectural improvement impels new types of algorithms for complex soft body simulation.

The topological relationship in (11) and (12) involves computing a large sparse system matrix (zero entries) which is very computationally expensive, and real-time simulation cannot be achieved with typical workstation. Hence, the method of assembly by nonzero entries using global memory [9] is used to optimize the simulation. Specifically, one computation thread is used to compute one tetrahedron and the result is stored independently in the corresponding vector matrix, and the forces on each vertex are accumulated using a pre-computed array that stores each vertex's indices of all connected tetrahedrons. By using GPU parallelization for general computing, real-time simulation of blood vessel deformation is realized.

III. Results and Discussions

A NVDIA Quadro FX 4600 graphics card in an Intel Xeon CPU @3.00GHz was used for the simulation of chemotherapy drug injection and the deformation of hepatic vessel. Fig. 2 shows the rendered flow visualization of drug injection in four discrete time steps using the proposed two-phase flow model. In order to evaluate the improved lumped element model in the context of real-time simulation, an experiment which applies external virtual force on a segment of hepatic vessel is designed by using mouse as input device. The result

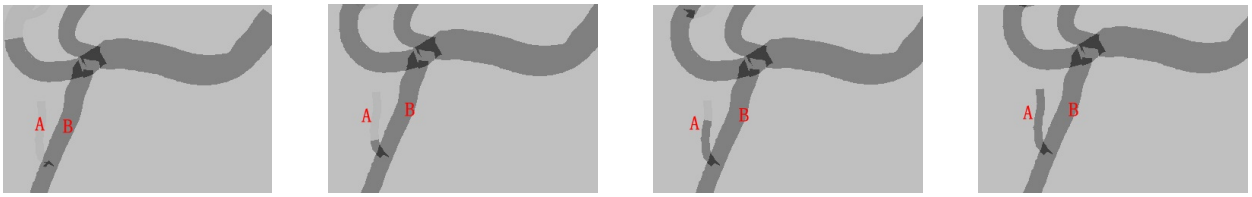

Fig. 2: Visualization of drug injection into vessels in four discrete time steps

shows that the maximum number of contact points that can be detected in real-time simulation is 128 , and they consist of 768 constrains. When the same simulation was performed in CPU-only system, only 8 contact points are required to be detected to maintain real-time simulation.

Fig.3 shows the surface tearing simulation of hepatic vessel under a sufficient large force. To ensure consistent measurement, a pre-defined cut-line in the center was used to progressively remove a section of vessel. It is observed that the whole process is performed under realistic and stable interactions.

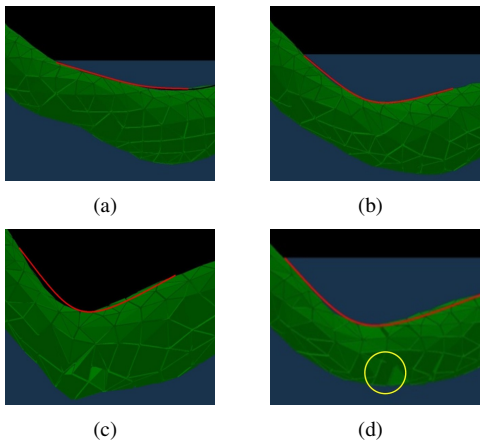

Fig. 3: Tearing simulation of vessel (The red line indicates the curvature of deformation for the hepatic vessel under a sufficient large force, and the yellow circle in (d) highlights the part where surface tearing occurred)

Tissue tearing in blood vessel is essentially a energy releasing process, and hence, the use of entropy as the tearing threshold closely resembles this biological mechanism. Conventional way of using the number of vertices as the threshold is basically a solid threshold with no regard to the law of physics. Entropy threshold claims more flexibility in simulating the deformation of soft tissue, and more comprehensively describing the structure randomness and energy state of the model.

IV. Conclusion

In summary, the proposed reconstruction method is shown to be adequate in obtaining high quality hepatic vessels model from CT-images. The post processing technique based on VDSM algorithm generates a unit-width skeleton which is suitable for fluidic simulation. The two-phase flow simulation of drug injection provides a fast and accurate estimation of overall injection pressure and flow rate.

Besides, realistic simulation of vessel deformation, self collision and surface tearing has been achieved using the proposed lumped element method. Our experiment results show that the simulation using GPU provides an over 20 fold speed-up than that of CPU-only system. Ease of parallelization on GPU, and the ability to balance simulation realism and computational speed are the main advantages of the proposed method.

For future work, the two-phase flow model can be improved by taking into account the frictional head loss due to the turbulent mixing of drug and blood, so that a more precise simulation of drug injection could be achieved. The improved lumped element method can be extended to other meshed models such as hexahedral elements, and there are some new reduction algorithm could be applied to optimize the parallel computing on GPU [9]. We are currently developing a hybrid system with haptics, which could be used for medical training of drug injection.

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