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NUMERICAL INVESTIGATION OF LIQUID EMBOLIZATION FOR INTRAVASCULAR TREATMENT USING A PARTICLE METHOD

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ABSTRACT

Subarachnoid hemorrhage is caused by cerebral aneurysm rupture. Since subarachnoid hemorrhage has not only high mortality but also high possibility for complications, it is important to prevent aneurysm rapturing by performing an intravascular treatment.

The advantage of the recently developed photocurable liquid embolization is its controllability compared to conventional treatment such as coiling or clipping. However, as the treatment has not been evaluated extensively, this study presents a nuermical method to investigate applicability of intravascular treatment using photocurable liquid embolization.

The process of photocurable liquid embolization is complicated and involves both geometrical and topological changes like separation. Therefore, a particle method, particularly the moving particle semi-implicit method was used in this paper. A stabilization method was investigated to reduce numerical instability caused by pressure oscillations. In addition, we developed an efficient numerical method that considers the effects of highly viscous fluids of liquid embolization and wettability of a catheter. The proposed method was applied to a simplified injection problem involving highly viscous fluid to replicate intravascular treatment. The simulation results were compared to experimental data and showed good agreement with the experiments.

Keywords: Moving Particle Semi-implicit method, highly viscous fluid, wettability, photocurable liquid embolization. Cerebral aneurysm

NOMENCLATURE

A_0	[m^2]	interfacial area occupied. by one particle
C	[N/m^2]	potential coefficient
F	[N]	force
H	[-]	approximated Heaviside function
N'	[m^2]	number of particles in the radius r_e^{ave}
P	[Pa]	pressure
S	[-]	coefficient for interface
d	[-]	dimension number
g	[m/s^2]	gravitational acceleration
h	[m]	interfacial thickness
n	[m^2]	particle density
n^{in}	[-]	normal vector to the interfacial surface
r	[m]	distance between particles
u	[m/s]	velocity
w	[-]	weighted function
λ	[1/m$^{2]}$	coefficient of the Laplacian term
μ	[P・s]	viscosity coefficient
ρ	[Kg/m^3]	density
ϕ	[-]	scalar

Subscripts and Superscripts

i, j	*x*-axis, *y*-axis, and *z*-axis coordinate system
k, l	particle number
inter	interfacial
$\overline{kl}$	variable between partilce k and particle l
e	effective
av	average
ad	adhesion
sp	spread
in b-c	interface between blood and embolization liquid
in l-c	interface between embolization liquid and

	catheter
k-st	particle in the structure (catheter)
k-l	particle in liquid (embolization liquid)
k-sl	particle in the interface between structure and liquid

1. INTRODUCTION

Subarachnoid hemorrhage is a serious cerebral disorder. Due to high mortality rate and sever subsequent complications after the incidence of subarachnoid hemorrhage, a patient at high risk is often recommended surgery [1]. Since subarachnoid hemorrhage generally occurs in patients with a cerebral aneurysm, surgery is necessary to prevent the aneurysm rupture. There are mainly two types of conventional surgical treatments: coiling and clipping. Coiling is less invasive than clipping and therefore more widely used. However, it is an expensive procedure that requires a large number of expensive coils. Further, it is difficult to estimate the number of required coils, and once inserted into the aneurysm, the coils cannot be retracted. In view of the limitations of coiling, liquid embolization methods like Onxy have gained attention [2]. In particular, photocurable liquid embolization has been developed as a novel treatment. The advantage of photocurable liquid embolization is its controllability compared to the conventional treatments.

Photocurable liquid embolization is still under development for a clinical application. A numerical simulation is a variable technique to elucidate the mechanism as well as to examine the procedure. However, replicating the flow behavior of photocurable liquid embolization by simulation is quite challenging. Since it is injected into aneurysm from catheter, its process is quite complicated and involves geometrical and topological changes like separation. Therefore, the simulation is required to perform not only flow analysis but also analyze interaction between blood flow and liquid for emolization and that between liquid for emolzaition and catheter with consideration of wettability.

To solve these complex issues, liquid embolization process was simplified. Specifically, to capture essential physics, viscous fluid imitating embolization liquid was injected into water instead of blood through a Teflon tube assumed to be a catheter. Since the embolization happened through topological changes, the moving particle semi-implicit (MPS) method was used in this study [3]. A stabilization method was investigated to reduce numerical instability caused by pressure oscillations [4-6]. In addition, we developed an efficient numerical method that considers the interfacial tension between water and highly viscous fluids of liquid embolization as well as the iterfacial tension with the wettability of the viscous fluid on the catheter.

There are mainly two types of interfacial models. The first one uses CFS model developed by Nomura et al. [7] , whcih considers a volume force. The other one uses the potential model developed by Kondo et al. [8] , which formalizes an intermolecular force. However, theses models are designed for force in single phase flow, and not for liquid-liquid two phase flow as in the present study. To cosider the influece of interaction between two-phase flows in the interfacial bounday, this study developed a method based on the potential model developed by Kondo et al., [8] combining it with the interfacial tension model developed by Ishii et al. [9].

Therefore, this paper aims to develop a new numeircal technique based on the MPS method to simulate the injection of embolization liquid into aneuryms by considering the interaction between blood and liquid embolization as well as between liquid embolization and catheter with wettability. In order to validate the prposed method, the simulation results were compared to experimental data, in which the problem was simplified as a viscous fluid injected into water through a cylidical Teflon tube [10].

2. NUMERICAL METHOD

In this study, the MPS method was used to simulate the injection of a highly viscous fluid into water through a cylindrical Teflon tube, mimicking the injection of embolization liquid into an aneurysm through the catheter. Since this is a complex multi-physic problem, the simulation not only models the flow of water and highly viscous fluid but also the interaction between them. In addition, the present study developed the interfacial tension model to account for interactions between water and the viscous fluid as well as one between the viscous fluid and cylindrical tube, while considering wettability effects.

2.1. Governing Equations

The present method was applied to an injection problem involving a highly visous fluid to replicate intravascular treatment. Therefore, flow simulations of incompressible Newtonian fluid are performed for water and viscous fluid. The governing equations consist of the continuity and Navier-Stokes equations as follows:

$$\frac{\partial u_i}{\partial x_i} = 0 \tag{1}$$

$$\frac{\partial u_i}{\partial t} + u_j \frac{\partial u_i}{\partial x_j} = \frac{1}{\rho}\frac{\partial P}{\partial x_i} + \frac{\mu}{\rho}\frac{\partial^2 u_i}{\partial {x_j}^2} + F_i \tag{2}$$

The external body force Fi is given by

$$F_i = \rho g_i + F_i^{inter} \tag{3}$$

The method to determine force F_i^{inter} is described in the later section.

2.1. MPS Method

The MPS method is based on particle interactions, discretizing the partial differential equation using a weight function to calculate an averaged interparticle distance. This study uses the weight function developed by Tamai et al. [11], which is given by

$$w^{\overline{kl}} = \begin{cases} \left(\dfrac{r^{\overline{kl}}}{r_e^k} - 1\right)^2 & \left(r^{\overline{kl}} \le r_e^k\right) \\ 0 & \left(r_e^k \le r^{\overline{kl}}\right) \end{cases} \tag{4}$$

where $r_{\overline{kl}}$ is the distance between particle k and l, and the effective radius r_e^k is the radius of influence for the particle k. In this study, r_e^k is set as 2.1 times the initial interparticle distance in the gradient and the divergence models while it is set as 3.1 times in the interfacial tension models.

The MPS method requires the density of particles, which is obtained by

$$n_k = \sum_{k \ne l} w^{\overline{kl}} \tag{5}$$

The particle interaction model comprises the gradient, the divergence, and the Laplacian terms. Each term of particle k for an arbitrary scalar ϕ^k can be described in the following partial differential operators:

$$\frac{\partial \phi^k}{\partial x_i} = \frac{d}{n^0} \sum_{l \ne k} \frac{\phi^l - \phi^k}{\left(r^{\overline{kl}}\right)^2} \left(r_i^l - r_i^k\right) w^{\overline{kl}} \tag{6}$$

$$\frac{\partial \phi^k}{\partial x_j} = \frac{d}{n^0} \sum_{l \ne k} \frac{\left(\phi^l - \phi^k\right) \cdot \left(r_j^l - r_j^k\right)}{\left(r^{\overline{kl}}\right)^2} w^{\overline{kl}} \tag{7}$$

$$\frac{\partial^2 \phi^k}{\partial {x_j}^2} = \frac{2d}{\lambda n^0} \sum_{l \ne k} \left(\phi^l - \phi^k\right) w^{\overline{kl}} \tag{8}$$

where λ is defined as

$$\lambda = \frac{\sum_{k \ne l} \left(r^{\overline{kl}}\right)^2 w^{\overline{kl}}}{\sum_{k \ne l} w^{\overline{kl}}} \tag{9}$$

When the governing Eqs. (1) and (2) are solved, numerical instability may increase depending on arrangement of particles. Therefore, the high-order MPS is used to avoid numerical instability. In general, the pressure Poisson equation is solved to obtain pressure. We applied the high-order method to the Laplacian term of pressure in the Poisson equations using the method based on Tanaka et al. [4-6]. In addition, the high-order gradient model developed by Iribe et al. [5] was applied to the pressure gradient term.

2.2. Interfacial Tension Model

The two types of interface force considered in this study are the one between blood (water) and liquid for embolization (viscous fluid) and the other between catheter (cylindrical Teflon tube) and liquid for embolization (viscous fluid). Therefore, when the interfacial tension F_i^{inter} is calculated, it is necessary to consider two types of the interfacial tension models :1) the one between liquid and liquid and 2) the other between solid and liquid considering wettability.

In this study, the simulation is simplified as described in the previous section. Figure 1 describes the interfacial force between water and viscous fluid. A blue particle represents water, and a red represents viscous fluid.

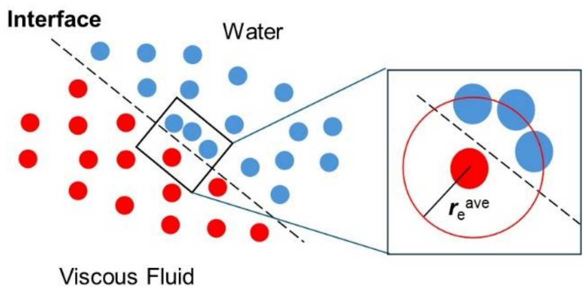


Figure 1. Schematic illustration of the interfacial force between blood and liquid for embolization

Let us consider force balance in the region within the radius of influence r_e^{ave} as shown in Fig.1 The interfacial tension model in the present study is based on the potential model [9]. The interfacial force between water and viscous fluid is given as follows:

$$(F^k)_i^{in\,b-e} = \frac{d\left(P^v - \overline{P^b}\right)}{h} n_i^{k-in} \tag{10}$$

$$P^b = \frac{1}{N'} \sum_{l \ne k}^{N'} \overline{P^b}\, w'^{\overline{kl}} \tag{11}$$

$$w'^{\overline{kl}} = \begin{cases} 1 & \left(r^{\overline{kl}} < r_e^{av}\right) \\ 0 & \left(r_e^{av} \le r^{\overline{kl}}\right) \end{cases} \tag{12}$$

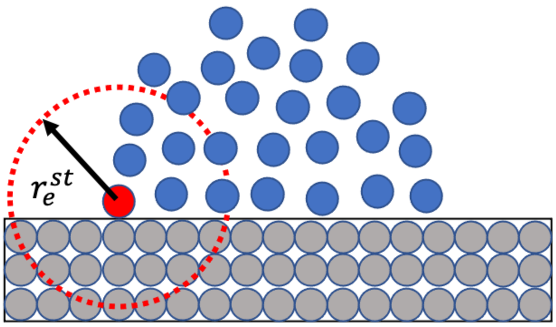
Since this study deals with the contact between a viscous fluid and cylindrical tube, it is important to include the effects of wettability. In general, the contact angle of the embolization liquid on the catheter is used to characterize wettability. However, it is difficult to obtain the angle. Therefore, this study developed a method to replicate the wettability by combining adhesion and spread of liquid on the solid surface of the catheter.

First, the particles at the interface between viscous fluid and cylindrical tube need to be distinguished. The following normal vector is used to identify the particles on such interface:

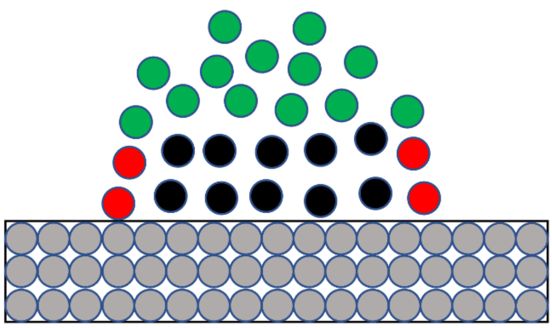
$$n_i{}^{k-in} = \begin{cases} \dfrac{F_i^{k-l}}{\left|F_i^{k-l}\right|} & \left(\left|F_i^{k-l}\right| \middle/ \left|F_i^{k-l}\right|^{flat} \geq \eta \right) \\ 0 & \left(\left|F_i^{k-l}\right| \middle/ \left|F_i^{k-l}\right|^{flat} < \eta \right) \end{cases} \quad (13)$$

In this paper, η is set to be 0.2[9].

Next, the identification of particles at the interfacial surface is performed to distinguish the particles, where wettability needs to be considered. Figures 2 (a) and 2 (b) show schematic illustration of how particles are identified using the effective radius r_e^{st}. The red particles in Fig. 2 (b) are required to consider wettability while the black particles represent the wall of catheter, and the green ones are the reference particles.



(a) Before identification



(b) After identification

Figure 2. schematic illustration of identification of particles subject to wettability

If the effects of adhesion in wettability are strong, the force balance can be expressed as follows:

$$F_i{}^{k-ad} = F_i^{k-st} + F_i^{k-l} + F_i^{k-sl} \quad (14)$$

On the other hand, when the effects of spreading out are stronger, the force balance becomes

$$F_i{}^{k-sp} = F_i^{k-st} - F_i^{k-l} - F_i^{k-sl} \quad (15)$$

Each term in Eqs. (14) and (15) can be obtained using the following poteital model:

$$F_i{}^{k-\alpha\alpha} = C^{\alpha\alpha} \sum_{l \neq k} f^{kl} \frac{r_i^l - r_i^k}{r^{\overline{kl}}} \quad (16)$$

The interfacial potential between viscous fluid and cylidical tube can be obtained by cobining $F_i{}^{k-ad}$ and $F_i{}^{k-sp}$, which is given by

$$(F^k)_i^{in\ l-c} = F_i{}^{k-ad} - HF_i{}^{k-sp} \quad (17)$$

Therefore, the interfacial tension model in this paper, which comprises the interfacial tension model between liquid and liquid and the one between solid and liquid considering wettability, can be defined as follows:

$$F_i^{k-inter} = S^k \frac{d}{hA_0} \left[\left(\frac{1}{\frac{1}{N} \sum_{l \neq k} \sin \varphi^{\overline{kl}}} - 1 \right) \right] (F^k)_i^{in\ s-c} \quad (18)$$

where S^k is set as 1.0 for convex shape and -1 for concave shape.

3. RESULTS

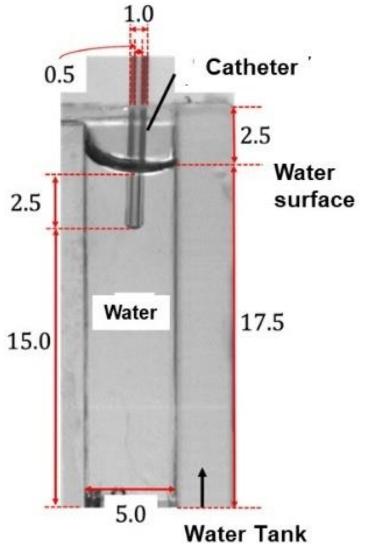
The simulation results were compared to the experimental data to validate the present numerical method. The conditions for the experiment were applied for the numerical simulations. In this study, epoxy resin was used as embolization liquid and a Teflon tube was used as catheter in the experiment [10]. Table 1 summarizes the experiment and simulation conditions.

Table 1. experimental and Simulation conditions

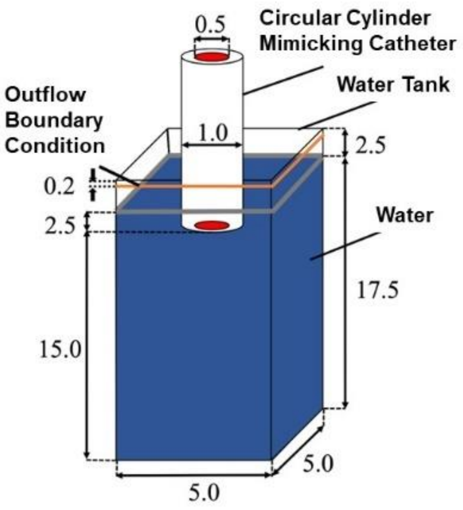
Physical Properties		
Density [kg/m^3]	Water	1.00×10^{3}
	Embolization Liquid	1.18×10^{3}
Viscosity Coefficient [Pa・s]	Water	1.00×10^{-3}
	Embolization Liquid	7.42×10^{-3}
Inflow Velocity [m/sec]		8.50×10^{-3}
Gravitational acceleration [m/s^2]		9.8

The particle size is 1.00 $\times$ 10^{-1} mm in the MPS simulation in this study.

The experimental setup is shown in Fig. 3(a). The simulation was performed using the analysis model as described in Fig. 3(b) under the same conditions as the experiment.



(a) Experimental model



(c) Analysis model

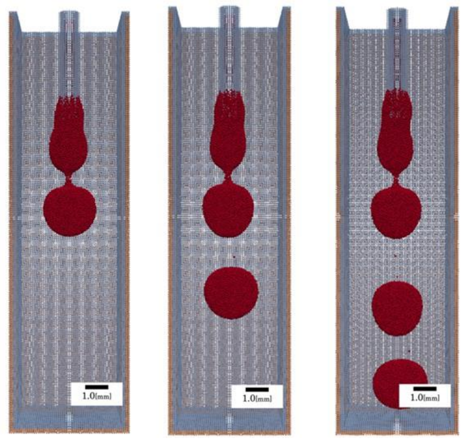
Figure 3. Experimental and analysis models

In order to examine the effects of wettability, a comparison was performed between simulation results obtained without and with considering wettability. Figures. 4 (a) and 4 (b) present the results of comparison.

Although it was difficult to compare behavior of two simulations at the same instantaneous time, the results were sufficiently close for comparison. If wettability is ignored, embolization liquid tends to be elongated due to gravitational force. On the other hand, if wettability is considered, embolization liquid tends to spread out at the edge of circular tube against gravitational force, which makes embolization liquid less elongated, as described in Fig. 4(b).

12.6 sec 22.9 sec 34.8 sec

(a) Without wettability



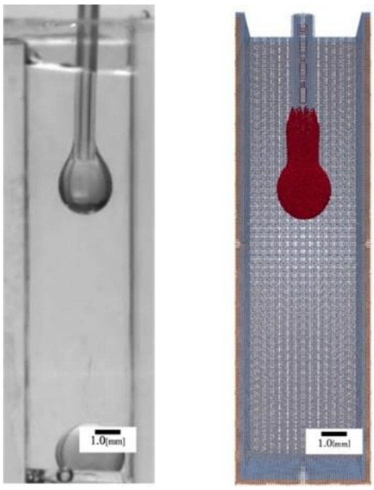
10.0 sec 15.6 sec 21.4 sec

(b) With wettability

Figure4. Comparison for investigation of wettability

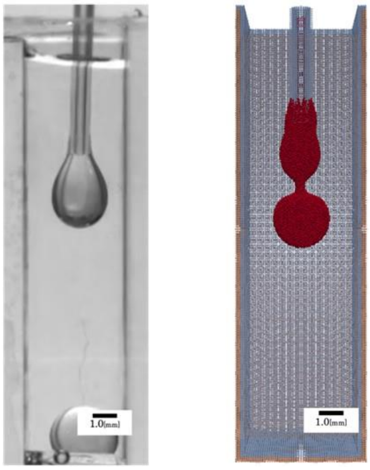
Next, we compared the simulation results with experimental data. First, we compared the separation time, which is the time taken by a droplet to detach from the cylindrical Teflon tube. The experiment was performed six times while the simulation using the present method was performed three times. The average separation times for the experiments and simulation were 10.32 sec and 7.09 sec, respectively. The results of droplet volume for experiment and simulation were also compared. The average droplet volume was calculated over three trials for both the experimental measurements and simulations. The averaged droplet volume of simulation was 8.84$\times$ 10^{-9} m^3 while it was 1.8 $\times$ 10^{-8} m^3 in the experiment. Both droplet volume and separation time were smaller than those in the experiments. Hence, separation in simulations tends to occur faster than that in experiments. This is caused by the underestimation of the interfacial tension compared to the gravitational force.

Let us compare the process of droplet formation at instantaneous times for both experiment and simulation. The images on the left in Figs. 5 (a) and 5(b) are the experimental measurement data while ones on the right are the simulation results. Both experimental data and simulation results are compared at the same instantaneous times 7.00 sec and 10.0 sec.



Experiment Simulation

(a) Instantaneous time at 7.00 sec



Experiment Simulation

(b) Instantaneous time at 10.00 sec

Figure 5. Comparison between experiment and simulation

As illustrated in Figs 5, the droplet tends to form faster than in the experiment, which leads to shorter droplet formation time and a smaller volume.

5. SUMMARY

In order to investigate flow characteristics of photocurable liquid embolization as a novel treatment for cerebral aneurysms, a simulation method has been developed based on the MPS method to solve topological changes. In addition, the interfacial tension model was developed. Since the treatment involves injecting embolization liquid into blood in the aneurysm, the interfacial tension needs to consider two types of interactions :1) the one between blood (water) and embolization liquid (viscous fluid) and 2) the other between embolization liquid (viscous fluid) and catheter (cylindrical Teflon tube) with wettability. The proposed interfacial tension model was incorporated into the MPS method and applied to the problem similar to the experimental setup, in which viscous fluid was injected into water through the cylindrical Teflon tube. The experiment and simulation were conducted under the same condtions and compared. The simulation results showed relatively good agreement with those of the experiment. However, the simulation results showed a tendency to underestimate the interfacial force compared to the experimental results. Therefore, volume and separation time in the simulation were smaller than those in the experiment.

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