Simulation of angiographic contrast agent propagation during virtual endovascular interventions

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***Abstract*—Angiography uses injection of contrast agent into blood vessels to visualise vascular anatomy and pathology for diagnosis and instrument guidance during endovascular procedures. Virtual simulators, introduced to acquire interventional radiology skills, rely heavily on virtual angiographic imaging. We propose a methodology for simulating contrast agent injection in three dimensional virtual vasculatures using a syringe haptic device. The contrast agent is modelled as a line of particles with adaptive radius propagating in the vessels and constrained by three forces produced by the initial injection, the collision with vascular walls, and the blood flow. Propagation is based on a vascular centreline giving the position of bifurcations, vascular network extremities and their radius. The blood flow, computed according to Poisseuille law, influences the contrast agent speed and opacity. The injection of contrast medium in three patients at five different vascular positions showed realistic rendering at an average of 20.8 frames per second.**

Keywords-Endovascular intervention; angiography; blood flow; contrast agent simulation; smoothed particle hydrodynamics.

I. INTRODUCTION

Vascular imaging (angiography) uses injection of contrast media to opacify blood vessels for diagnostic purposes, delineating vessel anatomy and any pathology. During angiography, the injected contrast medium is carried away by the blood flow and circulates through the vascular system. In interventional radiology (IR), minimally invasive procedures use this angiographic imaging to guide needles, catheters and guide-wires through vascular anatomy via tiny, pinhole incisions. Competency in IR vascular imaging techniques is attained over several years in an apprenticeship involving real patients: this training method is both expensive and carries a risk of complications. Computer-based simulation is an alternative to improve IR skills training, offering a safe framework to practice specific skills as often as needed.

The Simulation Group has worked on simulation tools for IR including an interactive catheter simulation and blood flow and contrast agent propagation via a 1D laminar flow model in a volumetric representation and multi-scale approach [1]. At Stanford, a surgical simulator has been implemented with realistic simulated blood using particle hydrodynamics [2]. This paper presents a new approach to model realistic 3D contrast agent injection and its interactions with blood flow. This work is part of a large effort (http://www.craive.org.uk/) aimed at training the IR core skills of navigating medical instruments within the vasculature.

II. MATERIALS AND METHODS

A. Contrast medium modelling

The contrast medium is discretised as a set of 3D particles. Each particle has a position, a velocity, a colour, a shape, a mass and a lifetime. The colour of a particle is black, and it fades as the contrast agent mixes with the blood. The particle shape is a sphere of adaptable radius to completely fill the vessel lumen. All particles have the same mass m.

Fig. 1 illustrates the particle line initialization and propagation after being injected at the catheter tip, noted as X0...X4. At the injection time *t0*, two contrast agent particles are created: P0 and P1. P0 is created at X1. Its radius is very small so that it will not be located outside the vasculature. P1 is created at the centre of the cross-section located in X0's plane. The radius of P1 is initialized according the cross-section radius at that point in order to fill the vessel.A new particle is created at X1 (same radius as P0) after the time step dt, while P0 takes the position of P1 and P1 moves to P2, at a distance L of P1, L being smaller or equal to the radius of P1 . The new position P2 is controlled by three forces, *Fd*, *Fb* and *Fe*. *Fd* represents the injection power. Its normalized direction *d* is obtained from the previous position (P1 Fig. 1) and the current particle (P0 in Fig. 2). The magnitude of *Fd* is related to the average injection speed *v*.

$$Fd = d.v \quad (1)$$

Fc originates from the collision response, discussed in section 2.3. Its direction is given by the opposite of the average normal *n* of the colliding vascular triangles, and its magnitude is proportional to the distance *dc* from the colliding triangle.

$$Fc = -n.dc \quad (2)$$

Fb represents the influence of the blood flow. Its direction *d* is the same (k=1) or opposite (k=-1) to that of *Fd*, depending on the contrast medium direction. The magnitude of *Fb* is related to the magnitude of blood flow velocity as described in section 2.4.

$$Fb = k.d.vb \quad (3)$$

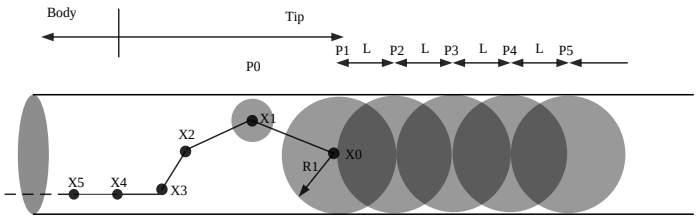


Fig. 1. Contrast agent particle line initialization at X1 and X0 and propagation towards P5.

At the next time step *dt*, a new particle is created at X0 and the other particles are moved forward. This process is repeated until the injection stops. If the contrast medium stays in one vessel, keeping one particle line is enough. Otherwise, new particle lines should be created at each bifurcation as discussed in section 2.5.

B. Radius pre-processing

Control The propagation is based on a centreline representation of the vasculature inspired by that in [6]. Control points are extracted during a pre-processing step and used to mark the beginning and end points of the vascular centreline and the bifurcation as shown in Fig. 2. This pre-processing allows speeding up the contrast agent initialization at time *t0* by computing the initial cross-section radius of the particle P1 (Fig. 1) as shown in Fig. 2. If the initial particle is between two control points (beginning, end, or bifurcation points - RP1 in Fig 2.)), its radius will be the average radius of the two control points. If the initial particle is between three control points of a bifurcation (RP2 in Fig. 2), its initial radius will be the average radius of the three control points.

This pre-processing is used only for initializing the radius of particle P1. The other particles have their own radius, estimated and adapted via the collision detection.

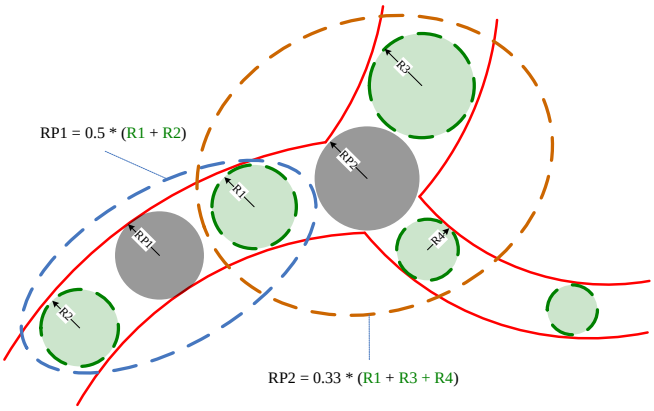


Fig. 2. Two cases occur when computing the cross-section of the initial particle (solid circles RP1 or RP2) using the centreline control points (dashed circles noted R1..R4).

C. Collision detection and collision response

Our collision detection partitions the vasculature in an Axis Aligned Bounding Boxes (AABB) tree before the simulation starts, and a bounding sphere around each contrast agent particle. When an intersection is detected between them, a collision is assumed. The collision response computes the external force *Fc* applied by the vasculature on the colliding particle(s). The collision detection checks if the contrast agent particles are within the vessel walls, while the collision response system ensures they stay inside the vessels, adjusting their radii to make the particles fit the vascular cross section entirely. . This adjustment could not be based on the centreline as it is not accurate enough across the whole length of the vessels. The particle radius is increased gradually at each time step *dt* if the particle failed to collide with the vessel walls. Conversely, the particle radius is reduced gradually at each time step if a collision is detected. Given that *dt* is small, the particles are able to cope with a satisfying radius change. Nevertheless, abnormal cases with sharp radius change (e.g. aneurysm or stenosis) would not be properly filled. In such cases, multiple centreline control points (typically three) need to be placed in the abnormality in order to smooth the radius adaptation, thus ensuring that contrast agent can always nearly fill the cross-section of the vasculature.

D. Influence of Blood flow

The main influence of the blood stems from the beating heart that results in the blood speed continuously changing from 0 to its maximum value at each cycle. A simple model of blood velocity *vb* at a time *t* for a heart beating at a rate *f* with volumetric flow rate φ can be given by a sine function which contributes to the real time rendering of the simulation:

$$vb = \varphi \,.|sin\,(t.\pi.f)| \qquad (4)$$

This function allows fast computation of the blood speed at each time step and therefore contributes to the real time rendering of the simulation. Using a more complex function could significantly slow down our simulation. The volumetric flow rate φ is given by Poisseuille's law. It assumes that the fluid flow is laminar viscous (η - dynamic fluid viscosity) and incompressible, through a constant circular cross-section (radius *R*) much smaller than the branch total length *L*:

$$\varphi = ((\pi R^4)/8\eta).(|\Delta P|/L) \qquad (5)$$

where ΔP is the pressure difference between its two ends. This rate is therefore variable along each branch. It is pre-computed prior to the simulation and stored on the centreline control points. The branch length is computed by adding the distance between centreline points from one control point to the other. The pressure drop depends on the blood density ρ, *g* (gravity acceleration), and vessel branch height *h*:

$$\Delta P = \rho.g.h \qquad (6)$$

The heart's pumping of blood into the arteries creates a force *Fb* that tends to push the contrast agent away from the heart. If contrast medium is injected in the direction of the blood flow, one particle line is enough to model it in a straight vessel. If contrast is injected against the blood flow, some of the medium will pass with, and some against, blood flow direction,

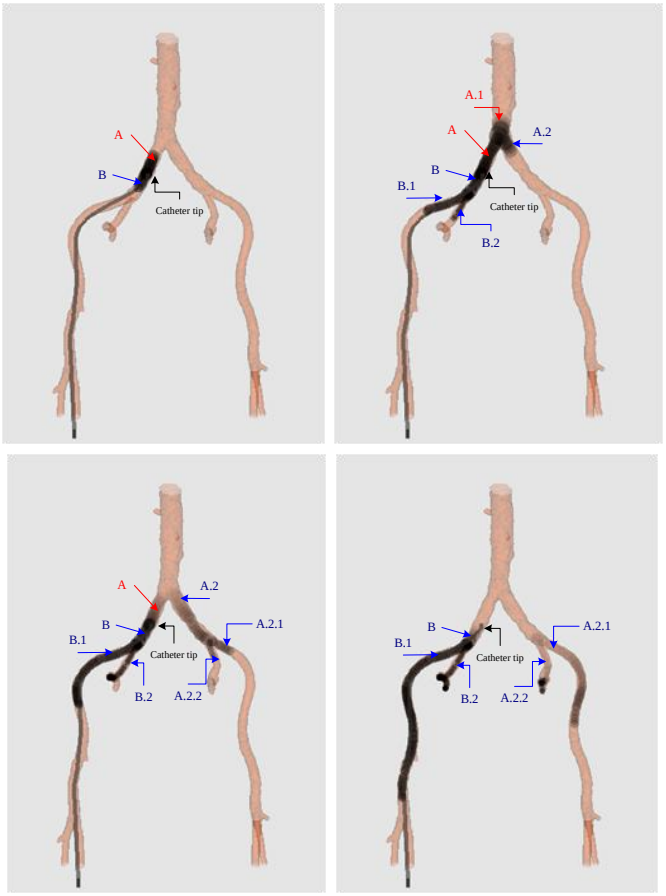
requiring the creation of a second line. The particles of the two lines are created as shown in Fig. 1, only the direction of propagation differs for each line. In each direction, the blood velocity vb_t influences the contrast agent velocity v_t at time t according to the following equation:

$$v_{t2}=v_{t1}+(vb_{t2}-vb_{t1}) \quad (7)$$

The presence of blood in the vessels also affects the concentration of contrast medium. At the initial injection time *t0*, the concentration is high, resulting in a deep black colour. However, it will gradually mix with the blood according to blood velocity *vb*. The velocity of the initial injection *vc* also slows down the mixing effect. In addition, if a large quantity of contrast medium *q* is injected, it will take longer time to mix than a small quantity. Therefore, each particle's colour *c* gradually reduces at each *dt* according to the following equation:

$$c_{t+dt}=c_t-k1.vb+k2.q+k3.vc \quad (8)$$

where *k1* is the blood mixing constant and *k2* and *k3* are constant values.



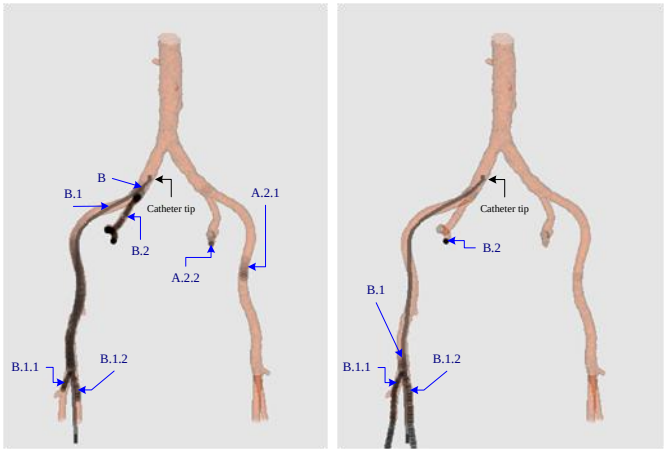


Fig. 3. Contrast agent injection in the right iliac artery and its propagation in the vasculature.

E. Bifurcation transition

Based on the model described in the previous sections, the contrast medium can propagate along a vascular branch. However, in a complex vascular system, it is necessary to deal with bifurcations to determine how the contrast medium spreads. This transition is based on the pre-computed centreline. When the particle meets a bifurcation, the current particle line is stopped and a new particle line is created for each bifurcating branch. New lines are created following the process described in Fig. 1 with the two first particles X0 and X1 replaced by the bifurcation point and the first control point of the new branch, respectively.

III. TESTS AND EVALUATION

Tests performed by interventional radiologists using 2 datasets available in the simulator showed properties of our virtual catheter were close to those of a real instrument. Fig. 3 illustrates contrast agent injection through the virtual catheter in a patient vascular model. The patient heart beat is set to 60 bpm, injection speed is 2.5m/s and injection volume is 3ml. Top left: contrast medium is initially injected in the right iliac artery. Top right: A meets the aorta bifurcation and splits in two directions: retrogradely in the aorta, and antegradely in the left iliac artery. B meets the iliac bifurcation, producing two fluid lines. Middle left: A.1 completely mixes with the blood, while A.2 mixes produces A.2.1 and A.2.2 in the external and internal left iliac arteries. B.1 and B.2 go on antegradely. Middle right: A and A.2 completely mix with the blood. The part of A which did not mix yet and which was pushed down towards the legs by the heart beat joins B and bifurcates into B.1 and B.2. Bottom left: B.1 meets a bifurcation and becomes B.1.1 and B.1.2 in the internal and external femoral arteries. Bottom right: B, A.2.2 and A.2.1 mix completely with the blood. The residual contrast medium in B.1.1 and B1.2 will pass antegradely and mix below the vessel end.

Table 1 describes the performance of the contrast medium simulation. There are two patient datasets: RLH002 and RLH007 with 54500 and 219904 surface triangles, respectively. CAP (catheter particles) corresponds to the length of the catheter in terms of particle number. Position describes the branch in which the injection happened. IV (injection

volume), IS (injection speed), BFS (blood flow speed) and HB (heart beats) are chosen based on experience. Finally, CMP (contrast medium particles) and Average FPS (Frames per second) show the rendering speed. The tests were performed on a laptop with an Intel Core 2 Duo CPU T1800 at 2.10GHZ and 2GB RAM. The average FPS for the 10 tests is 20.8 with standard deviation of 6.3. The contrast medium propagation simulations for the above two patients were shown to interventional radiologists: overall, feedback indicated that the simulations are very realistic and correspond to the propagation process.

IV. Discussion and Conclusions

This paper proposes an innovative contrast agent simulation and its interactions with blood which, to the best of our knowledge, represents the first full 3D model of contrast agent propagation. Contrast medium is modelled by particle lines where each particle has an adaptive radius to fit the vessel cross section. It uses prior information from a centreline representation of the vasculature. The contrast agent propagation depends on three forces: the initial injection force, the constraint from the vessel walls, and the blood flow. It leads to a realistic simulation with varying contrast medium speeds and fading of the contrast medium while mixing with the blood. Overall, with an average frame rate of 20.8 frames per second for the two datasets, the contrast agent injection is realistic, fairly close to real time and could help angiographic training.

Future work will focus on improving the speed and physical modelling of the simulation. The speed of the simulation could be improved by pre-processing the path of the particles or by using GPU acceleration to allow very large numbers of particles to be injected in the vasculature and keep a real time effect. The blood model could also be improved by replacing the current sine function with more complex functions able to introduce turbulences and arrhythmia. Improving the way the simulation deals with sharp radius changes is also contemplated. Finally, tests and further evaluation should be performed on more patient datasets.

Acknowledgment

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References

[1] X. Wu, J. Allard, S. Cotin, Real-Time Modeling of vascular flow for angiography simulation. MICCAI 07, Part I, LNCS 4791, pp.557-565, 2007.

[2] W. Liu, C. Sewell, N. Blevins, K. Salisbury Et al., Representing Fluid with Smoothed Particle Hydrodynamics in a Cranial Base Simulator. Stud Health Technol Inform. 132, 257-9, 2008.

Table 1. Analysis of the contrast medium simulation. CAP (catheter particles) corresponds to the length of the catheter in terms of particle number. Position describes the branch in which the injection happened. IV (injection volume), IS (injection speed), BFS (blood flow speed) and HB (heart beats) are chosen based on experience. Finally, CMP (contrast medium particles) and Average FPS (Frames per second) show the rendering speed.

Patient No	CAP	Position	IV (ml)	IS (m/s)	BFS (m/s)	HB (t/m)	CMP	Ave FPS
RLH002	19	Femoral	2	2.5	2.5	50	201	31
RLH002	67	Iliac	1	2.5	2.38	60	41	31
RLH002	92	Iliac	3	2.5	3.32	60	198	19
RLH002	118	Iliac	3	5	2.42	90	291	16
RLH002	136	Aorta	1	3.8	3.67	90	202	21
RLH007	33	Iliac	3	4	0.15	60	101	21
RLH007	77	Iliac	2.4	5	2.27	70	101	26
RLH007	127	Iliac	4.2	5	0	50	393	19
RLH007	165	Aorta	2.4	4	2.3	90	522	12
RLH007	165	Aorta	1	4	0.92	90	62	18