Numerical Simulation of Venous System Blood Flow for Hepatic Donor Patient and Portal Hypertension's Proposed Measurement

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1 Introduction

Concerning hepatic blood flow, clinicians require reproducible and non-invasive methods. In clinical practice, we quote two useful non-invasive methods in hepatic flow measurement despite their known limitations such as Doppler ultrasound (US) and Phase-contrast magnetic resonance imaging (PC-MRI). However, results in the literature suggest that these methods may in some cases show diverse analysis. Furthermore, in velocity measured by US doppler, there is great errors' percentage that may reach 20% in reference to blood velocity estimation [1]. In this case, a comparative study was done in order to set the variability of parameters using these methods [2, 3].

Nowadays, Numeric simulation of blood circulation opens up predictions and perspectives for doctors in terms of performing a surgery. Virtual simulation provides useful biomechanical models for clinicians in different blood flow systems. However, like any technic or method, the virtual simulation will be used with well-defined conditions in order to be verified in terms of capability and performance.

Little work has been done to study blood flow in the portal vein. The work reported in the literature has mentioned the hemodynamic simulation of blood flow in different organs such as the cerebral arterial circle [4], the carotids [5]

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and cardiovascular system [6] etc. The numerical simulation of venous system was rarely approached due to its complex geometry as well as its difference of elasticity with respect to the arterial system.

The purposes of this research will focus mainly on studying the normal venous blood circulation in the liver of a female donor through the determination of blood's pressure and velocity by the chosen hypothesis and condition, validating results, understanding the phenomenon of portal hypertension [7] that can be seen in several pathologies such as cirrhosis or thrombosis, as well as proposing portal pressure's measure related to a young receiver patient before and after chirurgical intervention.

2 Methods

We propose to simulate the 3D blood flow of the venous system based on 2D database. Our approach is as follows: We segmented series of anatomical images about the venous system, along with a 3D reconstruction extracting iso-surfaces. Subsequently, from the generated 3D triangular mesh, we modelled the volume of our structure by the "Reverse Engineering" method. Finally, we prepared a CFD study Fig. 1.

2.1 Segmentation and 3D Reconstruction

The anatomical structure was obtained from native cuts using a thoraco-abdominopelvic CT scan of liver donor patient before the chirurgical inter-

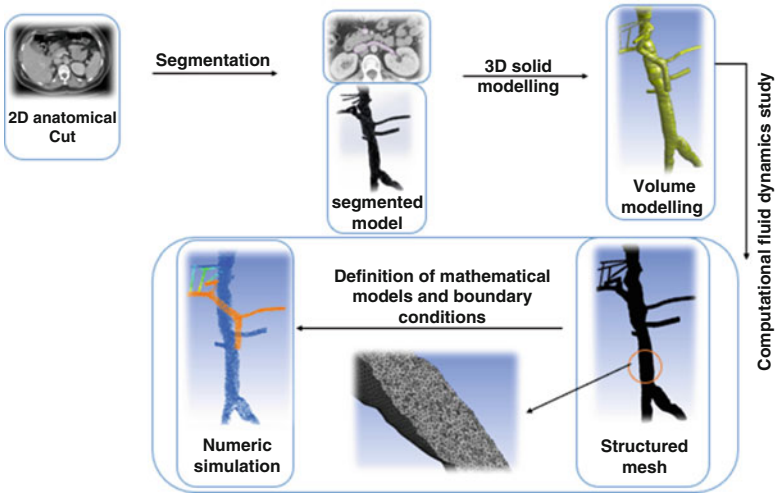

Fig. 1 Methods and approach

vention. We identified different veins and segmented the 2D anatomical cuts (0.7 × 0.7 × 1.25 mm) after injection of the contrast medium. Indeed, for each anatomical section we surround the concerned veins and control the appearance of the following line using AMIRA software. The segmentation has been validated by radiologist experts.

2.2 3D Volume Modelling

the next step is to prepare a volume model through the Reverse Engineering method which used to generate a three-dimensional digital model from 2D data [8]. This method consists of designing a series of parallel curves relating to the generated 3D triangular mesh and connecting them with surface sections in order to obtain a closed surface. It is noteworthy to mention; this surface model has been transformed into a volume model by filling the material.

2.3 CFD Study

In this section, we performed a CFD study consisting of the following steps:

2.3.1 Finite Elements Mesh

In CFD analysis, the flow domain is subdivided into smaller subdomains. For this geometry, a structured mesh has been conducted in which domain has been discretized into purely tetrahedral elements.

This figure depicts a mesh quality study about the size and number of geometric primitives which forms the tetrahedral mesh after choosing one of these five meshes composed of 2,138,653 elements from which the pressure is almost constant. This volume mesh allowed us to have a reliable result along with optimizing the calculation time Fig. 2.

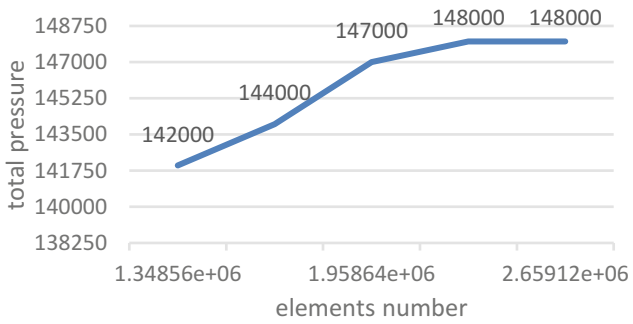

Fig. 2 Pressure variation according to the number of elements

2.3.2 Parameter Definition: Mathematical Model and Boundary Conditions

The flow is assumed to be isothermal so the energy equations have been neglected, while the Navier Stokes equations of continuity and momentum equations 1 and 2 have been solved numerically under stable conditions and for an incompressible fluid.

$$\frac{\partial \rho}{\partial t} + div \rho \vec{V} = 0 \tag{1}$$

$$\rho \frac{\partial \vec{V}}{\partial t} + \bar{=}grad \vec{V}.\rho \vec{V} = \rho \vec{F} - \vec{grad}\, p + \mu \left(\Delta \vec{V} + \vec{grad}\, div \vec{V} \right) \tag{2}$$

V: Vector fluid velocity, p: momentum, F: External forces exerted on the fluid, ρ: density of the fluid.

We use the hypothesis of a perfectly rigid wall. Blood is a non-Newtonian fluid its viscosity depends on mechanical shear stress. Several models can describe this behaviour between mechanical stress and deformation. Recently, comparative research has been conducted in order to identify the most useful model within blood circulation. Power Law model was recommended which defined by the following Eq. (3) and parameters [9].

$$\mu = \mu_0 (\dot{\gamma})^{n-1} \tag{3}$$

γ: Shear rate, n: power-law index, $\mu_0$initial Viscosity.

Blood density usually depends on the sex and age group of human beings. We took on average density equal to 1060 kg / m3. The flow is defined laminar since the Reynolds coefficient Re (4) does not exceed 1500.

$$R_e = \frac{\rho V D}{\mu} \tag{4}$$

ρ: density (kg.m-3), μ: Dynamic viscosity, V: Characteristic velocity (m / s), D: Largest vein diameter (m).

Due to the complexity and orientation of venous system's shapes, the flow must be considered as turbulent. That is why we examined both laminar and turbulent cases. Although there are numerous turbulence models, we used two turbulence models named K Epsilon and K Omega [10].

In addition, we know that the definition of boundary conditions plays an indispensable role in the numeric experience. We have affected the entry velocities of the veins that feed the liver and vena cava. In CFD study, the input velocity values are obtained from in vivo measurement and are specified in Table 1. For the output values, we used a flow ratio of 1 at the level of the right atrium.

Table 1 Boundary Conditions

Veins	Velocity	Diameter
Left iliac vein	14 (cm/s)	13 (mm)
Right iliac vein	14 (cm/s)	13 (mm)
Left renal vein	15 (cm/s)	8 (mm)
Right renal vein	15 (cm/s)	8 (mm)
Superior mesenteric vein Splenic vein	18 (cm/s) 16 (cm/s)	11 (mm) 7 (mm)

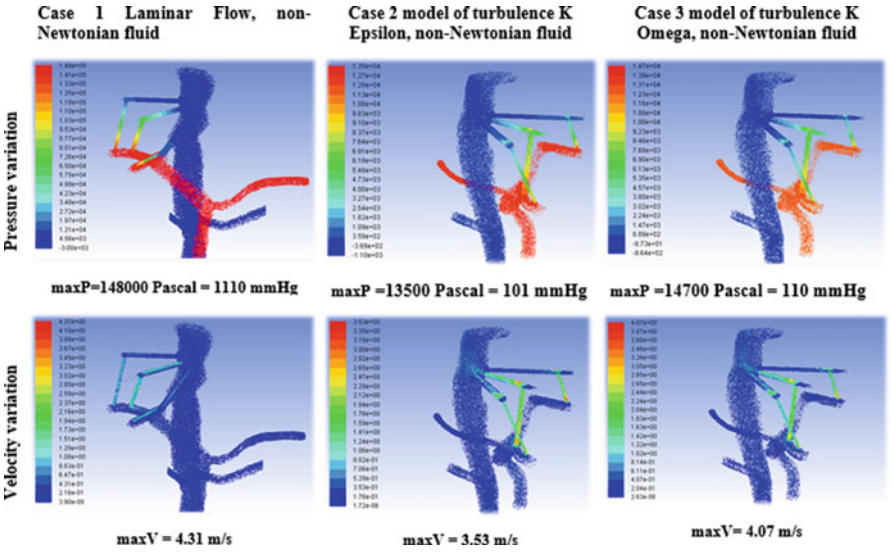

Fig. 3 Results of virtual simulation

3 Results and Conclusions

Regarding the figure below, it is concluded that the laminar model diverged from measured data. In contrast, turbulence models provide an accurate result. Indeed, virtual velocity in the portal vein is 20 cm/s, this result shows a significant correlation with the result measured by Us Doppler which is between 15 and 20 cm/s. Maximum pressure is concentrated at the liver while maximum velocity appears in capillaries situated between sus-hepatics veins and portal branches. Moreover, this research opens perspectives on studying the phenomena explaining portal hypertension through a mechanical model starting by data acquisition, segmentation, 3D reconstruction and numerical simulation. Indeed, there is a growing need for clinicians to detect portal hypertension by measuring hepatic pressure gradient using numeric simulation. Furthermore, we are going to simulate a hepatic pressure gradient for a receiver patient suffering from portal hypertension after and before chirurgical interventions, and to determinate wall shear stress in order to predict variceal bleeding risk (Fig. 3).

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