Published in final edited form as:

Chem Eng Sci. 2013 August 9; 99: 59–66. doi:10.1016/j.ces.2013.05.025.

A Porous Media Model for Blood Flow within Reticulated Foam

J.M. Ortega*

*Staff Scientist, Computational Engineering Division, Lawrence Livermore National Laboratory, Livermore, CA

Abstract

A porous media model is developed for non-Newtonian blood flow through reticulated foam at Reynolds numbers ranging from 10^{-8} to 10. This empirical model effectively divides the pressure gradient versus flow speed curve into three regimes, in which either the non-Newtonian viscous forces, the Newtonian viscous forces, or the inertial fluid forces are most prevalent. When compared to simulation data of blood flow through two reticulated foam geometries, the model adequately captures the pressure gradient within all three regimes, especially that within the Newtonian regime where blood transitions from a power-law to a constant viscosity fluid.

Keywords

Porous media; Non-newtonian fluid; Blood; Foam; Computational fluid dynamics; Biomedical engineering

1 Introduction

Reticulated foams are used in a number of biomedical applications that require the passage of blood flow through the foam structure. For example, during a balloon angioplasty or stent placement, embolic particulates can become dislodged from the stenosed artery being treated. To reduce the risk of subsequent stroke, intravascular devices have been proposed to filter these particulates from the bloodstream by means of reticulated foam [32]. Another example occurs during a cardiopulmonary bypass procedure, in which excess blood is collected from the surgical site. Before this blood can be reintroduced into the patient's artery, it must first be purified of debris, such as tissue, bone fragments, air bubbles, and thrombus. This can be achieved by passing the blood through a filter comprised of multiple reticulated foam layers, which both purify and de-foam the blood [47, 53, 58]. Similarly, transfusion blood must be filtered to eliminate micro-aggregates of platelets and fibrin that form over the course of the storage period [35, 51]. Several in-line reticulated foam filters have been manufactured in the past to accomplish this task [50].

In addition to blood filtration, reticulated foam has also been studied as an alternative to endovascular coils for the treatment of intracranial aneurysms. Metcalfe *et al.* [34] invasively inserted reticulated shape memory polymer (SMP) foam into canine model aneurysms, demonstrating that the open cellular foam structure favors an in-growth of the

© 2013 Elsevier Ltd. All rights reserved.

Corresponding author: Jason M. Ortega, Lawrence Livermore National Laboratory, P.O. Box 808, L-090, Livermore, CA 94551; ortega17@llnl.gov; phone: 925-423-3824.

Publisher's Disclaimer: This is a PDF file of an unedited manuscript that has been accepted for publication. As a service to our customers we are providing this early version of the manuscript. The manuscript will undergo copyediting, typesetting, and review of the resulting proof before it is published in its final citable form. Please note that during the production process errors may be discovered which could affect the content, and all legal disclaimers that apply to the journal pertain.

cells involved in neointima formation. For the treatment of type II endoleaks, Rhee *et al.* [39] evaluated the efficacy of a procedure in which reticulated polyurethane foam elements are released within a canine abdominal aortic aneurysm model. The procedure successfully induced thrombus formation between the aneurysm wall and an impermeable stent graft used to maintain patent flow through the aorta. An endovascular technique for delivering reticulated SMP foam to an aneurysm has also been developed by our research group [27]. During the treatment procedure, a compressed piece of SMP foam is delivered through a catheter to the aneurysm where it is heated by laser or thermal energy from a radially-diffusing fiber optic cable or electrical wires, respectively. The energy deposition releases the SMP foam from its secondary compressed shape, allowing it to expand to its primary shape and, thereby, fill the aneurysm. Following the treatment, the blood flow within the aneurysm is decelerated and the conditions for thrombus formation and subsequent healing are established. An example of this is shown in a recent animal model investigation in which a complete re-endothelialization occurs across the neck of foam-filled aneurysm by 90 days after the procedure [40].

To gather additional insight into the post-treatment hemodynamics of these aneurysms, we recently conducted a computational study of two patient-specific basilar aneurysms [36]. A prediction of the thrombus formation patterns throughout the foam-filled aneurysms is obtained by means of a low fidelity thrombosis model that is based upon the residence time and shear rate of blood. When compared to the simulation results of non-Newtonian blood flow through the SMP foam geometry, the results obtained from a porous media model using a Newtonian viscosity significantly underestimate the amount of thrombus predicted to form within the treated aneurysms. Despite this shortcoming, the porous media model has a major advantage in that it requires substantially fewer computational resources since the complex, three-dimensional foam geometry does not need to be modeled. Therefore, to maintain this computational advantage and to improve the predictive capability of these simulations, the purpose of the present study is to develop a porous media model that is applicable for non-Newtonian blood over the range of Reynolds numbers present inside an aneurysm treated with reticulated SMP foam.

2 Theoretical Background

The non-Newtonian behavior of blood arises from the suspension of red blood cells (RBCs) within the blood plasma medium [17]. These RBCs, which have a biconcave disk shape, display a range of biomechanical behaviors that depends upon the shear rate, $\dot{\gamma}$, to which they are exposed, where

$$\dot{\gamma} = \sqrt{2\mathbf{S}:\mathbf{S}} \quad (1)$$

and $\mathbf{S} = \frac{1}{2}(\nabla\mathbf{u} + \nabla\mathbf{u}^T)$ is the symmetric portion of the velocity gradient tensor, $\mathbf{u}$. At very small rates of shear, the RBCs aggregate into stacks, forming an arrangement that resists deformation. Hence, the effective viscosity of blood is relatively large at smaller shear rates. As the shear rate is increased, the stacks of RBCs are broken apart and blood becomes less resistant to shear. At even larger shear rates, hemodynamic forces align the long axes of the RBCs to the background velocity field. As more of the RBCs become oriented in the same direction, they flow more easily past one another, causing the effective viscosity to asymptotically approach a constant Newtonian value (≈ 0.0035 Pa·s). In general, the shear rate threshold for this Newtonian behavior is approximately 100 s^{-1} .

This dependence of blood viscosity upon the shear rate has been characterized and investigated with several analytical models [2, 3, 6, 19, 20, 22, 26, 52, 55]. For this particular study, we employ the generalized power-law viscosity model [2]

$$\mu = \lambda |\dot{\gamma}|^{n-1} \quad (2)$$

where λ is the power-law consistency index,

$$\lambda(\dot{\gamma}) = \mu_0 + \Delta\mu \exp \left[- \left(1 + \frac{|\dot{\gamma}|}{a} \right) \exp \left(\frac{-b}{|\dot{\gamma}|} \right) \right] \quad (3)$$

n is the power-law index,

$$n(\dot{\gamma}) = n_0 - \Delta n \exp \left[- \left(1 + \frac{|\dot{\gamma}|}{c} \right) \exp \left(\frac{-d}{|\dot{\gamma}|} \right) \right] \quad (4)$$

$n_0 = 1.0$, $\mu_0 = 0.0035$ Pa·s, $n = 0.45$, $\mu = 0.025$ Pa·s, $a = 50$, $b = 3$, $c = 50$, $d = 4$ (Fig. 1). This model, which is based upon a curve fit to the rheological data of [5, 9, 10, 11, 21, 33, 38, 49, 55], behaves as a simple power-law fluid at very small shear rates with $\mu_0 + \mu$ and $n = n_0 - n$ as 0 . At shear rates greater than ≈ 200 s⁻¹, the model approaches a constant Newtonian value of μ_0 .

While there are numerous non-Newtonian porous media models describing the pressure gradient as function of flow speed [4, 24, 46, 57], a starting point for the current study is an expression of the form of Sabiri and Comiti [41, 42, 43] and Sabiri *et al.* [44, 45], which has been applied not only to synthetic reticulated foams, but also to angular-shaped crushed rocks, irregular-shaped sand particles, and beds of packed plates and mixed-sized spheres [7, 12, 25, 54]. For fully-developed laminar flow of a simple power-law fluid with a viscosity of μ^{n-1} , Sabiri and Comiti [41] show that the pressure gradient, $\Delta p/L_s$, across a sample of length L_s is given by the expression

$$\frac{\Delta p}{L_s} = N' u_0^{n'} + M' u_0^2 \quad (5)$$

where u_0 is the superficial or Darcy velocity and N and M depend upon constant parameters, such as the fluid properties (μ , n , and the fluid density, ρ), the geometric details of the porous media (the porosity, ϵ , and the dynamic specific surface area), and the tortuosity, τ , of the flow through the porous media. The two terms on the right-hand side of Eq. 5 account for pressure losses due to non-Newtonian viscous forces and inertial forces, respectively. For a Newtonian fluid ($n = 1$, $\mu = \mu_0$), Eq. 5 has the form shown by Comiti and Renaud [12],

$$\frac{\Delta p}{L_s} = \widehat{N} u_0 + \widehat{M} u_0^2 \quad (6)$$

One significant limitation of Eq. 5 is that it assumes the power-law index and the consistency index to be constants, when, in fact for blood, these properties are both functions of the shear rate (see Fig. 1a). Due to the no-slip boundary condition, the shear rate is not a constant throughout the foam, but varies, such that blood closest to the pore walls

experiences shear rates larger than that at the pore centers. This, in turn, leads to a spatial distribution of the power-law index and the consistency index. As the flow speed and, therefore, the shear rate, increase, blood near the walls is the first to transition from a non-Newtonian to a Newtonian fluid. For even larger flow speeds, the viscosity near the wall no longer decreases in the power-law manner assumed by Eq. 5 and, thus, the foam imposes a relatively larger resistance to blood flow. Should this transition occur at a range of flow speeds below which the inertial forces become predominant, Eq. 5 will under predict the resulting pressure gradient.

This deficiency can be alleviated in an *ad hoc* manner by adding a Newtonian viscous force term, Nu_0 , to Eq. 5, thereby accounting for the Newtonian behavior exhibited by blood at these larger shear rates. The resulting expression for the pressure gradient is

$$\frac{\Delta p}{L_s} = N^* u_0^{n^*} + Nu_0 + M^* u_0^2 \quad (7)$$

where the variables N^* , N , and M^* are empirically determined by fitting Eq. 7 to the p/L_s versus u_0 data and n^* is taken to equal $n_0 - n = 0.55$, the power-law index of blood for small shear rates (see Eq. 4). Including both the non-Newtonian and Newtonian viscous force terms effectively divides the pressure gradient versus flow speed curve into three regimes: values of u_0 for which non-Newtonian viscous forces dominate at smaller shear rates, *i.e.* $N^* u_0^{n^*} > Nu_0 + M^* u_0^2$; values of u_0 for which inertial forces dominate at larger shear rates, *i.e.* $M^* u_0^2 > N^* u_0^{n^*} + Nu_0$; and a transition regime in between these two where Newtonian viscous forces dominate. Defining the Reynolds number as

$$Re_p^* = \frac{M^* u_0^2}{(N^* u_0^{n^*} + Nu_0)} \quad (8)$$

and non-dimensionalizing Eq. 7 by $M^* u_0^2$ produces

$$f_p^* = \frac{1}{Re_p^*} + 1 \quad (9)$$

where

$$f_p^* = \frac{1}{Re_p^*} + 1 \quad (10)$$

is the friction factor. For a Newtonian fluid, the Reynolds number is defined as

$$\widehat{Re}_p = \frac{\widehat{M}u_0}{\widehat{N}} \quad (11)$$

and the friction factor as

$$\widehat{f_p} = \frac{\Delta p}{L_s} \frac{1}{\widehat{Mu}_0^2} \quad (12)$$

Subsequently non-dimensionalizing Eq. 6 by $\widehat{Mu}_0^2$ gives

$$\widehat{f_p} = \frac{1}{\widehat{Re_p}} + 1 \quad (13)$$

3 Computational Setup

To assess the validity of the porous media model in Eq. 7, we simulate blood flow through the reticulated SMP foam using computational fluid dynamics (CFD). A digitized cylindrical geometry of the foam (5.4×10^{-3} m diameter $\times$ 4.8×10^{-3} m long) is acquired by means a micro-CT scanner (Skyscan 1172, Micro Photonics Inc.) at a spatial resolution of 4.15×10^{-6} m. Two levels of virtual reticulation are achieved by applying two separate intensity thresholds to the micro-CT images (Fig. 2a), thus removing the faces that exist between individual foam pores (Fig. 2b). These images are processed with Amira software (Visage Imaging, Inc.) and the resulting foam surface geometries are exported as CAD files that are analyzed (Rhino 3D, Robert McNeel & Associates) to determine their surface area and volumetric properties. For the first level of reticulation (R1, Fig. 2c), the sample porosity, surface area per unit bulk volume, and static specific surface area are 0.985, 1300 m²/m³, and $1300/(1 -) = 8.67 \times 10^4$ m²/m³, respectively. Likewise, the respective values are 0.954, 4900 m²/m³, and 1.07×10^5 m²/m³ for the second level of reticulation (R2, Fig. 2d).

A cube-shaped foam sample is digitally cut from the cylindrical CAD geometry and centered within a hexahedral-shaped computational domain, which has the dimensions $15L_s \times 1L_s \times 1L_s$, where $L_s = 3.8 \times 10^{-3}$ m is the length of one side of the sample (Fig. 3a). At the inlet to the computational domain, a steady, uniform velocity boundary condition is specified and the pressure is extrapolated from the inside of the domain. The inlet velocity, u_0 , ranges from 1×10^{-7} to 1.2 m/s, the upper bound of which is slightly greater than the maximum fluid velocity in a healthy basilar artery (BA) [23]. A slip velocity boundary condition is specified on the four walls and the computed variables are extrapolated at the outlet face of the computational domain. On the surface of the foam, a no-slip velocity boundary condition is specified and the foam is taken to be rigid. Since the foam geometry may be anisotropic, it is oriented along each of the three coordinate directions in separate simulations, thus allowing the pressure gradient to be measured in each of these directions. Blood is modeled as a continuum since the red blood cell diameter ($\approx 8 \times 10^{-6}$ m) is approximately 0.8% of the nominal foam pore size [17]. The flow field is solved using the conservation of mass

$$\nabla \cdot \mathbf{u} = 0 \quad (14)$$

and momentum

$$\rho \frac{D\mathbf{u}}{Dt} = -\nabla p + \mu \nabla^2 \mathbf{u} \quad (15)$$

equations for an incompressible flow using a finite-volume code (STAR-CCM+ v. 7.04.006, CD-Adapco), where ρ is taken to equal 1060 kg/m³ for blood. The code solver uses a co-

located variable arrangement and a Rhie-and-Chow-type pressure-velocity coupling combined with a SIMPLE-type algorithm [13, 14, 18, 29, 30, 37]. The convective fluxes in Eqs. 14–15 are evaluated with a second-order upwind scheme. The computational domain is discretized onto an unstructured, hexahedral cell mesh, which is comprised of a Cartesian core mesh and several layers of hexahedral prismatic cells extruded from the foam surface. Between these two meshes are transitional, polyhedral cells. For inlet velocities that require the solution of an unsteady flow field in the foam, the time derivative in Eq. 15 is discretized using a second-order implicit scheme with a time step of 1×10^{-3} s. Both the Newtonian ($\mu = \mu_0$) and non-Newtonian viscosity models (Eq. 2) are utilized for blood. The simulations are computed on parallel processing machines using up to several hundred 2.8 GHz Intel CPUs. To evaluate the sensitivity of the finite-volume solver to the mesh refinement level, the solutions for the R1 reticulation with the non-Newtonian viscosity model are repeated on three mesh resolutions, which have sizes of 27×10^6 (mesh 0), 107×10^6 (mesh 1), and 263×10^6 (mesh 2) cells.

4 Results

A least-squares fit of Eqs. 5–7 is subsequently performed on the respective simulated pressure gradients for each coordinate direction. The resulting coefficients for Eq. 7 demonstrate a mesh convergent trend (Fig. 3b–c) and, hence, the spatial resolution corresponding to mesh 1 (107×10^6 cells) is subsequently utilized for both reticulation levels. These simulations are performed until the running average of the pressure measurements changes less than 0.02%. The resulting pressure gradient in, for example, the x -direction is plotted as a function of u_0 in Fig. 4. Within the range of u_0 where Newtonian viscous forces are dominant, Eq. 7 represents the data better than Eq. 5 ($n = 0.55$), which under predicts the pressure gradient as anticipated. For the R1 and R2 reticulation levels, this range of u_0 lies between 1×10^{-2} to 0.4 m/s and 2×10^{-3} to 0.1 m/s, respectively. At the lower end of this Newtonian range, the non-Newtonian viscous forces, which are nevertheless present within the foam, have a spatial distribution that is dependent upon the level of reticulation. The non-Newtonian forces ($n < 1$ in Fig. 5a) for the R1 reticulation reside within the pore centers, while the Newtonian viscous forces cover the foam surface. The R2 reticulation has a larger distribution of n , such that a few dead-end pores are almost entirely filled with quiescent, non-Newtonian blood (Fig. 5b). As u_0 is subsequently increased to the upper end of the Newtonian viscous force range, the non-Newtonian viscous forces all but vanish as the flow field becomes characterized by tubules of relatively larger shear flow that wind through adjacent pores (Fig. 5c–d).

Further insight can be obtained by considering the fitting coefficients shown in Table 1 for each of the coordinate directions and reticulation levels. A comparison of the relative values in each direction demonstrates a transverse anisotropy of the foam in the z -direction for both reticulation levels. This anisotropy is produced by the foam forming process [48], whereby the foam is expanded vertically in the z -direction against gravity and is confined in the x - and y -directions by the walls of a cylindrical container. Consequently, the pores are relatively smaller in the x - y plane and present a greater flow resistance to blood passing through the foam in the z -direction, especially within the inertial regime as indicated by the larger values of M^* .

This transverse anisotropy also manifests itself in the flow tortuosity, which is assessed by integrating an array of 256×256 velocity streamlines forward in time from the inlet of the computational domain through the SMP foam by means of a fourth-order Runge-Kutta scheme (Fig. 6). The tortuosity, $\tau = L_a/L_s$, is typically largest in the z -direction, where L_a is the average fluid path length within the SMP foam (Table 2). It is also evident that τ is not a constant parameter, as assumed within the porous media models of Comiti and Renaud [12],

Sabiri and Comiti [41, 42, 43], and Sabiri *et al.* [44, 45], but rather increases quite substantially with u_0 in the Newtonian viscous and inertial regimes (Fig. 7). Moreover, exhibits sensitivity to the viscosity model at smaller flow speeds, which is especially true for the R2 reticulation. This variability in Δp is consistent with the discussion of Dharmadhikari and Hale [15], who postulated that the tortuosity depends upon the flow speed and that a shear-thinning, power-law fluid will have a tendency to follow a shorter path through the porous media than a Newtonian fluid. Similar thoughts have been presented by several others [8, 16, 31].

Using the terms in Table 1, the Reynolds number and friction factor are computed for the non-Newtonian (Eqs. 8 and 10) and Newtonian (Eqs. 11–12, $n = 1$, $\mu = \mu_0$) viscosity simulation results. For Reynolds numbers ranging from 10^{-8} to 10, the (f_p^*, Re_p^*) and (f_p, Re_p) data collapse onto Eqs. 9 and 13, respectively, for both viscosity models and foam reticulation levels (Fig. 8). In addition, the Newtonian viscous forces are prevalent within the foam for Reynolds numbers ranging between 10^{-2} to 1.

5 Discussion

The porous media model given by Eq. 7 marks an improvement for predicting the pressure gradient of blood flow through reticulated foam. In a previous study [36], we utilize Eq. 6 to simulate the post-treatment hemodynamics of an aneurysm filled with the R1 reticulation of the SMP foam. Near the aneurysm dome where blood travels very slowly and non-Newtonian viscous forces are prevalent, significant differences appear between the results of the Newtonian viscosity porous media model (Eq. 6) and those of non-Newtonian blood flow (Eq. 2) through the actual foam geometry. The data shown in Fig. 4a explain why: Eq. 6 yields a much larger value of u_0 for a given pressure gradient, allowing blood to travel more easily through the foam-filled aneurysm. Implementing Eq. 7 will not only alleviate this shortcoming, but also provide an improvement over Eq. 5 closer to the aneurysm neck where Newtonian viscous forces are prevalent and Eq. 5 under predicts the pressure gradient.

In spite of these advantages, there are several underlying assumptions made in the development of this porous media model. First, the generalized power-law viscosity model (Eq. 2) does not reproduce the yield stress of blood, an effect which arises from RBC aggregation at very low shear [17]. This could potentially impact the pressure gradient within the foam at the smallest flow speeds. To account for this yield stress, the Casson model [17] can be utilized, though the viscosity predicted by this model for large shear rates is slightly larger than the Newtonian value of 0.0035 Pa·s [22]. Another assumption is the rigid foam boundary condition, which is qualitatively supported by a previous study on the *in vitro* foam treatment of an aneurysm [27]. Within that experiment, the maximum flow speed impinging upon the foam is comparable to that in the present study and no large-scale foam deformation is observed. However, small-scale, local deformation of the pore struts and faces may be occurring and thus altering the pressure gradient relative to that of a completely rigid foam structure. The impact of such deformation on the results of the present study can be assessed in subsequent simulations by considering the material properties of the foam and coupling the mechanical foam-blood interactions. Lastly, the most significant assumption of this study is that the simulations neglect the foreign body response to the foam. Following exposure to blood, numerous biochemical reactions are initiated that lead to the adhesion of fibrinogen, proteins, and red blood cells to the foam surface [1, 28, 56]. For example, in a previous *in vivo* study [28], scanning electron microscope images show red blood cell adhesion to the foam approximately 30 minutes following implantation. As these bio-materials accumulate, the pressure gradient at a given

flow speed will steadily increase over time as the foam becomes less permeable to blood flow.

6 Conclusions

Through this computational investigation, we have presented an empirical porous media model for non-Newtonian blood flow through reticulated foam. By including terms for both the Newtonian and non-Newtonian viscous forces, this model adequately represents the pressure gradient for Reynolds numbers ranging from approximately 10^{-8} to 10 and provides an improved prediction within the Newtonian regime where blood transitions from a power-law to a constant viscosity fluid. The porous media model has an advantage of requiring substantially fewer computational resources than simulations that incorporate the actual foam geometry. This, in turn, allows for the efficient evaluation of novel SMP foam devices used to treat aneurysms. The approach described by this study can be readily extended to endovascular coil geometries. Therefore in future studies, we will have the opportunity to make measurements of blood flow through both the foam and coils, thereby allowing us to further assess the underlying assumptions of this porous media model, as well as differences between these two treatment devices.

Acknowledgments

The authors thank R. Cook, W. Small, and T.S. Wilson of Lawrence Livermore National Laboratory and D.J. Maitland and J.N. Rodriguez of Texas A&M University for their assistance in this study. This work was supported by the National Institutes of Health/National Institute of Biomedical Imaging and Bioengineering Grant R01EB000462 and partially performed under the auspices of the U.S. Department of Energy by Lawrence Livermore National Laboratory under Contract DE-AC52-07NA27344. LLNL-JRNL-617022.

References

1. Anderson JM. Biological responses to materials. *Annu Rev Mater Res*. 2001; 31:81–110.
2. Ballyk PD, et al. Simulation of non-Newtonian blood flow in an end-to-end anastomosis. *Biorheology*. 1994; 31(5):565–586. [PubMed: 7833458]
3. Banerjee, RK., et al. Proceedings of the ASME Winter Annual Meeting, BED. Vol. 20. Atlanta: Dec 4–8. 1991 Effect of the non-Newtonian viscosity of blood on steady and pulsatile flow in stenosed arteries; p. 103-106.
4. Chhabra, RP. Bubbles, Drops, and Particles in Non-Newtonian Fluids. Taylor & Francis Group, LLC; Boca Raton: 2007.
5. Chien S, et al. Effects of hematocrit and plasma proteins on human blood rheology at low shear rates. *J Appl Physiol*. 1966; 21:81–87. [PubMed: 5903948]
6. Cho YI, Kensey KR. Effects of the non-Newtonian viscosity of blood on flows in a diseased arterial vessel. Part 1: Steady flows. *Biorheology*. 1991; 28:241–262. [PubMed: 1932716]
7. Ciceron D, et al. Pressure drops for purely viscous non-Newtonian fluid flow through beds packed with mixed size spheres. *Chem Eng Commun*. 2002; 189:1403–1414.
8. Ciceron D, et al. Non-Newtonian fluidisation of spherical particles. *Chem Eng Sci*. 2002; 57:3225–3234.
9. Cokelet G, et al. The rheology of human blood—measurement near and at zero shear rate. *Trans Soc Rheol*. 1963; 7:303–317.
10. Cokelet, G. Biomechanics: Its Foundation and Objectives. Prentice-Hall; New Jersey: 1973.
11. Cokelet, G. The rheology and tube flow of blood. In: Skalak, R.; Chien, S., editors. *Handbook of Bioengineering*. McGraw-Hill; New York: 1987.
12. Comiti J, Renaud M. A new model for determining mean structure parameters of fixed beds from pressure drop measurements: application to beds packed with parallelepipedal particles. *Chem Eng Sci*. 1989; 44(7):1539–1545.

13. Demirdzic I, et al. A collocated finite volume method for predicting flows at all speeds. *Int J Numer Meth Fl*. 1993; 16:1029–1050.
14. Demirdzic I, Musaferija S. Numerical method for coupled fluid flow, heat transfer and stress analysis using unstructured moving meshes with cells of arbitrary topology. *Comput Methods Appl Mech Eng*. 1995; 125:235–255.
15. Dharmadhikari RV, Kale DD. Flow of non-Newtonian fluids through porous media. *Chem Eng Sci*. 1985; 40(3):527–529.
16. Epstein N. Letter to the editor. *Chem Eng Sci*. 1998; 53:1469.
17. Ethier, CR.; Simmons, CA. *Introductory Biomechanics: From Cells to Organisms*. Cambridge University Press; New York: 2008.
18. Ferziger, JH.; Peric, M. *Computational Methods for Fluid Dynamics*. 3. Springer-Verlag; Berlin: 2002.
19. Fisher C, Stroud Rossmann J. Effect of non-Newtonian behavior on hemodynamics of cerebral aneurysms. *J Biomech Eng-T ASME*. 2009; 131:091004.
20. Fung, YC. *Biomechanics: Mechanical Properties of Living Tissues*. 2. Springer; Berlin: 1993.
21. Huang C, et al. Rheogoniometric studies of whole human blood at shear rates down to 0.0009 sec⁻¹. Part II. Mathematical interpretation. *Biorheology*. 1973; 10:17–22. [PubMed: 4724172]
22. Johnston BM, et al. Non-Newtonian blood flow in human right coronary arteries: steady state simulations. *J Biomech*. 2004; 37:709–720. [PubMed: 15047000]
23. Kato T, et al. Contrast-enhanced 2D cine phase MR angiography for measurement of basilar artery blood flow in posterior circulation ischemia. *Am J Neuroradiol*. 2002; 23:1346–1351. [PubMed: 12223376]
24. Kembrowski Z, Michniewicz M. A new look at the laminar flow of power-law fluids through granular beds. *Rheol Acta*. 1979; 18:730–739.
25. Legrand J. Revisited analysis of pressure drop in flow through crushed rocks. *J Hyd Eng ASCE*. 2002; 128:1027–1031.
26. Liepsch D. An introduction to biofluid mechanics—basic models and applications. *J Biomech*. 2002; 35:415–43. [PubMed: 11934411]
27. Maitland DJ, et al. Prototype laser-activated shape memory polymer foam device for embolic treatment of aneurysms. *J Biomed Opt*. 2007; 12(3):030504. [PubMed: 17614707]
28. Maitland, DJ., et al. Design and realization of biomedical devices based on shape memory polymers. In: Lendlein, A.; Shastri, P., editors. *Proceedings of the Material Research Society Symposium*. Vol. 1190. 2009. p. NN06-01
29. Mathur SR, Murthy JY. Pressure-based method for unstructured meshes. *Numer Heat Tr B-Fund*. 1997; 31(2):195–214.
30. Mathur SR, Murthy JY. Pressure boundary conditions for incompressible flow using unstructured meshes. *Numer Heat Tr B-Fund*. 1997; 32(3):283–298.
31. Mauret E, Renaud M. Transport phenomena in multi-particle systems—1. Limits of applicability of capillary model in high voidage beds—application to fixed beds of fibers and fluidized beds of spheres. *Chem Eng Sci*. 1997; 52:1807–1817.
32. McInnes, PR. Embolic Filter. United States Patent 6,558,405 BI. May 6. 2003
33. Merrill E, et al. Rheology of human blood, near and at zero flow: Effects of temperature and hematocrit level. *Biophys J*. 1963; 3:199–213. [PubMed: 13935042]
34. Metcalfe A, et al. Cold hibernated elastic memory foams for endovascular interventions. *Biomaterials*. 2003; 24:491–497. [PubMed: 12423604]
35. Mosely RV, Doty DB. Changes in the filtration characteristics of stored blood. *Ann Surg*. 1970; 171:329. [PubMed: 5416575]
36. Ortega JM, et al. Virtual treatment of basilar aneurysms using shape memory polymer foam. *Ann Biomed Eng*. 2013; 41(4):725–743. [PubMed: 23329002]
37. Peric M, et al. Comparison of finite-volume numerical methods with staggered and colocated grids. *Comput Fluids*. 1988; 16(4):389–403.
38. Rand P, et al. Viscosity of normal human blood under normothermic and hypothermic conditions. *J Appl Physiol*. 1964; 19:117–122. [PubMed: 14104265]

39. Rhee JY, et al. Treatment of type II endoleaks with a novel polyurethane thrombogenic foam: induction of endoleak thrombosis and elimination of intra-aneurysmal pressure in the canine model. *J Vasc Surg.* 2005; 42(2):321–328. [PubMed: 16102634]
40. Rodriguez JN, et al. Opacification of shape memory polymer foam designed for treatment of intracranial aneurysms. *Ann Biomed Eng.* 2012; 40(4):883–897. [PubMed: 22101759]
41. Sabiri NE, Comiti J. Pressure drop in non-newtonian purely viscous fluid flow through porous media. *Chem Eng Sci.* 1995; 50(7):1193–1201.
42. Sabiri NE, Comiti J. Experimental validation of a model allowing pressure gradient determination for non-Newtonian purely viscous fluid-flow through packed beds. *Chem Eng Sci.* 1997; 52(20): 3589–3592.
43. Sabiri NE, Comiti J. Newtonian and non-Newtonian fluid flows through fixed beds of plate-shared particles. *Can J Chem Eng.* 1997; 75(6):1030–1037.
44. Sabiri NE, et al. Pressure drops of non-Newtonian purely viscous fluid flow through synthetic foams. *Chem Eng Comm.* 1996; 156:59–74.
45. Sabiri, N-E., et al. Fluidization of various shaped particles by non-Newtonian purely viscous fluids. 6th World Cong. on Chem. Eng; San Diego, CA. 1996. p. 225
46. Savins JG. Non-Newtonian flow through porous media. *Ind Eng Chem.* 1969; 61(10):1847.
47. Servas, FM., et al. Blood filter. United States Patent 4,743,371. May 10. 1988
48. Singhal P, et al. Ultra low density and highly crosslinked biocompatible shape memory polyurethane foams. *J Polym Sci Part B: Polym Phys.* 2012; 50:724–737.
49. Skalak R, et al. Mechanics of blood flow. *J Biomech Eng.* 1981; 103:102–115. [PubMed: 7024641]
50. Soeter JR, et al. Comparison of filtering efficiency of four new in-line blood transfusion filters. *Ann Surg.* 1975; 181:114–117. [PubMed: 1119858]
51. Solis RT, Gibbs MB. Filtration of microaggregates in stored blood. *Transfusion.* 1972; 4:245. [PubMed: 5039459]
52. Soulis JV, et al. Non-Newtonian models for molecular viscosity and wall shear stress in a 3D reconstructed human left coronary artery. *Med Eng Phys.* 2008; 30:9–19. [PubMed: 17412633]
53. Thor, EJ., et al. Combined cardiotomy and venous blood reservoir. United States Patent 5,411,705. May 2. 1995
54. Wahyudi I, et al. Darcy and post Darcy flows within different sands. *J Hyd Res.* 2002; 40:519–525.
55. Walburn FJ, Schneck DJ. A constitutive equation for whole human blood. *Biorheology.* 1976; 13:201–210. [PubMed: 953256]
56. Ward WK. A review of the foreign-body response to subcutaneously-implanted devices: the role of macrophages and cytokines in biofouling and fibrosis. *J Diabetes Sci Technol.* 2008; 2(5):768–777. [PubMed: 19885259]
57. Wu, Y-S.; Pruess, K. Flow of non-Newtonian fluids in porous media. In: Corapcioglu, MY., editor. *Advances in Porous Media.* Elsevier; New York: 1996.
58. Zupkas, PF., et al. Unitary venous return reservoir with cardiotomy filter. United States Patent 4,642,089. Feb 10. 1987

Figure 1.
 (a) Power-law index, n , and consistency, λ , for the (b) generalized power-law viscosity model of blood (Eq. 2).

Figure 2.

(a) Raw and (b) processed micro-CT image (zoomed and cropped to show details) of a SMP foam sample. CAD geometries of the (c) R1 and (d) R2 reticulation levels of the foam sample.

Figure 3. (a) Computational domain for the simulations. (b–c) Mesh sensitivity of the fitting coefficients (flow in the x -direction) in Eq. 7 for the R1 reticulation of the foam.

Figure 4.

Pressure gradient as a function of flow speed in the x -direction for the (a) R1 and (b) R2 reticulation levels of the foam.

Figure 5. Power-law index (Eq. 4) for the non-Newtonian viscosity model (Eq. 2) of blood within the (a) R1 ($u_0 = 0.01$ m/s) and (b) R2 ($u_0 = 0.003$ m/s) reticulation levels. Iso-surfaces of velocity magnitude for the (c) R1 ($u_0 = 0.3$ m/s, iso-surface of $|\mathbf{u}| = 0.45$ m/s) and (d) R2 ($u_0 = 0.1$ m/s, iso-surface of $|\mathbf{u}| = 0.3$ m/s) reticulation levels of the foam.

Figure 6.

Velocity streamlines and tortuosity, τ , for non-Newtonian blood (Eq. 2) flow in the z -direction through the (a–b) R1 and (c–d) R2 reticulation levels of the foam. For clarity, only 16 streamlines are visualized for each case. The vertical dashed lines denote the extent of the foam.

Figure 7. Tortuosity as a function of flow speed for blood flow through the R2 reticulation level of the foam (x - and z -directions). For clarity, the simulation results for the y -direction are not shown.

Figure 8.

Friction factor as a function of Reynolds number for both foam reticulation levels and viscosity models of blood.

Variables and coefficients computed from the curve fits of Eqs. 6 and 7 to the simulated (u_0 , pL_s) data for the Newtonian ($\mu = \mu_0$) and non-Newtonian (Eq. 2) viscosity models, respectively, of blood.

R1 reticulation						
	N (N·s/m ⁴)	M (N·s ² /m ⁵)	n^*	N^* (N·s ^{n^*} /m ^{3+n^*})	N (N·s/m ⁴)	M^* (N·s ² /m ⁵)
x	1.41×10^5	3.11×10^5	0.55	1.43×10^4	9.76×10^4	3.42×10^5
y	1.46×10^5	3.23×10^5	0.55	1.46×10^4	1.02×10^5	3.53×10^5
z	1.40×10^5	4.85×10^5	0.55	1.63×10^4	9.85×10^4	5.11×10^5

R2 reticulation						
	N (N·s/m ⁴)	M (N·s ² /m ⁵)	n^*	N^* (N·s ^{n^*} /m ^{3+n^*})	N (N·s/m ⁴)	M^* (N·s ² /m ⁵)
x	2.01×10^6	1.49×10^7	0.55	1.28×10^5	1.66×10^6	1.51×10^7
y	2.45×10^6	1.80×10^7	0.55	1.33×10^5	1.92×10^6	1.85×10^7
z	4.25×10^6	4.47×10^7	0.55	1.96×10^5	3.49×10^6	4.53×10^7

Table 2

Tortuosity from the simulation data for both reticulation levels and viscosity models.

R1 reticulation		
	($\mu = \mu_0$)	($\mu = \text{Eq. 2}$)
x	1.05–1.07	1.05–1.07
y	1.05–1.07	1.05–1.07
z	1.06–1.08	1.06–1.07

R2 reticulation		
	($\mu = \mu_0$)	($\mu = \text{Eq. 2}$)
x	1.11–1.28	1.09–1.28
y	1.10–1.27	1.09–1.28
z	1.22–1.54	1.17–1.53