



Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Numerical modelling of blood clot extraction by aspiration thrombectomy. Evaluation of aspiration catheter geometry

Carlos Talayero^{a,*}, Gregorio Romero^a, Gillian Pearce^b, Julian Wong^c

^a Department of Mechanical Engineering, Universidad Politécnica de Madrid, Madrid, Spain

^b Department of Mechanical Engineering, School of Engineering, University of Birmingham, Birmingham, United Kingdom

^c Department of Cardiac, Thoracic & Vascular Surgery, National University Heart Centre Singapore, Singapore



ARTICLE INFO

Article history:

Accepted 29 July 2019

Keywords:

Blood clot removal
Aspiration thrombectomy device
Stroke
Bond-graph
Multi-body

ABSTRACT

Aspiration thrombectomy is one of the most effective systems for blood clot removal and vessel recanalization. We present the results of a study involving the modelling and extraction of blood clots in the arteries of the human body using the following computer tools: Bond-Graph methodology for the fluid domain and Multi-Body Simulation for the mechanical domain. The modelling for the mechanical domain focuses on the clot and the distal end section of an aspiration device. Our final model considers an elastic characterization of the blood clot with progressive detachment from the vessel wall. We conclude that the results of such modelling could potentially improve the effectiveness of blood clot removal by reducing the risk of clot fragmentation. Such modelling could also potentially provide an adjunct technique in improving recanalization of arteries over a range of given parameters (mechanical properties of the vessel, mechanical properties of the blood clot, blood clot length, suction pressure, catheter – clot distance, catheter shape, catheter diameter and vessel occlusion).

© 2019 Elsevier Ltd. All rights reserved.

1. Introduction

Stroke is a common cause of death (Scarborough et al., 2009). Strokes arise in the brain when the blood supply to a given area is interrupted. Several devices have been developed and implemented to remove clots that arise during stroke (Fanous and Siddiqui, 2016; Spiotta et al., 2015). Stentrievors, classified as mechanical thrombectomy devices (MTDs), are a recent class of devices that work by recanalizing cerebral vessels (Mokin et al., 2016). However, such devices may in some instances have associated potential risks which could include breakage of moving parts and distal embolization, due to thrombus fragmentation. That is one reason why they are sometimes combined with aspiration devices (Mokin et al., 2015). Aspiration devices alone can also achieve high recanalization rates (Lapergue et al., 2016). Entry through the femoral artery is the standard method used to introduce catheters which then pass into the vessels of the heart or brain. This site is chosen because it is the safest method of approach and there are fewer vital structures in proximity that

could be damaged during insertion, compared to other areas of the body. An aspiration device called the GPTAD (GP Thrombus Aspiration Device) attempts to overcome some of the potential problems associated with clot aspiration (Pearce et al., 2009; Pearce and Perkinson, 2006; Pearce et al., 2010). The device can potentially reduce the risk of distal embolization because it does not have to make contact with the clot, during clot removal. In addition, it has no moving parts, and this potentially reduces the risk of breakage that may occur in the device. In-vitro studies by Tennucci et al. (Tennucci et al., 2011) have also shown that it potentially reduces the risk of clot fragmentation.

Preliminary studies (Talayero et al., 2018; Romero et al., 2011, 2013a, 2013b) have dealt with the formation, composition and shape of clots to find some general parameters that can reliably define their behaviour. Recently, Chitsaz et al. (2018) developed a model where the fluid-solid interaction was used for studying aspiration thrombectomy by comparing existing Penumbra catheters. One of the points addressed in their study was related to the computation cost of FEM (Finite Element Method) models.

The study we present is specifically focused on suction removal of the clot and on its behaviour. Our modelling shows that the Bond-Graph technique is useful in representing different simulation conditions, enabling the incorporation of different parameters in an effective manner. It also allows us to quickly obtain the

* Corresponding author at: Universidad Politécnica de Madrid, C/José Gutiérrez Abascal 2, 28006 Madrid, Spain.

E-mail addresses: ctalayero@gmail.com (C. Talayero), gregorio.romero@upm.es (G. Romero).

boundary conditions, to use in more complex modelling based on three-dimensional methods such as FEM, Agent Based Simulation (ABP) or Multi-body Simulation (MBS).

We have recently simulated a clot as a two-dimensional model which incorporates several components including the relationship between platelet-platelet in longitudinal and transversal directions, and between platelet-wall (Talayero et al., 2018). In this way, we can investigate the parameters and conditions that might potentially damage the artery and the thrombus. Any such potential damage can be assessed by examining different criteria related to stress, eg wall shear stress due to friction, wall normal stress due to suction pressure, clot equivalent stress (i.e. Von Mises) or, as in our case, by the strain of the elements that form the clot (platelet-platelet bond that represents the fibrin clotting network in our modelling).

The Bond-Graph modelling technique involves using a differential equation for each partition. Although this approach is successful, it suffers from the drawback that for models with a large number of partitions, a great deal of computing time is needed. Preliminary studies analysed a model of 5×5 partitions, which was considered sufficient given the relatively small size of the artery (from 1 to 3 mm in diameter) involved in this instance, and the practical possibilities of manufacturing the catheter. When the model was solved, we obtained the load and displacement for each pair (platelet-platelet or platelet-wall). However, if two-dimensional Bond-Graph modelling is used in conjunction with MBS, suction, deformability, and other parameters of the clot, can be modelled more effectively. Previous studies have involved using FEM and co-simulation algorithms between Computer Fluid Dynamics (CFD) and nonlinear dynamic mechanical representations (Talayero et al., 2018) which involves high computing costs.

2. Methods

2.1. Modelling the device

2.1.1. Fluid domain

A resistance – compliance model was chosen for modelling the fluid, comprised of the pump, the catheter, the distal end of the catheter, the deflecting nozzle and the artery.

The pump creates the pressure needed to undertake clot extraction. It is represented by a pressure source ($Se = P$) that varies over time reaching 100 kPa in 1 s, after which time the pressure provided by the pump remains constant.

The catheter is represented as a cylindrical tube about 1 m in length and a diameter determined by the size of the blood vessel (from 1 to 2 mm approximately). In the model it is represented as a pipe with the following parameters in the interior of the vessel: load (R), the inertia loss (I), and the compressibility of the blood (C). The friction between the liquid particles and the pipe walls is responsible for the primary load losses. The distal end of the catheter is taken to be of the same diameter as the rest of the catheter and is of length 10 mm. Additionally, the distal end of the catheter is deemed to have the same three parameters (R, I, C) and with the corresponding values. Finally, because the artery is surrounding the distal end of the catheter, it is necessary to consider the transition between both elements as a pressure loss caused by the difference in diameter, and the subsequent variations in flow (Eq. (1)).

$$\Delta P = \frac{\lambda \cdot v^2 \cdot \rho \cdot L}{2 \cdot D} \quad (1)$$

where ' λ ' is the load loss coefficient, ' v ' is the average fluid velocity inside the catheter, ' ρ ' is the blood density, ' L ' the length of the section and ' D ' the inner diameter of the catheter.

From (1) and considering a laminar flow in a cylindrical section, the resistance and the inertia are as follows:

$$R_{Blood_catheter} = \frac{128 \cdot \mu \cdot L}{\pi \cdot D^4} \quad (2)$$

$$I_{Blood_catheter} = \frac{\rho \cdot L}{\pi \cdot \left(\frac{D}{2}\right)^2} \quad (3)$$

where ' μ ' represents the dynamic viscosity of the fluid.

The compressibility of the catheter is considered to act as a spring producing a decrease in volume when the pressure is increased. The blood's bulk coefficient 'B' is defined by a capacitance with a value given by the following expression to represent the longitudinal stiffness:

$$K_{Blood_catheter} = \frac{4 \cdot B}{\pi \cdot D^2 \cdot L} \quad (4)$$

The pressure losses corresponding to the transition from the distal end of the catheter to the artery (which have different diameters) are represented as a secondary loss and can be represented by adding a new resistance (R_{nozzle}).

$$R_{Nozzle} = 8 \cdot \rho \cdot \zeta \cdot \frac{Q}{\pi \cdot D^4} \quad (5)$$

where ' Q ' is the flow that exists in a pipe segment and D is the mean diameter between the cylinder and the artery. The coefficient ' ζ ' is the load loss coefficient (dimensionless) which depends on the difference between the artery diameter and the catheter distal end diameter. The coefficient can be considered as 0.3 in the best case (Romero et al., 2013b). This modelling allows consideration of the blood profusion around the catheter and the flow between the catheter and the clot.

Ideally, the distal end of the catheter is positioned at a distance of approximately 3 mm from the blood clot when the suction begins (Pearce et al., 2010; Tennuci et al., 2011). However, while the clot is traversing this distance of 3 mm, the artery can become deformed too due to the forces acting. We therefore treat this section of the artery in the model as another section, and as such it is necessary to insert a parallel capacitance (stiffness) that represents the compressibility of the artery, in accordance with its Young's modulus (E), its thickness (h), the artery initial volume (V_0) and the initial radius (r_0), as represented in equation (6).

$$K_{Artery_wall} = \frac{E \cdot h}{V_0 \cdot 2 \cdot r_0} \quad (6)$$

Fig. 1 shows a representation of the aspiration thrombectomy intervention. In Fig. 2 the Bond-Graph model for the fluid domain is represented.

2.1.2. Mechanical domain

The clot interaction is the most complex part of the modelling. The blood clot is approximated by a cylindrically shaped element in the range 5–50 mm in length, and a mass in the range 0.05 and 0.1 g. The model considers a spring-damper system in parallel in order to simulate the elastoplastic behaviour of the clot (that it undergoes in resisting traction due to the applied suction pressure). To model the interaction between the clot and the artery wall, we need to consider the static and dynamic friction at each instant. While the clot is in the position of static friction, each inertia will suffer a force due to suction, which should be compensated for in the model by inserting an equal (and opposite) force of friction.

The three-dimensional blood clot representation enables a better study of the clot stress. By connecting several elements with a spring-damper link, the bidimensional model of the Bond-Graph

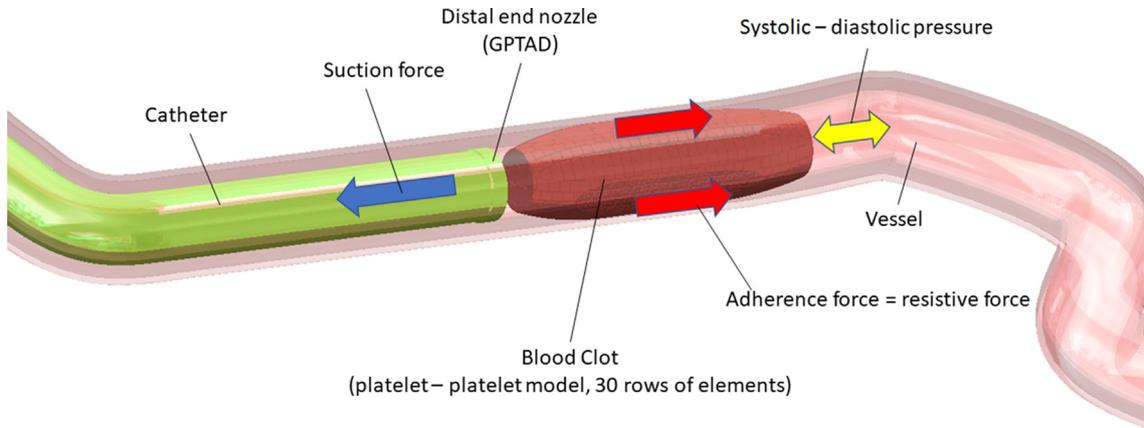


Fig. 1. Aspiration Thrombectomy Device representation showing all the components and interactions that are modelled in the fluid and mechanical domains.

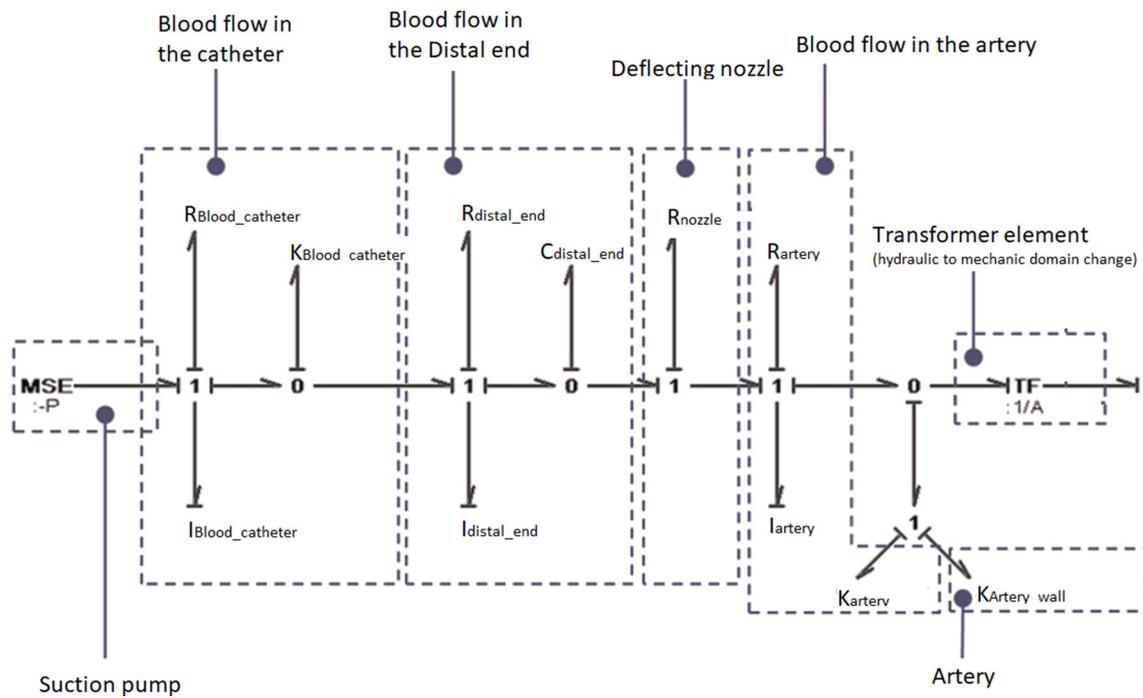


Fig. 2. Model representation with Bond-Graph. Fluid domain.

technique can be converted into a three-dimensional geometry arrangement, where not only the longitudinal elastoplastic behaviour is reproduced, but also the transversal behaviour is taken into account (Fig. 3).

The clot is 5–50 mm long, which means that (for convenience) it can be broken down into several “spheres” joined together, all with the same constant, as shown in Fig. 4. Since over the whole surface of the clot there are forces of adhesion to the artery wall, in order to obtain a correct approximation, it is necessary to consider the existence of a “sphere” for every 0.2 mm. Consequently, between 25 and 250 “spheres” would need to be included. The number of “spheres” will give the final K_{union} (stiffness) and R_{union} (damping coefficient). The parallel-series system equations are solved at a previous stage in the modelling process using the equations:

$$K_{eq_parallel} = K_1 + K_2 + \dots + K_n \quad (7)$$

$$\frac{1}{K_{eq_series}} = \frac{1}{K_1} + \frac{1}{K_2} + \dots + \frac{1}{K_n} \quad (8)$$

As established in previous models (Romero et al., 2013b), the stiffness can be assumed as being $K_{union} = 3.41 \pm 1.5$ N/m and the damping coefficient as $R_{union} = 0.035$ kg/m.s. The number of elements for the analysed model is chosen by comparing the spheres with a hexahedral mesh used in a FEM model. Thus, with at least 19 elements per row for the chosen diameter, convergence is achieved.

In a complete Bond-Graph model, the transition from fluid domain to mechanical domain requires a transformer (TF) element. To calculate the value of the coefficient defining this element (R_{TF}), the change in the definition of the flow before and after the TF element is evaluated. Before the TF element, the flow is in the hydraulic domain, while after, it is in the mechanical domain. The coefficient R_{TF} is obtained by evaluating the required change between both domains. The equations that relates both flows are:

$$f_2 = f_1 \cdot R_{TF} \quad (9)$$

$$f_1 = Q = v_{fluid} \cdot A_{clot} \quad (10)$$

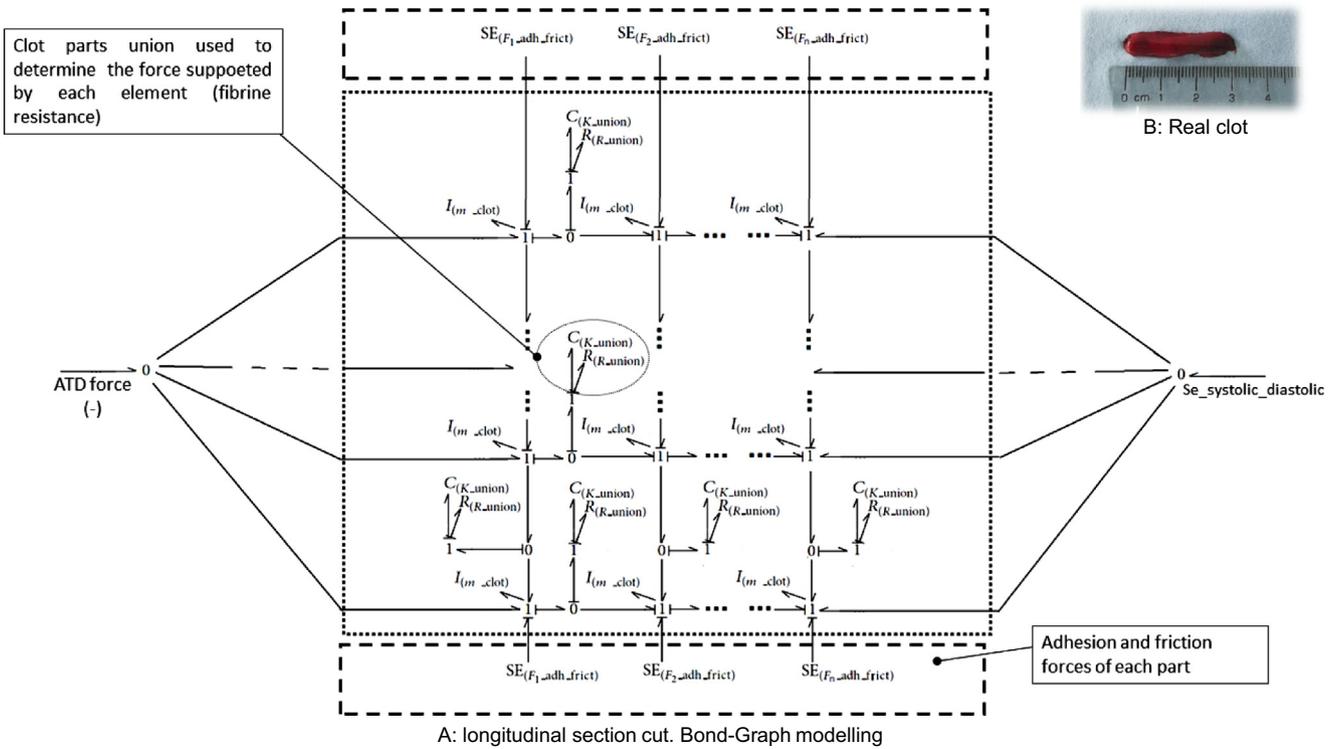


Fig. 3. Model evolution process. (A) Real blood clot. (B) Longitudinal section cut. Bond-Graph modelling.

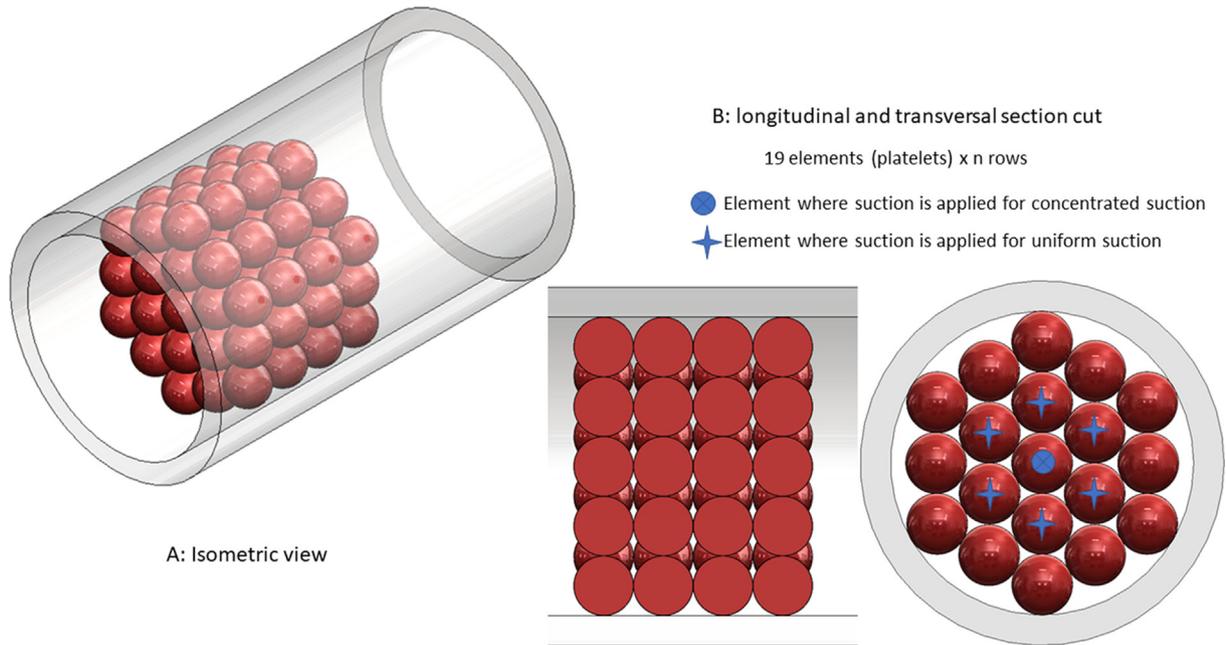


Fig. 4. Multi-Body geometrical model. (A) Isometric. (B) Longitudinal and transversal section cut.

$$f_2 = v_{clot} \tag{11}$$

$$R_{TF} = 1/A_{clot} = 1/(\pi \cdot R_{clot}^2) \tag{12}$$

where “ f ” is the flow (in hydraulic and mechanical domains), “ v ” is the volume (for both domains), “ A_{clot} ” is the transversal area of the clot, calculated as a circular section where ‘ R_{clot} ’ is the clot radius. In the case of 100% arterial occlusion, it is equal to the artery radius.

We note that in an actual artery (as in our model also), the fluid does not acquire the same velocity in the centre of the catheter compared to a point near to the walls of the artery, and as such in some previous models, a mean value of velocity was used for purposes of modelling. However, our model takes a different approach. We apply a force according to a theoretical pressure field depending on the catheter inner geometry (cylindrical, conical, helical or drilled) (Talayero et al., 2018). Consequently, the force

can be applied with a combination of values of axial load and torsional load due to the configuration of the helical aspiration device (the GPTAD). The applied suction pressure distributions are based on existing or patented solutions (Bonnette and Morris, 2014; Cosigny, 2013; Pearce and Perkinson, 2006; Milner and Barnett (2015)). In our case, since we are applying the suction to the platelets, the force calculated from the fluid domain is divided into the number of elements. Thus, in the case of “concentrated” suction, the load is applied in the central element. In the case of “uniform suction”, that can be achieved with geometries as in (Talayero et al., 2018), the load is divided into the six elements that surround the central element (Fig. 4).

The interaction of the blood clot and the artery is evaluated by including in the model an equal (and opposite) force of friction to eliminate the force needed to overcome the inertia. Having defined the clot, the adhesion force is calculated by the platelet adhesion force; using from Flannery (Flannery, 2005) the equivalent number of platelets per area in the equation:

$$\text{No. of platelets per area} = fp \cdot \frac{SA}{MPA} \quad (13)$$

where “fp” is the percentage of platelets in the clot, “MPA” is the mean platelet area, and “SA” is the area of the surface in contact with the vessel, that is obtained from the clot length and the vessel diameter. This approach allows us to have different forces of adhesion depending on the geometry and size of the clot, as modelled in previous models (Romero et al., 2013a, 2013b). If the occlusion of the vessel is not 100%, the model allows us to assign a force due to suction pressure to the elements that are not directly attached to the walls.

2.2. Computational methods

2.2.1. Computational domain and solver

As shown in Fig. 1, the computational domain comprises the fluid domain (catheter and vessel) and the mechanical domain (vessel and clot). The fluid domain is shown in Fig. 2 in the Bond-Graph representation. It begins at the pump and finishes at the space between the catheter and the blood clot. The mechanical domain is shown in Fig. 4, where a section of the vessel is shown with the platelets in the inside. The total number of elements depends on the clot length.

A WSTIFF solver is used for the multibody problem, similar to other stiff solvers. They use a backwards difference formulation. The coefficients of other solvers are calculated assuming a constant step size, whereas WSTIFF coefficients are a function of the step size. If the step size changes suddenly during integration, other solvers introduce a small error, while WSTIFF can handle step size changes without loss of accuracy.

2.2.2. Numerical parameters

We aim to investigate the minimum pressure required for blood clot extraction under given conditions, and to determine the most efficient way of applying the suction to achieve the recanalization with a low risk of damage to both the vessel and the clot. We study several options by using a three level DOE (Design of experiments), investigating such factors as the suction pressure (50, 80 and 100 kPa), the distance between the distal end of the catheter and the clot (0, 2 and 4 mm) and the diameter of the catheter in relation to the section of the vessel, represented as the percentage of free area between them (0, 5 and 10%). This factor allows us to predict the behaviour in different vessel diameters, since the catheter diameter can be easily chosen in accordance with standard sizes available. Other factors are related to the clot properties derived from other studies (Merritt et al., 2018) and models (Chitsaz et al., 2018; Romero et al., 2011, 2013a, 2013b), such as the clot length (3; 10 and 30 mm), the K_{union} (1.91; 3.41 and 4.91 N/m)

and R_{union} (0.025; 0.035 and 0.045 kg/m·s). The factor chosen to analyse the vessel integrity is the Young Modulus (0.5; 0.8 and 1.1 MPa). Finally, two factors that relate the vessel and the clot are considered: namely, the occluded area of the vessel by the clot, measured as a percentage (80, 90 and 100%) and the occlusion force (0.01; 0.05 and 0.1 N). The Middle Cerebral Artery is about 2.5 mm in diameter, so this diameter is considered. In previous studies, consistent results were obtained for arterial diameters varying from 0.5 to 5 mm (Romero et al., 2011). The diameter of the artery determines the size of the catheter that may be used to remove the clot.

The values of the constant parameters used in the simulation are listed in Table 1.

2.2.3. Model tuning and verification

Bond-Graph models have previously been verified and validated with in-vitro experiments (Pearce et al., 2010; Tennuci et al., 2011). In order to verify the behaviour of the clot, an undamped one-dimensional mass-spring model is created which is compared to a cylinder modelled with MBS. An elongation of 1.68 mm is obtained for a 3 mm length cylinder with the elastic properties of the clot. This is consistent with the results shown in Fig. 7. The number of elements to be modelled in MBS are also verified as described in Section 2.1.2.

3. Results and discussion

The variables we use to compare the solutions modelled are: time to obtain clot extraction, vessel integrity, clot behaviour, and the maximum load on it. In the Multi-Body analysis, the load is given by the total elongation of the spring between two elements and the speed of elongation, that directly affects the load (damping force). When high strains or high speeds are detected, the risk of embolization is higher.

3.1. Time to extraction

The time taken for clot extraction will depend directly on the mass and length of the clot. Since in this study we are considering a fixed diameter vessel, the mass and the length are directly related, and typical values are shown in Fig. 5. It is also noted that, for a given size, the higher the stiffness of the clot (defined by K_{union} and R_{union}), the shorter the required time. This is due to a smaller elongation that arises during the suction; this case having a greater similarity to a rigid body than to an elastic element. These factors are also related to the viscosity and composition of the clot. Our modelling verifies that the time for clot extraction is shorter with higher pressure values (Fig. 5).

3.2. Vessel integrity

The suction pressure created in the catheter nozzle affects the vessel, and it generates a radial narrowing that depends on the $K_{\text{Artery_wall}}$ as defined in (6). The vessel is also affected by the shear loads due to the axial flow. It is noted that for the highest pressures (100 kPa) and a catheter positioned at about 4 mm from the clot,

Table 1
Parameters definition.

Blood density (ρ)	1060 kg/m ³
Artery diameter	2.5 mm
Blood viscosity (η)	0.0035 Pa·s
fp	0.96
MPA	5.31·10 ⁻⁶ mm ²
F _{adhesion_platelet}	32·10 ⁻⁹ N

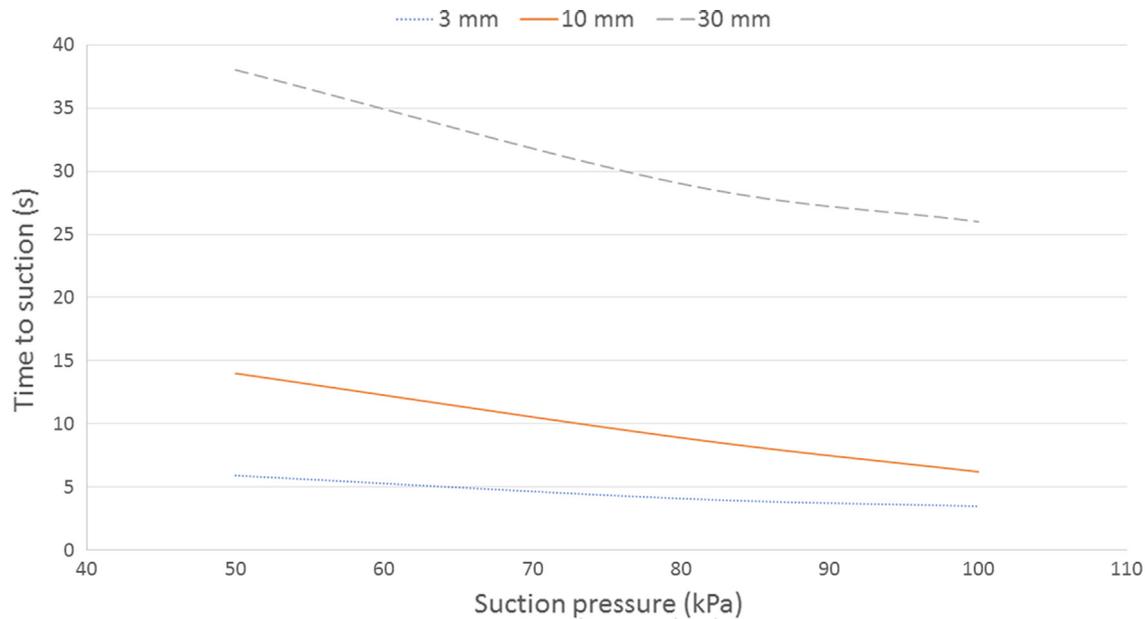


Fig. 5. Suction pressure vs time and blood clot size plot.

the vessel could potentially be damaged due to the resultant high radial pressure, and consequently a big strain would result. However, for 50 and 80 kPa no such potential risk is apparent.

3.3. Clot behaviour

Fig. 6 shows the sequence of movement of the blood clot (a 3 mm clot is shown for a 100% vessel occlusion). This movement can be seen clearly in this type of modelling, since the movement of the clot not only depends on the movement of each slice, but on the internal disaggregation that can lead to a loss of contact between the clot and the artery wall. The disaggregation can be assessed by comparing the value of the internal force between two elements and comparing it with the clot limits. The force values are directly related to the way the suction load is applied. If the pressure is concentrated on a small area close to the revolution axis (Fig. 6A, B, C, D) (as in a direct aspiration catheter), the clot will collapse and that will lead to movement of the clot due to the lack of contact between the clot and the artery wall. However, if the internal loads are much higher, there is an increased chance of clot breakage (fragmentation) when clot movement occurs, thus resulting in a potential risk of small pieces of the fragmented clot moving upstream. On the other hand, if the pressure is applied on an area far from the axis and near to the wall (“uniform suction”) (Fig. 6E, F, G, H), the clot will remain in contact with the wall during the first stages of suction (Fig. 6F and G). This leads to a less elongated clot (Fig. 6H shows a more compact clot than Fig. 6D), thereby reducing the risk of clot breakage (fragmentation). These results are qualitatively similar for all the simulated lengths, so we can imply that certain catheter geometries can potentially improve the recanalization results, independent of the clot properties. With respect to catheter size, the most effective suction is achieved when the catheter diameter is similar size to the blood clot diameter (i.e. there is 0% free area between clot and vessel wall). Other options result in a pressure loss that results in a decrease in the efficiency of the intervention process, as predicted by Eq. (1).

3.4. Internal loads

Fig. 7 shows the clot elongation during an initial suction time of 3.5 s for the model in Fig. 6. It is represented as the behaviour of an

element at the proximal end of the clot (the highest spring elongation), and an element at the distal end of the clot (smallest spring elongation). As can be seen by noting the length of the clot, when the pressure is concentrated on a point, the clot becomes slightly longer compared to when the pressure is distributed over a greater area. The “concentrated” or “uniform suction” conditions are explained in 2.1.2.

The oscillatory behaviour shown in the last element is due to two factors: the systolic–diastolic pressure and the elastic links between the elements. When the force due to suction (or due to the movement of an element of the $i + 1$ row) overcomes the friction force, a linked element of the $i + 1$ row begins its movement. When the distance between the elements changes, in accordance with Hooke’s law, the force of the spring changes proportionally.

When we consider the case of a 100% clot occlusion and the maximum clot length is considered (30 mm), and we apply the suction far from the artery walls, the indications are that the clot would potentially be broken (i.e. fragmented). In this instance, more than one extraction attempt may be necessary to completely remove the clot from the artery, on account of the higher mass (inertia) and greater adhesion surface between the clot and the artery. However, for smaller lengths (up to 10 mm) the influence of the force of adhesion (due to the surface in contact) is not so significant as in other cases. The results of such modelling potentially provide a useful adjunct to physical measurements in future clinical studies.

4. Conclusions

From the study undertaken we conclude that:

- Our results confirm that aspiration thrombectomy is potentially a good option for recanalization of arteries after stroke. Suction appears to allow the removal of the clot without potentially damaging the artery, except when the suction pressure is relatively high, and the distal end of the catheter is relatively far from the clot.
- Our 3D modelling provides a relatively streamlined approach and enables general parameters to be found that potentially describes clot extraction behaviour. The most significant inclusion in the model is the link between unidimensional calculus for the fluid domain, and MBS for the mechanical domain. Good

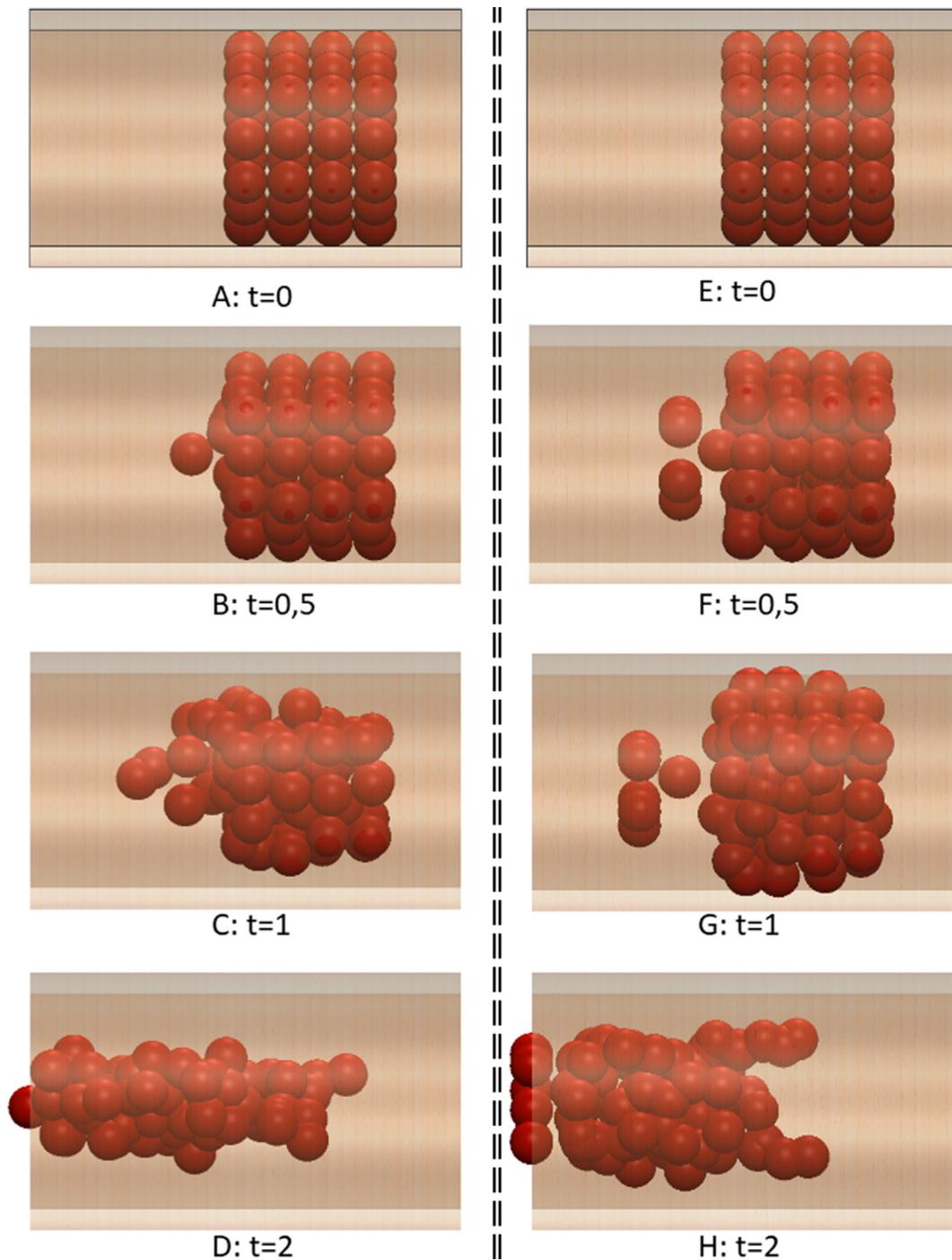


Fig. 6. Clot movement sequence. (A) Concentrated suction, $t = 0$ sec. (B) Concentrated suction, $t = 0.5$ sec. (C) Concentrated suction, $t = 1$ sec. The clot collapse and the movement as whole begins. (D) Concentrated suction, $t = 2$ sec. Shape of the clot when it reaches the catheter. (E) Distributed suction, $t = 0$ sec. (F) Distributed suction, $t = 0.5$ sec. (G) Distributed suction, $t = 1$ sec. Some elements (platelets) are disaggregated from the clot. The clot remains still. (H) Distributed suction, $t = 2$ sec. Shape of the clot when it reaches the catheter.

agreement was found with previous Bond-Graph and coupled CFD and FEM models. The MBS allows to us include the adhesive force between the clot and the vessel wall, and the internal cohesive forces, that are important parameters governing clot extraction.

- The error margin of this modelling is of the order $\pm 20\%$ based on the load and time for extraction obtained from previous in-vitro tests. The factors that impact on the error are mainly the geometry discretisation, the fluid domain Bond-Graph mod-

elling, the MBS solver and the material properties (tissue and blood clot).

- Solutions can be obtained in a faster way than with other computational tools (a 100 elements clot is solved in less than 6 min in a standard Core i7 processor).
- This modelling can be improved by the addition of the behaviour of potentially broken particles (“spheres”) and studying the resulting movement after clot breakage.

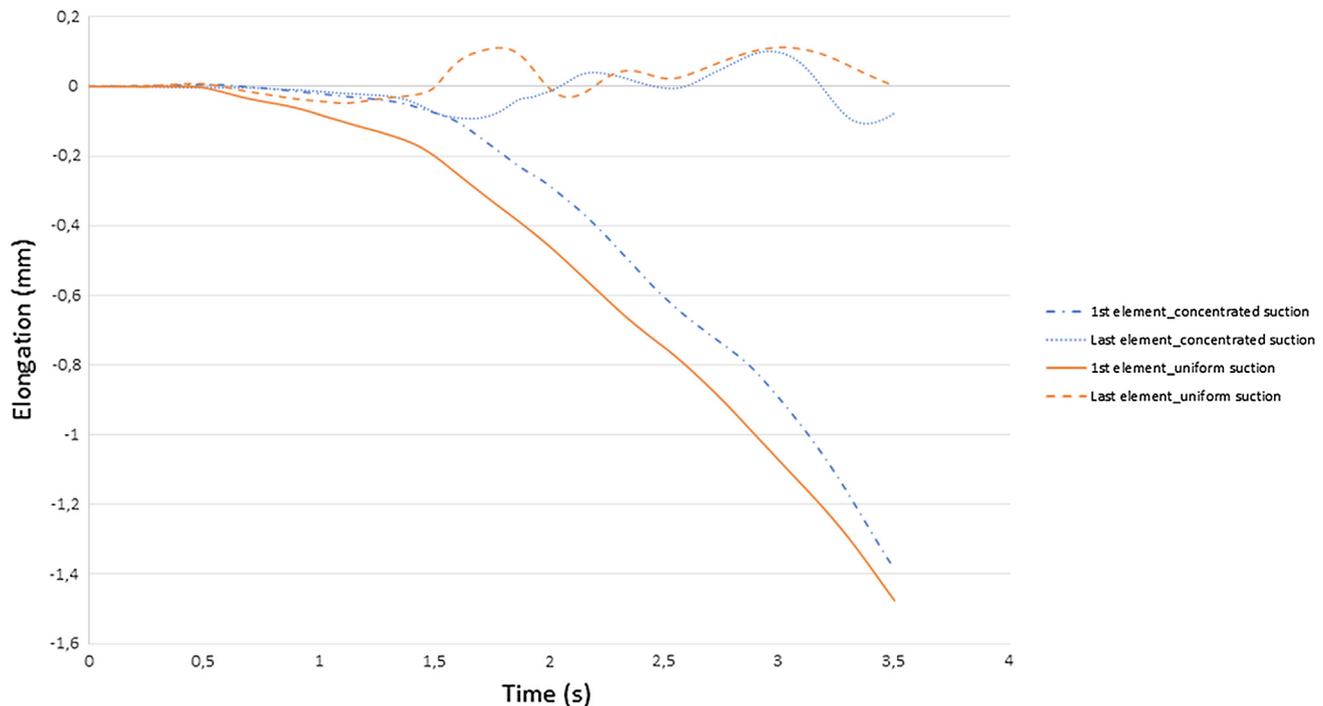


Fig. 7. Elongation vs time plot for different suction pressure distribution.

- Applying the suction pressure in an area close to the interface between the clot and the vessel may potentially improve recanalization results.
- Although our results are based solely on mathematical modelling and computer simulations, we conclude that the results of such modelling are potentially useful in providing constraints on physical parameters that could be used in future laboratory or clinical studies. Further studies relating to different physical blood clot properties could also potentially be useful in predicting clot extraction outcomes in different patients.
- Since a robust design process is undertaken, based on a DOE, the results are relatively independent of the blood clot properties in respect of comparing the parameters studied and the combination of these in relation to each other.

Funding statement

This research received no specific grant from any funding agency in the public, commercial or not-for-profit sectors.

Contributorship statement

All authors fulfill the following requirements:

- Substantial contributions to the conception or design of the work; or the acquisition, analysis, or interpretation of data for the work; AND
- Drafting the work or revising it critically for important intellectual content; AND
- Final approval of the version to be published; AND

In particular, Talayero and Romero are responsible for the mathematical models and simulation. Pearce and Wong oriented the research in the medical aspects and results analysis.

Declaration of Competing Interest

The authors declare that they have no conflict of interest.

References

- Bonnette, M., Morris, D., 2014. Thrombectomy Catheter System (patent US2014303658A1).
- Chitsaz, A., Nejat, A., Nouri, R., 2018. Three-dimensional numerical simulations of aspiration process: evaluation of two penumbra aspiration catheters performance. *Artif. Org.* <https://doi.org/10.1111/aor.13300>.
- Cosigny, P., 2013. Spiral thrombectomy catheter (patent US2013060206A1). doi:10.1016/j.jstrokecerebrovasdis.2009.03.017.
- Fanous, Andrew A., Siddiqui, Adnan H., 2016. Mechanical thrombectomy: stent retrievers vs. aspiration catheters. *Cor Vasa* 58 (2), e193–e203. <https://doi.org/10.1016/j.crvasa.2016.01.004>.
- Flannery, C.J., 2005. Thrombus Formation Under High Shear in Arterial Stenotic Flow [Ph.D. thesis]. Georgia Institute of Technology, Atlanta, Ga, USA.
- Lapergue, B., Blanc, R., Guedin, P., Decroix, J.P., Labreuche, J., Preda, C., Piotin, M., 2016. A direct aspiration, first pass technique (ADAPT) versus stent retrievers for acute stroke therapy: An observational comparative study. *Am. J. Neuroradiol.* 37 (10). <https://doi.org/10.3174/ajnr.A4840>.
- Merritt, W., Holter, A.M., Beahm, S., Gonzalez, C., Becker, T.A., Tabor, A., Frenklakh, S., 2018. Quantifying the mechanical and histological properties of thrombus analog made from human blood for the creation of synthetic thrombus for thrombectomy device testing. *J. NeuroIntervent. Surg.* <https://doi.org/10.1136/neurintsurg-2017-013675>.
- Milner, K., Barnett, A., 2015. Thrombectomy catheter with flow directing mechanism (patent US2015133973A1).
- Mokin, M., Ionita, C.N., Nagesh, S.V.S., Rudin, S., Levy, E.I., Siddiqui, A.H., 2015. Primary stentriever versus combined stentriever plus aspiration thrombectomy approaches: in vitro stroke model comparison. *J. NeuroIntervent. Surg.* 7 (6), 453–457.
- Mokin, M., Nagesh, S.V.S., Ionita, C.N., Mocco, J., Siddiqui, A.H., 2016. Stent retriever thrombectomy with the cover accessory device versus proximal protection with a balloon guide catheter: in vitro stroke model comparison. *J. NeuroIntervent. Surg.* 8 (4), 413–417.
- Pearce, G., Perkinson, N.D., 2006. Biomechanical probe international patent corporate treatise (WO2006120464); European patent (ep1893195 (a2)); Japanese patent (jp2008639924 (t)); Chinese patent (cn101208049 (a)).
- Pearce, G., Brookfield, P., Burley, M., Murcott, D., Perkinson, N.D., Spence, J., Wong, J., 2009. The “GP” mechanical thrombectomy device: measurements and modelling, using laboratory models, applications of geometric algebraic techniques and cadaveric brains. *J. Stroke Cerebrovasc. Disease* 18 (4), 288–293.
- Pearce, G., Perkinson, N.D., Wong, J., Roffe, C., Brooker, L., Jones, K., Dodd, M., Spence, J., Rai, M., Brookfield, P., 2010. In vitro testing of a new aspiration thrombus device. *J. Stroke Cerebrovasc. Disease* 19 (2), 121–129.

- Romero, G., Martínez, M.L., Felez, J., Pearce, G., Perkinson, N.D., 2011. Applicability of the GP device to the circle of willis arteries by using a mathematical model. In: Proceedings of 2011 UkSim 13th International Conference on Computer Modelling and Simulation (UKSim) [Internet]. EEUU: IEEE, pp. 48–53.
- Romero, G., Martínez, M.L., Pearce, G., Mera, J.M., 2013b. An investigation into the performance of a new mechanical thrombectomy device using bond graph modelling: application to the extraction of blood clots in the middle cerebral artery. *Simulation* 89 (3), 381–391.
- Romero, Gregorio, Martínez, M. Luisa, Maroto, Joaquin, Felez, Jesus, 2013a. Blood clot simulation model by using the bond-graph technique. *Sci. World J.* 2013, 1–10. <https://doi.org/10.1155/2013/519047>.
- Scarborough, P., Peto, V., Bhatnagar, P., Kaur, A., Leal, J., Luengo-Fernandez, R., Gray, A., Rayner, M., Allender, S., 2009. Stroke Statistics. British Heart Foundation and Stroke Association, pp 12–13.
- Spiotta, A.M., Chaudry, M.I., Hui, F.K., et al., 2015. Evolution of thrombectomy approaches and devices for acute stroke: a technical review. *J. NeuroIntervent. Surg.* 7, 2–7.
- Talayero, C., Romero, G., Pearce, G., Wong, J., 2018. Simulation of Blood clot removal by aspiration thrombectomy in cerebral vessels. Geometry optimization of the aspiration device. *Int. J. Simul. Syst. Sci. Technol.* 19 (5). <https://doi.org/10.5013/ijssst.a.19.05.18>.
- Tennuci, Christopher, Pearce, Gillian, Wong, Julian, Nayak, Sanjeev, Jones, Tom, Lally, Frank, Roffe, Christine, 2011. Comparison of the effectiveness of three methods of recanalization in a model of the middle cerebral artery: thrombus aspiration via a 4F catheter, thrombus aspiration via the GP thromboaspiration device, and mechanical thrombectomy using the solitaire thrombectomy device. *Stroke Res. Treat.* 2011, 1–6. <https://doi.org/10.4061/2011/186424>.