SIMULATION OF THE 'GP' MTD DEVICE INTENDED FOR THE EXTRACTION OF BLOOD CLOTS BY USING THE BOND GRAPH TECHNIQUE.

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ABSTRACT

This article covers the analysis and research into a device recently developed by the University of Wolverhampton (UK), called a *'GP' MTD Mechanical Thrombectomy Device*, under the direction of Dr G. Pearce. This device will improve the process of extracting thrombosis clots in the cerebral arteries. On the one hand, the development of the simulation model of this device is shown by using Bond-Graph formalism and, on the other hand, the optimization of its performance in the very near future, from the interpretation of the results.

KEYWORDS

Health sciences; Bond Graphs, Blood-clotting, Multi-formalism modelling

INTRODUCTION

This article Approaches the study and research into a device recently developed under the supervision of Dr G. Pearce from Wolverhampton University (United Kingdom,) called a *'GP' Mechanical Thrombectomy Device (MTD)* (Pearce et al. 2007, Pearce et al. 2008, Pearce et al. 2009, Rai et al. 2009). This device allows improving the process of extraction of typical thrombosis clots in cerebral arteries. Presented in this paper is the development of the model of this device by means of the Bond Graph technique, as well as its simulation and interpretation of the results obtained with the purpose of optimizing its operation for future use.

The aim of the simulation model that is presented is to obtain the minimum pressure necessary to extract the clot and to check that, both this pressure and the time required to complete the operation are reasonable for use in patients, and are in line with experimentally obtained data. It is therefore necessary to consider aspects from the domains of hydraulics and mechanics. The Bond Graph technique (Karnopp et al. 1990) was chosen for its simplicity and suitability for a combined study of both domains.

STATE-OF-THE-ART

Thrombosis is produced by the formation of a clot inside the blood vessels causing an abrupt interruption of the blood flow. In the cerebral arteries this occlusion takes place due to the presence of a clot that has formed at another location of greater diameter which obstructs the cerebral artery due to its smaller cross section.

The process to eliminate this obstruction is called catheterism and different devices exist to carry out this operation. All of them function by introducing a catheter into the artery to eliminate the clot, generally by pushing it. The device under study in this paper, compared to those currently in use, is based on the suction of the obstructing element by creating a vortex, the advantage being that smaller risks are associated with its use.

For the study of this device, different existing techniques have been considered for the modelling and simulation. Some methods which have been considered for their applicability are the Boltzmann flow simulation technique, and finite elements modelling and its implementation in *Matlab* © software, with 2D or 3D models; or by means of Laplace transformations using Dynamic Motion Solver software.

Finally, the method chosen for the representation and simulation of this model is the Bond Graph technique. Its choice is based on the fact that this technique allows assimilating the model to an electric circuit made up of resistances, capacitances and inductances. Therefore, it is possible to obtain the results in a simple way by evaluating flows and efforts that join and connect the components of the model.

This paper gives a brief description of the device under study, as well as the parts comprising it. Next, the model used for the simulation is described and the phenomena considered to define the device, and, in addition, the values of the parameters used are defined. Lastly the results obtained and the conclusions of this study are attached.

MODEL DESCRIPTION

The specific device shown in this article is formed by a pump that provides the necessary suction pressure for the operation, joined to a very long catheter. The 'GP' is located at the end of this catheter.

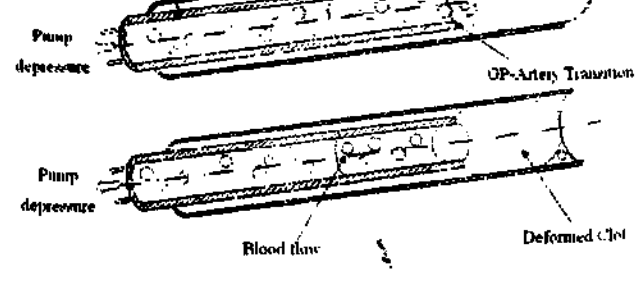

Figure 1. 'GP' Device.

It is a hollow cylinder with the same diameter as the catheter. Its interior is the place where the vortex is created to carry out the extraction. This device is introduced into the cerebral artery at the place where the clot is, and is positioned at a distance of 3mm from it. Then the suction begins until the clot is extracted. The clot crosses the 3mm that separate it from the 'GP' and becomes encrusted with it due to the difference in diameters. Once it is encrusted, the device is removed from the body.

To obtain the simulation of the model, *Bondin* © software will be used (Romero et al. 2009). This program allows obtaining the evolution of the characteristic parameters of the model as well as letting them be compared.

Listed below are the components of the model.

Pump

The pump is the component that creates the necessary pressure to carry out the extraction. It is represented by a variable pressure source whose value will increase from zero to 40 kPa, a figure which experience shows to be suitable for carrying out this operation. The time taken to reach 40kPa is 3 sec., after which time the pressure provided by the pump remains constant.

Catheter and 'GP'

The catheter is a 110 cm long 1 mm diameter hollow cylindrical tube. It is joined to the 'GP' cylinder of the same diameter and a length of 20mm. In order to represent both elements, they are considered as several pipe sections bearing in mind the different phenomena that take place in their interior: load and inertia loss, and fluid compressibility

Linear load loss is due to the friction between the liquid particles and the pipe walls. Due to their being straight pipes, only linear load losses are taken into account. As this pipe is horizontal and of constant cross section in each section, the load loss is reduced to a pressure loss as the fluid advances along the pipe, the loss being progressive and proportional to the length of the pipe. It is represented by a resistance R, and a type 1 junction.

To determine the equation that governs its behaviour, it is necessary to know if the behaviour of the blood flow is laminar or turbulent. This is evaluated by the Reynolds number, giving the following value:

$$\mathrm{Re} = \frac{V \cdot D}{\nu} \approx 1000 < 2200 \qquad (1)$$

The behaviour is laminar, and can be determined by the following expression:

$$R = \frac{128 \cdot \eta \cdot L}{\pi \cdot D^4} \qquad (2)$$

where η is the dynamic viscosity of the blood flow, L the length of the pipe section and D its diameter.

Secondly, the flow inertia to be overcome in its movement is taken into account. It is represented by a type I, port and a type 1 junction. The expression is:

$$I = \rho \cdot L / A \qquad (3)$$

where ρ is the blood density, L the length of the pipe and A its cross section. Considering this section with circular geometry:

$$A = \pi \cdot \left(\frac{D}{2}\right)^2 \qquad (4)$$

Lastly, the blood compressibility is included. It acts as a spring producing a decrease in volume when the pressure required for compression is increased. This behaviour is dependent on Bulk's blood coefficient (B) and it is defined by a capacitance C with a type 0 junction, by the following expression:

$$K = 4 \cdot B / \pi \cdot D^2 \cdot L \qquad (5)$$

In the model, first the pump is positioned then the catheter. Due to its great length it is represented by ten identical sections that include the three previously described phenomena. Thanks to this representation, it is possible to study the evolution of the pressure loss along the catheter. Later the 'GP' is positioned and is represented by the three previous phenomena.

'GP'-Artery junction

The artery is located at the end of the 'GP'. The transition between both elements is considered as a secondary load loss caused by the difference in diameter of both elements and the subsequent variations in flow. These load losses are modelled by a resistance R and can be calculated with the following expression:

$$R = 8 \cdot \rho \cdot \xi \cdot \frac{Q}{\pi^2 \cdot D^4} \qquad (6)$$

where P is the load loss, V the mean speed of the flow in this section and ξ, the load loss coefficient. As the pressures are not equal, the junction is type 1.

The load loss coefficient ξ is an adimensional parameter that quantifies the loss produced and depends on the geometry of the junction. Since this is a narrowing, this value is 0.4.

The flow is not constant during the extraction, so to calculate this expression, its value at each instant is considered like the flow of the inertance that represent the GP device (*flow(I_{GP})*).

The diameter indicated in the expression, is the mean diameter between the cylinder and the artery, calculated in the following way:

$$D_{medium} = \frac{D_{cilinder} + D_{artery}}{2} \qquad (7)$$

Artery

The artery located between the end of the 'GP' and the clot is included in the model as another section of a pipe, similar to the catheter and the 'GP'. It is defined by the loss of linear load (R), the inertia (I) and the compressibility of the blood (C).

In addition, it is necessary to insert a parameter that represents the compressibility of the artery, in line with its Young's modulus:

$$K = \frac{E \cdot h}{V_0 \cdot 2 \cdot r_0} \qquad (8)$$

Domain change

Once all the elements are defined by fluid mechanics, it is necessary to change from the domain of hydraulics to mechanics, to be able to evaluate the movements and efforts in the clot, as well as to define the physical friction between the clot and the artery.

This domain change is carried out by a *Transformer (TF)* element. To calculate the value of the coefficient defining this element, the change in the definition of the flow before and after this element is evaluated. Before the transformer, the flow is in the hydraulics domain, while after, it is in the mechanics domain. The coefficient will be determined by evaluating the required change between both domains. Since the equation that relates both flows is $f_2 = f_1 \cdot r$, where $f_1 = Q_1 = v_1 \cdot A_1$ and $f_2 = v_2$, then,:

$$r = 1/_{A_1} = 1/_{\pi \cdot R^2} \qquad (9)$$

where R is the artery radius.

Clot

Representing the clot is the most complex part of the model. Firstly, the existence of a spring is considered (C port). It measures the force support by the beginning of the clot. Experimental data indicate that the clot begins its movement when this force is equal to 0.01N, from which the value of the constant of this spring is determined by the expression:

$$K = F / q \qquad (10)$$

Additionally a resistance R is positioned. It represents the friction between the clot and the arterial wall. The value of this parameter is variable depending on whether the clot has not begun its movement (static friction) or if it is already in movement (dynamic friction). This value is obtained starting from the Stokes equation.

An inertia is inserted that represents the mass of the clot.

$$I = m \qquad (11)$$

Finally, a spring-damper system is used to ensure that the clot remains at rest while the force existing at its beginning is less than 0.01 N. The spring-damper system is joined to a zero flow source. While the clot does not receive the force of minimum suction, it has a zero speed. However, when it begins its movement, the spring-damper system is cancelled allowing its extraction.

Complete model representation

The implementation of the model required for the simulation is shown and it has been made connecting the different components show previously. The full model is presented divided into three ordered blocks.

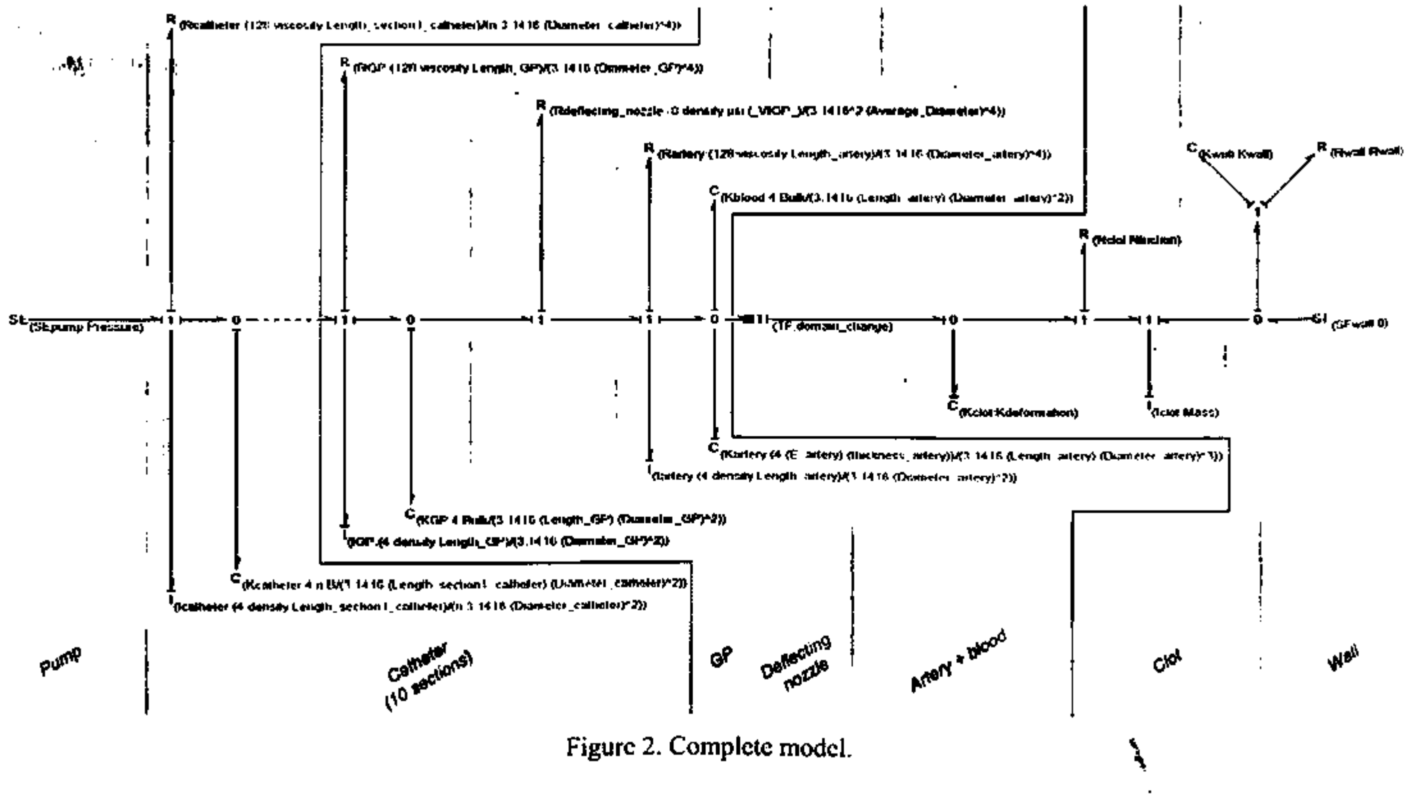

Figure 2. Complete model.

First block shows the junction between the pump with the catheter. Since this is a repeated a section, only first section are shown, each one characterized by the previously described R, C and I ports . Second block includes the 'GP', the loss between the 'GP' and the artery, and the artery and blood. Finally, the third block represents the parameters describing the clot and its movement, which includes the change from hydraulics to mechanics.

MODEL VALIDATION

The object of this study consists in determining the minimum pressure required for the extraction of a blood clot. To do this, by varying the values of the pressure source, the movement of the clot and the time required for its extraction are measured, thereby obtaining the optimum minimum pressure.

To carry out the model validation, the values of the parameters used in the simulation are listed in the following table.

Pressure	0 - [-40, -60] kPa
Blood Viscosity (η)	0.0035 Pa·s
Blood Density (ρ)	1060 kg/m^3
Bulk's coefficient	2200000000 N/m
Catheter length (L)	110 cm
Catheter diameter (D)	0.001 m
'GP' length (L)	0.020 m
'GP' diameter (D)	0.001 m
'GP' thickness (h)	0.0001 m
Artery Young modulus (E)	2800000000N/m
Artery thickness (h)	0.0001 m
Artery diameter (Da)	0.003 m
Artery length (La)	0.003 m
Load loss coefficient (ξ)	0.4
Flow load loss (Q)	flow(I_{GP})
'GP'-artery mean diameter (Dm)	0.002 m
Domain change coefficient (r)	141471.06000
Static friction	0.0000025 N·s/m
Dynamic friction	0.000000025 N·s/m
Clot weight	0.001 kg

Table I. Parameter values

Likewise, by introducing gauges into the model, pressure loss is evaluated through the sections determining where these losses are concentrated. Also evaluated is if the artery possesses the necessary strength to support the pressure to which it is subjected in this operation.

RESULTS

The results obtained are shown after the model simulation using a 40 kPa suction pressure and considering a 5cm long obstructive clot of 1gr. mass.

In figure 3 the evolution of the suction pressure supported by the clot can be observed. It can be seen that it undergoes an increase over time until it reaches a value of 1.41 kPa at 112 seconds. This pressure corresponds to 0.01 N, the point at which the clot starts its movement.

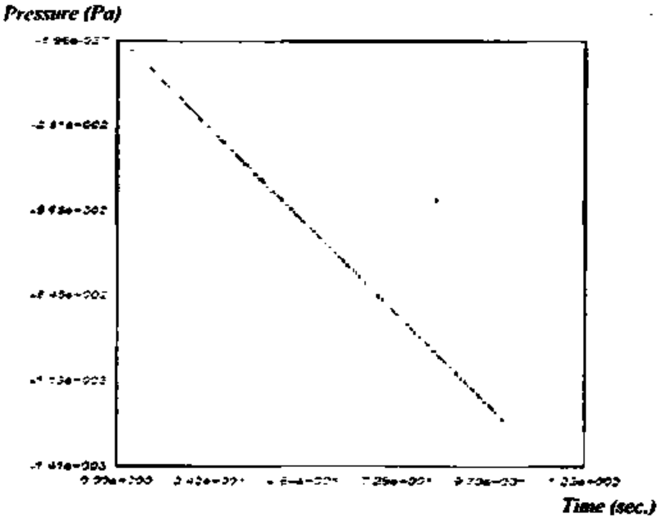

Figure 3. Clot pressure at beginning of movement.

Figure 4 shows clot movement. It can be seen that its movement is zero until 112 seconds, a point at which it reaches a force of 0.01N. From this instant it begins its movement. The clot must travel 3mm through the artery, which is the distance that separates it from the end of the 'GP', in which the clot will be encrusted. It can be observed that the clot needs 114 seconds. to travel this distance.

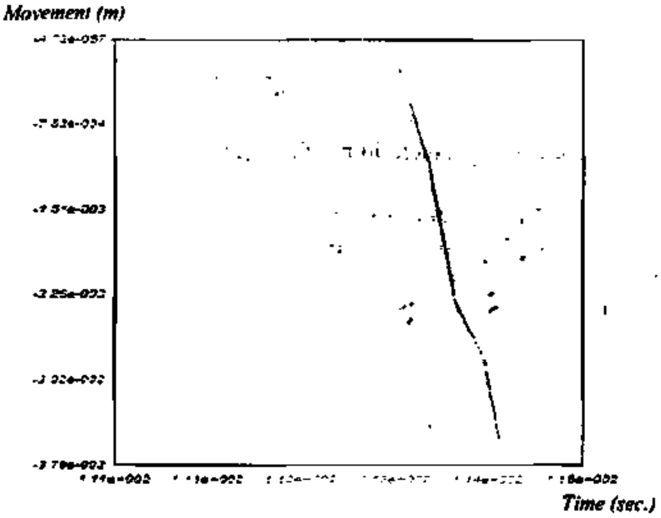

Figure 4. Clot movement.

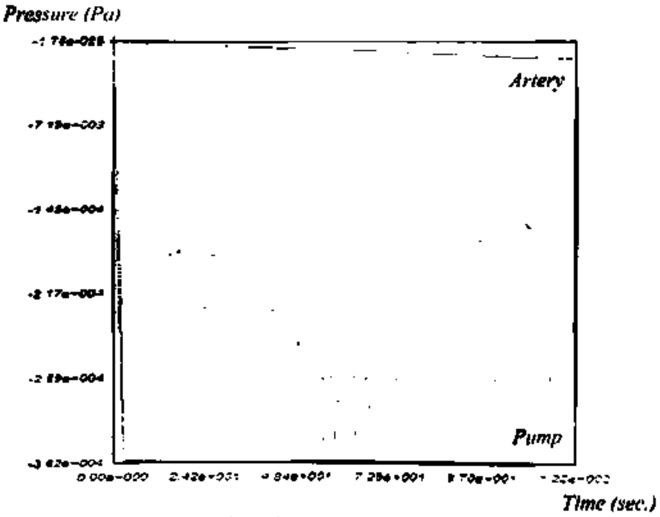

Figure 5. Pressure loss between the pump and the artery.

Figure 5 shows the pressure loss between the pump outlet and the artery.

A significant pressure loss occurs as can be seen in figure 6. It is mainly concentrated in the catheter joining the pump to the 'GP'. As the blood is sucked away, the device progressively loses pressure in each section it flows through due to load loss, blood compressibility and to the blood's own inertia.

Figure 6 shows the loss in pressure in three catheter sections, with a behaviour proportional to the distance being observed.

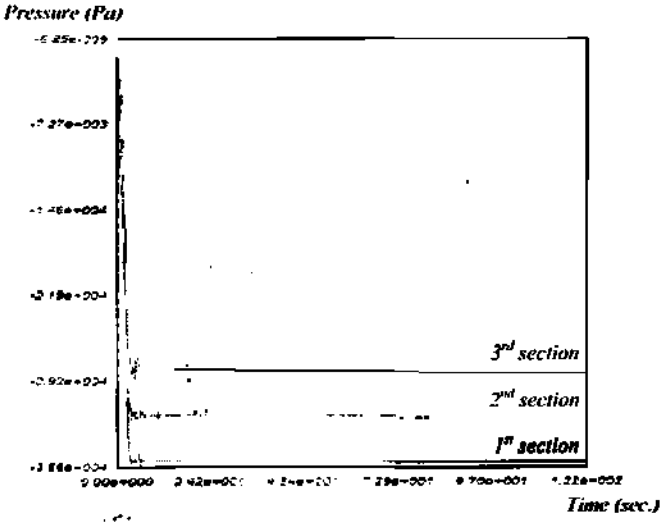

Figure 6. Loss in pressure in the first sections of the catheter.

Finally, in figure 7 the loss in pressure is shown from the pump up to the clot . The proportionality of the loss in pressure is observed in the catheter and its greater magnitude in respect of the loss experienced in the 'GP' and artery.

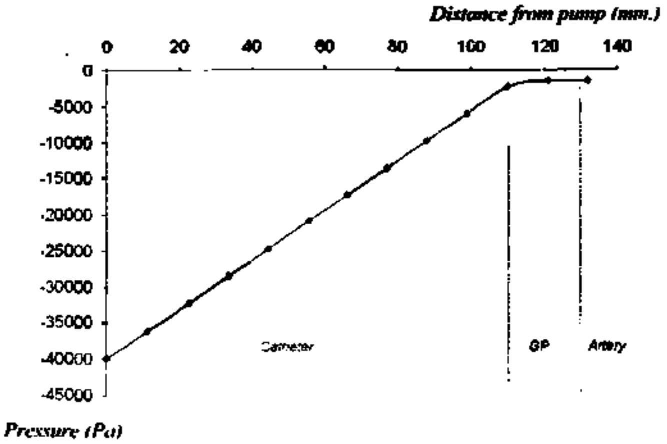

Figure 7. Loss in pressure in the catheter, GP and artery.

Previous figure has been obtained when the clot is moving and it's possible see that the loss of pressure appear mainly in the catheter part due to the length of this component and how the loss of pressure is the same in each division too. After this loss of pressure, the pressure are 99% stabilized in the GP (the less of pressure are lower than in the catheter). Finally a little percentage of the de-pressure pump remove the clot.

CONCLUSIONS

Experimental data indicate that the extraction of the blood clot takes place in an interval of time of [60,120] sec. In figure 4 it can be seen that the simulation produces the extraction in 114 seconds, a figure that is coherent with experience.

The model indicates that a pressure of 40 kPa is enough for the extraction of a 5cm long 1g clot. It would be necessary to check how this pressure evolves by varying the mass and length of the clot.

On the other hand, studies demonstrate that the artery has a resistance of 750 mmHg, which is equivalent to 100 Kpa. In figure 3 it is seen that the pressure in this area rises to 1.41 kPa, there being a wide danger margin for rupture of the artery.

Finally it is shown that an important pressure loss takes place in the catheter joining the pump to the 'GP'. These values obtained can be used to optimize its geometry.

FUTURE WORKS

This work is a first attempt to optimize the operation of the 'GP' device, ensuring its future applicability and compatibility with the experimental data obtained by Dr Pearce. The subsequent lines of work should focus on developing a highly accurate model, the study of different clot sizes and the rechanneling of the blood flow after clot removal.

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BIOGRAPHY

IRENE HIGUERA received her Mechanical Engineering degree from the Technical University of Madrid (Spain) in 2009. She has worked as researcher at U.P.M. since 2008. She developed her research in the field of simulation techniques of multi-domain systems based on bond graph methodology. She has been involved in others tech papers.

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