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FLUID - STRUCTURE MODELING OF A VENTRICULAR CARDIAC ASSISTANCE DEVICE OPERATED IN A BIOMECHANIC FORM

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1.INTRODUCTION

The cardio-insuffiency is one of the diseases that has reported one of the greatest mortality rates in the word. The heart transplant is one of the procedures that has reached very good results, but, at the same time, is very limited due to both the escarce number of available donors and its high operation costs. Other current possibilities of artificial heart assistance has not been the most adecuated ones, due to its high risk of infection through catheters and other conections. A new solution altenative is proposed: artificial pumping device with biomechanical action of the wide dorsal muscle and in conection with the left ventricule in series. The model must give the necessary ejection to improve the output cardiac. There is identified in this system the interaction of the Muscle – device and hemodynamic fluid. Some simulations of the behavior were realized Fluid - structure of the dynamics of the blood fluid and of the ventricular device. The results obtained of this structural design delivered result of pressures and velocity to the output of the device that go in conformity with the required values in order that a good functioning is achieved at the artificial device , the heart and the circulatory system. It's fluid-structure problem that represents an internal wall composed of a fine flexible membrane, for the device, and that is the blood in movement because of a few hemodynamic fluid tensions, with besides a few edge conditions of the action of the dorsal muscle.

2. FLUID-STRUCTURE MATHEMATICAL MODEL

Like Considers a representative model hiper-elastic of device in 2D, put under pressure on its external superficial layer. Model that counts on four defined regions affluent: entrance, $\mathbf{\Gamma}^1$, exit, $\mathbf{\Gamma}^2$, superior $\mathbf{\Gamma}^S$,e inferior $\mathbf{\Gamma}^I$. The fluid enters the subsystem by asistolic action of the heart, through $\mathbf{\Gamma}^1$ and one lodges in the enclosure that is closed momentarily in $\mathbf{\Gamma}^2$. Soon the action of the wide dorsal muscle takes place on $\mathbf{\Gamma}^S$. After it manages to evacuate great part of the fluid by $\mathbf{\Gamma}^2$. With the system of single control of valves it's possible that the fluid circulates of unidirectional way. the region is considered $\mathbf{\Gamma}^I$ in all the mathematical treatment. $\forall t \in I$, a dislocation will exist of $\mathbf{\Gamma}_t^S$, represented breakup will exist of η respect to the reference $\mathbf{\Gamma}_0^S$. Guaranteeing the states of continuity of the stress principle of action and reaction and the perfect adhesion of the fluid to the structure. Also initial conditions will be considered for, Ω_0 y Γ_0^S. Effects of despicable movements for the sub-regions $\mathbf{\Gamma}^1$ y $\mathbf{\Gamma}^2$ and the device of ventricular cardiac. Be a mappings family A_t for $t \in I$ defined with $A_t : \Omega_0 \subset R^n \longrightarrow \Omega_f \subset R^n \quad \Rightarrow x(Y,t) = A_t(Y)$. The vectorial spaces are defined formally that are required within the subsystem: The global equation of coupling for the problem fluid-structure, the Neumann case and the case of Dirichlet: : The condition of coupling in the interphase $u \circ A_t = \dot{\eta}\boldsymbol{e_r}$ en Γ_0^S, guaranteeing the continuity of speed between the fluid and structure. The condition of the construction of the mapping $\int_{\Omega_0} \nabla A_t : \nabla \boldsymbol{z} d\Omega = 0$ in the dominion Ω_0 y $A_t = \omega_\eta$ and in the contour $\partial\Omega_0$ for $\forall t > 0$. In addition the speed to the dominion of the mapping is represented by , in a dominion Ω_t. The initial conditions $\vec{u} = \vec{u}_0, \eta = \eta_0$ for $t = 0$ en Ω_0, y $\dot{\eta}\boldsymbol{e}_r = \vec{u}_o$ for $t = 0$ en Γ_0^S.The coupled equation fluid-structure for neumann case, $\forall(\mathbf{v}, \xi) \in \boldsymbol{V}(t)$ y $\forall q \in Q(\Omega_t)$; and for the Dirichlet case, $\forall(\mathbf{v}, \xi) \in V_0(t)$, $\forall q \in Q(\Omega_t)$:

$$\int_{\Omega_t} div\ uqd\Omega = 0 \tag{1}$$

$$\begin{aligned}&\int_{\Gamma_0^S} \rho_m h \frac{\partial^2 \eta_r}{\partial t^2}\xi dz + kGh\,\xi \frac{\partial \eta}{\partial z}_{\substack{z=0\\z=L}} + \int_{\Gamma_0^S} kGh \frac{\partial \eta}{\partial z}\frac{\partial \xi}{\partial z}dz + \int_{\Gamma_0^S}\frac{Eh}{1-\nu^2}\frac{\eta_r}{R_0^2}\xi dz \\ &+ \frac{\partial^2 \eta_r}{\partial t \partial z}\xi_{\substack{z=0\\z=L}} + \int_{\Gamma_0^S}\frac{\partial^2 \eta_r}{\partial z \partial t}\frac{\partial \xi}{\partial z}dz + \int_{\Gamma_0^S} p_{ext}\xi dz + \int_{\Omega_t}\rho \partial \vec{u}_t \cdot \mathbf{v} d\Omega + \int_{\Omega_t} div\left[(\vec{u}-\vec{w})\otimes u\right]\cdot \mathbf{v} \\ &- \int_{\Omega_t} div(\mathbf{v}p) - 2\mu\int_{\Omega_t} D(u):\nabla \mathbf{v} d\Omega - \int_{\Gamma^2_{Neumann}} 2\mu\left[\underbrace{D(u)n}_{\equiv \mathbf{0}}\right]\cdot \mathbf{v} = \int_{\Omega_t}\vec{F}\cdot \mathbf{v} d\Omega + \int_{\Gamma_0^S} p_{Musc}\xi dz\end{aligned} \tag{2}$$

and Dirichlet:

$$\begin{aligned}&\int_{\Gamma_0^S} \rho_m h \frac{\partial^2 \eta_r}{\partial t^2}\xi dz + \int_{\Gamma_0^S} kGh \frac{\partial \eta}{\partial z}\frac{\partial \xi}{\partial z}dz + \int_{\Gamma_0^S}\frac{Eh}{1-\nu^2}\frac{\eta_r}{R_0^2}\xi dz \\ &+ \int_{\Gamma_0^S}\frac{\partial^2 \eta_r}{\partial z \partial t}\frac{\partial \xi}{\partial z}dz + \int_{\Gamma_0^S} p_{ext}\xi dz + \int_{\Omega_t}\rho \partial \vec{u}_t \cdot \mathbf{v} d\Omega + \int_{\Omega_t} div\left[(\vec{u}-\vec{w})\otimes u\right]\cdot \mathbf{v} \\ &- \int_{\Omega_t} div(\mathbf{v}p) - 2\mu\int_{\Omega_t} D(u):\nabla \mathbf{v} d\Omega = \int_{\Omega_t}\vec{F}\cdot \mathbf{v} d\Omega + \int_{\Gamma_0^S} p_{Musc}\xi dz \\ &\int_{\Omega_t} div\ uqd\Omega = 0\end{aligned} \tag{3}$$

3. MODELING OF A CARDIAC VENTRICULAR ASSISTANCE DEVICE

The constructed model can be occupied by a maximum volume of 100 millilitres of blood. It is selected like permissible value of viscosity for the blood: 3.0 centiposes, density: 1,04 grams by millilitre and pressure of preload: 10mmHg. Can be used for I module longitudinal of the material, poisson radio $\nu = 0.49$, like constants of Mooney Rivlin, a value of 0.2×10^{6} . On the surface considers Γ^{I} surface it's considered null displacements, like in the subregions Γ^1 y Γ^2 . See fig. 1 :

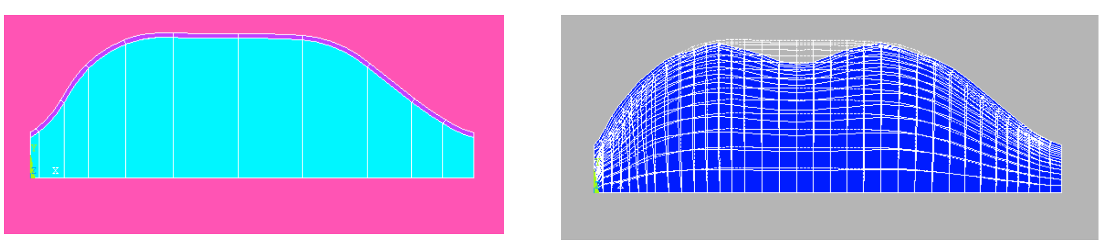

Figure 1 . Bidimensional geometry of Ventricular attendance device

It's considered as the current hemodinamyc within the ventricular cardiac device like a

Newtonian fluid. It will be assumed for the conditions of edge of the dominion of the hemodinamyc fluid Ω_f plus is constituted with limits of border $\partial\Omega_f$ sufficiently regulars. Other interaction conditions of both subsystems: coincidence of the fluid-structure interface, principles of action and reaction in the interface and Kinematics conditions was utilized for the mapping of quadrilateral structural element of reference 141. See fig 2.

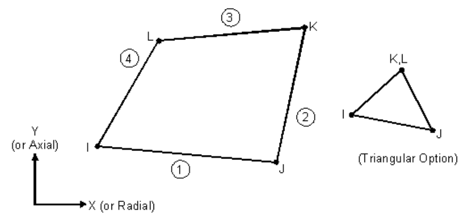

Figure 2 . Quadrilateral structural element of reference 141

4. RESULTS OF THE VENTRICULAR ASSISTANCE DEVICE SIMULATION

The The field of velocity indicate that indeed the greater speed takes place in the exit of the model because of the external pressures. velocity were achieved with forces of 64 grsF; results that were found adjusted of the range work necesary. See fig. 3 .

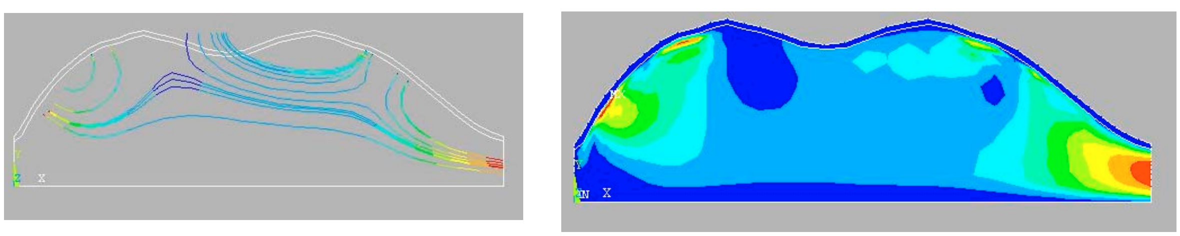

Figure 3. Results of speeds and hematíes particles

CONCLUSIONS

It is observed that the displacement of the mapping happens mainly in the normal direction of the subregion, but as well as in the walls nonloaded to muscular pressure they under go an expantion by effects of the internal pressure of the fluid acting on the inner walls Another part of the fluid leaves by the subregion. It is observed that the geometric form quickly facilitates the increase of velocity in the zone of the exit. On the inner walls other speeds take place which are those that guarantee the states of continuity of the stress: principle of action and reaction, and the perfect adhesion of the fluid structure. The displacement of hematíes particles can be determined from streamlines directly acting on the inner walls of the model in and another marking a route of complementary logout. The bidimensional model is considered within the interpretation of the phenomenon fluid-structure sufficiently representative. The numerical implementation in two dimensions gives results that generate certain security of which it is tried to reach. With the purpose of having a greater clarity of the action of the muscle on the surface of the ventricular pump, it's had a model in the three dimensions. This design as it already have been mentioned, has a great particularity and it allows that the smaller power of the skeletal muscle on the surface in contact, maximums can be reached displacements of drained and greater velocity of the hemodinamyc fluid. This new condition prohangs to that the skeletal muscle quickly, does not reach a process of fatigue in the time. The reason of this special characteristic is that its cross-sectional form is not completely uniform, but that in addition has two located axial-asymmetric axes under certain geometric strategy, on which it is allowed to guarantee the mentioned particularity previously. The geometric form does not offer great resistance to the contraction of the dorsal muscle. The arcs that constitute initially a cylindrical surface that is smoothed by means of other arcs joined in the inferior part of the model. Its base in flat form allows to comply him in any rough and irregular surface. The areas in form of semicylindrical plate allow an effective contact of the muscle against the surface without irrigations of instabilities of the global model. The longitudinally seen model presents/displays of a form similar to a cone that facilitates the increase of speed with respect to its diametrical variation. Is possible to be observed that the model is consistent with the studies previously made in the mathematical and numerical model, since its geometric principle is the same principle implemented in the study in two dimensions. See fig. 4.

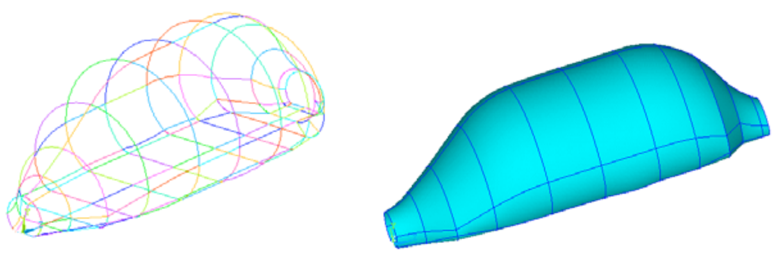

Figure 4 . Model three dimensions

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