Blood Coagulation Time Dependence of the Polymeric-Surface Contacting Area on a Left Ventricle Assist Device Implanted In Calves

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Abstract. A miniaturized artificial heart is being tested in our laboratories, the Auxiliary Total Artificial Heart (ATAH). This device can be implanted with two artificial ventricles, as Total Artificial Heart (TAH), or with only one pumping chamber, as Left Ventricle Assist Device (LVAD). Five *In Vivo* tests were performed in calves. The device was implanted as LVAD in their abdominal cavities in series with their native hearts. The implantation were performed connecting a flexible polymeric graft from the left ventricle apex to the LDAV inlet port and another polymeric graft from the LVAD outlet port to the descending aorta. The LVAD internal blood-contacting surface are also made of polymeric materials. During all the tests, heparin was administrated to control the blood Activated Coagulation Time (ACT). A strong relationship between the polymeric area and the heparin dosage was observed. Also, a relationship between the blood flow through the LVAD and the heparin infusion flow was observed. Besides, with the LVAD helping the natural heart to obtain the appropriate blood flow, the native heart beat decreased. It was observed that the LVAD and the native heart worked synchronously, with the left artificial ventricle and the natural heart contracting alternatively.

Keywords. Artificial heart, Ventricle assist device, Blood, Polymers, Polyurethane.

1. Introduction

Dr. Michael DeBakey developed the first continuous blood pump for transfusion in 1934 (DeBakey, 1934). Since that time many research groups have developed a great variety of blood pump systems. Several years later, with the surgical technique improvements, studies for new generation pulsatile ventricular assist devices (VAD) were initiated and the first clinical implantation was performed by Dr. Domingo Liotta in 1961 (Liotta, 1963), followed by Dr. DeBakey in 1963 (DeBakey, 1971). However, for more severe clinical cases new cardiac prostheses are necessary and had to be developed. In 1965, Dr. Yukihiko Nosé demonstrated that it was possible to implant a Total Artificial Heart (TAH) inside the pericardial sac of calves (Nosé, 1965). Dr. Denton Cooley performed the first TAH clinical implantation, in 1969 (Cooley, 1969). Nevertheless, in 1983 at Utah University, Dr. Devries performed the first successful TAH implantation because the patient, even in an unstable condition, survived for 112 days (Devries, 1984). That fact motivated the leading research groups to concentrate their efforts to develop their own TAH project. Presently, various VAD and TAH designs are being developed. One of the most difficult problems encountered by those groups is the device dimension (Jarvik, 1986; Shiono, 1991). Some of the commercially available pulsatile VADs are not readily implantable into the thoracic cavity due to size limitation. Most of the TAHs, which are under development, have dimensions requiring the removal of the patients' native heart.

The TAH research groups experienced many other difficulties. One such problem is the TAH control system; this fact is aggravated by the removal of the native heart. Thus, eliminating the natural cardiac output control performed through the mechanoreceptors and chemoreceptors, and also, the right-left flow balance which is helped by the natural heart (Michelini, 1986; Chalmers, 1991; Fukamachi, 1993)

2. Materials and Methods

A miniaturized artificial heart is being tested in our laboratories, the Auxiliary Total Artificial Heart (ATAH). The ATAH is an electromechanically driven artificial heart with reduced dimensions (outer diameter is 85 mm and the thickness is 65 mm), being able to be implanted in the right thoracic cavity (in parallel) or in the abdominal cavity (in series) of an average human patient without removing the natural heart or the heart neuro-humoral inherent control mechanism for the arterial pressure. When in the abdominal cavity (in series), the device can be implanted with two artificial ventricles (as Total Artificial Heart - TAH) or with only one pumping chamber (Left Ventricle Assist Device – LVAD). Figure (1) shows a schematic drawing of the implantation options.

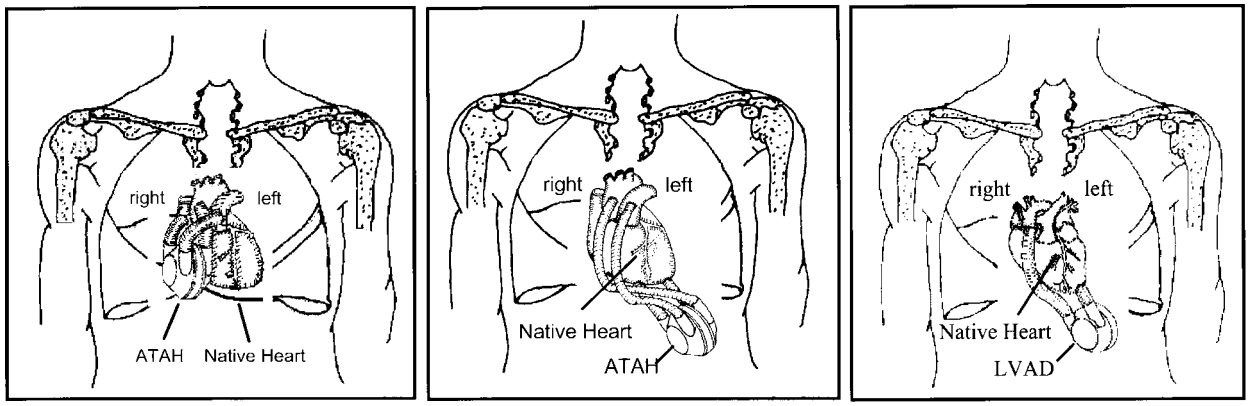

Figure 1. Schematic Drawing showing the implantation options for the Auxiliary Total Artificial Heart (ATAH) in parallel (left) or in series (center) and with only one pumping chamber as Left Ventricle Assist Device (LVAD) (right).

The ATAH's beating frequency is regulated through the change of the left preload, based on Frank-Starling's law, assisting the native heart in obtaining adequate blood flow. The ATAH left and right stroke volumes are 38 ml and 34 ml, respectively, giving approximately, 5 L/min of cardiac output at 160 bpm (Andrade, 1999).

This device is an electromechanical pulsatile blood pump with left and right chambers. A brushless direct current (DC) motor which is fixed in a metallic centerpiece provides the actuation. A mechanical actuator, the planetary roller screw, converts the motor rotation into a rectilinear motion that advances the left and right diaphragms. A support plate with three stabilizer rods is welded to one edge of the roller screw to avoid rotation. The diaphragms are adhered to conical pusher plates. The left and right housings are made of an epoxy resin.

The filling and ejection phases for the right and left chambers alternate. Three magnetic Hall effect sensors are installed in the centerpiece, two magnets are fixed to the stabilizer rods and one to the left diaphragm (Fig. 2).

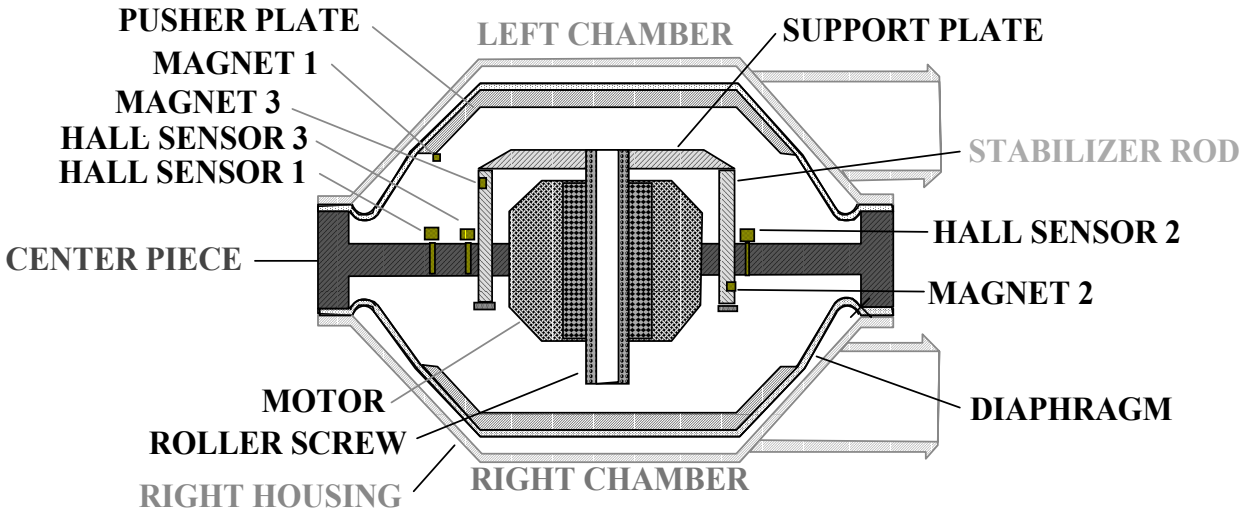

Figure 2- Schematic drawing showing the assembly of the ATAH mechanical parts and the position for three Hall effect sensors and the respective magnets.

When the left auxiliary ventricle is completely filled, Hall sensor #1 detects the magnet #1, at this moment, the motor is turned on and the left ejection phase commences. When sensor #2 detects the proximity of magnet #2, the motor inverts its rotation and the support plate moves down toward the centerpiece. The left pusher plate detaches from the support plate allowing the left chamber passive filling. The Hall sensor #3 with magnet #3 detects the down position of the support plate and the motor is turned off waiting for the down position of the left diaphragm to start a new cycle. This device operates in left master alternate mode (LMA), with the left auxiliary ventricle, as the master, setting the ATAH pumping rate and the right beating alternatively with a fill-limited flow (Ohashi, 1997). The ATAH frequency can be constant, operating in a fixed rate mode (FR) or variable when in a variable rate mode (VR) (Andrade, 1999). In FR, the waiting time is pre-fixed by the ATAH electronic controller (Ohashi, 1997). In VR, the left ejection phase

commences when the Hall sensor (sensor #1 – Fig. 2) detects a complete filling of the auxiliary left ventricle. Therefore, the ATAH pulse rate depends on the left chamber filling time and is dictated by the left preload.

Five *In Vivo* tests were performed in calves with 80 ± 5 Kg of weight. The device was implanted as LVAD in their abdominal cavities in series with their native hearts to verify the device performance and operating functions. The implantation were performed connecting a flexible polymeric graft from the left ventricle apex to the artificial ventricle inlet port. From the LVAD outlet port, the blood was pumped to the descending aorta through another polymeric graft. Inside the LVAD inlet port was a biological prosthetic valve with 23 mm of diameter and inside the outlet port was a 21 mm artificial valve. The LVAD internal blood-contacting surface are also made of polymeric materials. The pumping diaphragm is made of polyurethane rubber and the housing is made of acrylic resin. During all the tests, heparin was administrated to control the blood Activated Coagulation Time (ACT). The ACT was maintained in 350 ± 50 s by adjusting the heparin (anticoagulant) infusion flow.

3. Results and Conclusions

Important data were obtained with those tests. A strong relationship between the polymeric area and the heparin dosage was observed. The polymeric materials initiate the blood coagulation process that must be reversed by the anticoagulant (heparin). Also, a relationship between the blood flow through the LVAD and the heparin infusion flow was observed. Fig. (3) shows the relationship between Activated Coagulation Time (ACT) and infusion flow of heparin (2000u/250ml) for 4 calves.

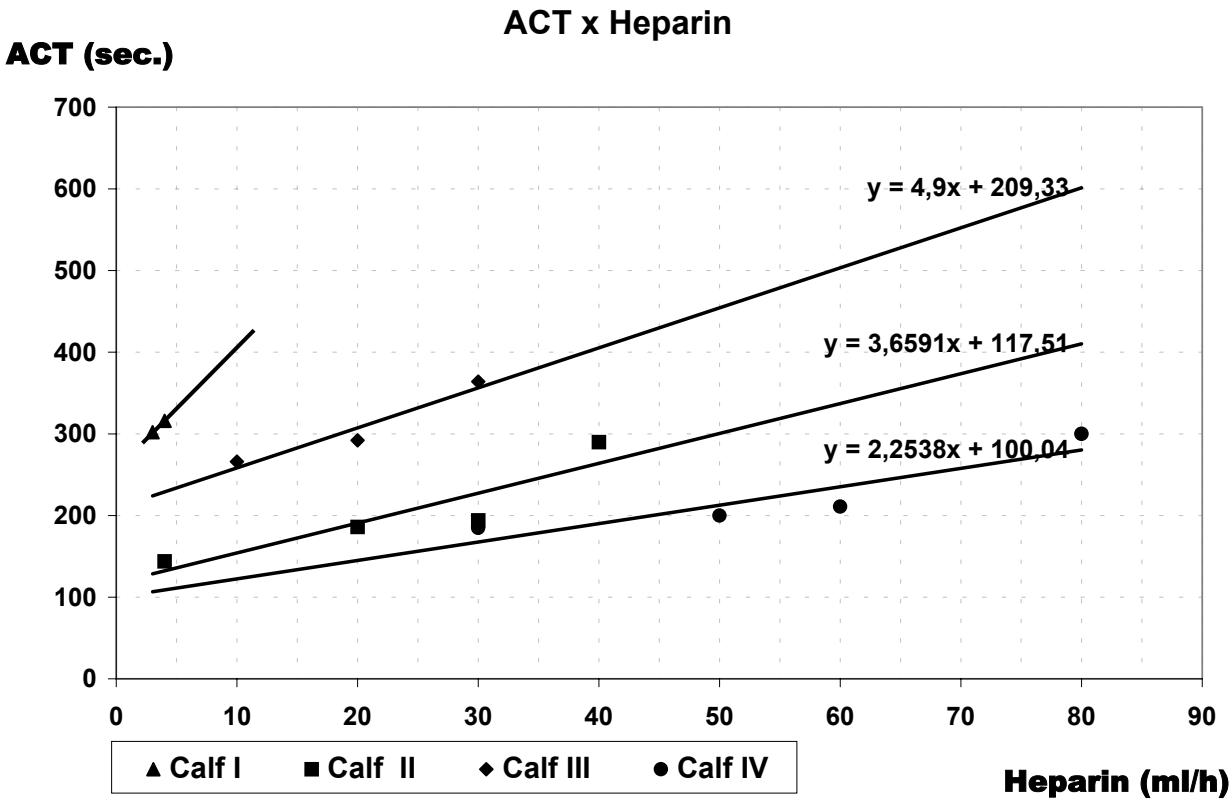

Figure 3. Relationship between ACT and Heparin infusion

In calves II, III and IV, the device was implanted using polymeric tubes with 300 mm of length and 15 mm of diameter. However, in calf I, the device was implanted with no polymeric tubes, we used grafts with a layer f biological tissue. Comparing the results between calf I and the others, it is easy to observe that the calves with polymeric tube needed higher volumes of heparin to annul the coagulation activation due to the contact between the blood and this polymeric surface. During the experiments, the protocols fixed a optimum Activated Coagulation Time of 300 sec. Calf I had infusion of 3 or 4 ml/h of heparin solution to maintain the ACT around 300 sec. However, to achieve this ACT, calf II had 50 ml/h, calf II had 20 ml/h and calf IV had 80 ml/h.

Another important data can be observed comparing all calves with polymeric tubes (calves II, III and IV). They show different sensibility to the heparin. Calf III had less heparin infusion (20 ml/h to achieve ACT of 300 sec.) and the blood flow through the polymeric tube was 2 L/min. Calf IV has higher heparin infusion (80 ml/h to achieve ACT of

300 sec.) and the flow through the polymeric tube was 5 L/min. Calf III had 40 ml/h of heparin and the blood flow was 4L/min. Those data suggest a strong relationship between blood flow trough the polymeric tube and the activation of blood coagulation. It is possible to speculate that higher flow means more blood cells in contact to polymeric surface and those cells have the coagulation process activated and needs more heparin to inhibit and reverse this process.

In addiction, with these experiments it was possible to observe the LVAD behavior. With the LVAD helping the natural heart to obtain the appropriate blood flow, the native heart beat decreased. It was observed that the LVAD and the native heart worked synchronously, with the left artificial ventricle and the natural heart contracting alternatively. Thereby, it is possible to predict that the ATAH will help a sick heart to recover its pumping function. With the ATAH and the natural heart working simultaneously, the control system is simplified and the surgical risks are reduced.

4. Acknowledgement

We would like to acknowledge CNPq and FAPES for partially support our project

5. References

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